



Relationships between lumbar inter-vertebral kinematics
and paraspinal myoelectric activity during sagittal flexion:
a quantitative fluoroscopy and surface electromyography
study

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Abstract

Title. Relationships between lumbar inter-vertebral kinematics and paraspinal myoelectric activity during sagittal flexion: a quantitative fluoroscopy and surface electromyography study

Introduction.

Previous investigations that have attempted to relate mechanical parameters to NSLBP groups are often contradictory of each other, and currently clear mechanical markers for LBP remain elusive. In order to move forward in this area, it may be necessary to take a step back, and improve understanding of 'normal' spinal biomechanics (i.e. in low back pain free populations). Indeed, Peach et al. (1998) stated "*By knowing what is "normal" and what is "abnormal" it may be possible to provide objective evaluation of rehabilitation protocols, and possibly classify different low back pathologies*" (Peach et al. 1998). Therefore, an improved understanding of biomechanical behaviours in groups of back pain free people is desirable, particularly at an inter-vertebral level, an area where clear knowledge gaps still exist.

Control of the spine during voluntary movement requires finely-tuned coordination of numerous trunk muscles. This dynamic control is believed to be achieved via communication between three sub-systems, the passive (vertebrae, discs and ligaments), the active (muscles and tendons) and the control (central and peripheral nervous system) systems. Investigating the interplay between these sub-systems however is difficult, as the spine is a complex structure with a hidden kinematic chain. Quantitative fluoroscopy (QF) is an imaging technology capable of measuring continuous spinal kinematics at the inter-vertebral level, and surface electromyography (sEMG) provides a non-invasive means of objectively quantifying muscle activity. This study used QF and sEMG technologies concurrently to investigate relationships between and amongst lumbar kinematic (QF determined) and muscle activity (sEMG determined) variables, during weight-bearing active forward flexion. This was the first time such technologies have been combined to investigate the biomechanics of the lumbar spine *in vivo*. An improved understanding of normal lumbar kinematic and myoelectric behaviour, will assist in the interpretation of what is abnormal in terms of inter-vertebral spinal mechanics.

Methods. Contemporaneous lumbar sEMG and QF motion sequences were recorded during controlled active flexion of 60° in 20 males with no history of low back pain in the previous year. Electrodes were placed adjacent to the spinous processes of T9, L2 and L5 bilaterally, to record the myoelectric activity of the thoracic and lumbar erector spinae (TES and LES) and lumbar multifidus (LMU) respectively. QF was used concurrently to measure the maximum inter-vertebral rotation during flexion (IV-RoMmax) and initial attainment rate for the inter-vertebral levels between L2 and S1, as well as each participant's lordotic angle. The sEMG amplitude data were expressed as a percentage of a sub-maximal voluntary contraction (sMVC). Ratios were calculated between the mean sEMG amplitudes of all three muscles examined. Each flexion cycle was also divided into five epochs, and the changes in mean sEMG amplitude

between epochs were calculated. This was repeated to determine changes between all epochs for each muscle group. Relationships between IV-RoMmax and all other kinematic, morphological (i.e. lordosis) and muscle activity variables were determined using correlation coefficients, and simple linear regression was used to determine the effects of any significant relationships. The reliability and agreement of the IV-RoMmax, initial attainment rate, and normalised RMS sEMG measurements were also assessed.

Results. The reliability and agreement of IV-RoMmax, initial attainment rate and sEMG amplitude measurements were high. There were significant correlations between the IV-RoMmax at specific levels and the IV-RoMmax at other lumbar motion segments ($r = -0.64$ to 0.65), lordosis ($r = -0.52$ to 0.54), initial attainment rate (-0.64 to 0.73), sEMG amplitude ratios ($r = -0.53$) and sEMG amplitude changes ($r = -0.48$ to 0.59). Simple linear regression analysis of all significant relationships showed that these variables predict between 18% and 42% of the variance in IV-RoMmax.

Conclusion. The study found moderately strong relationships between kinematic, morphological and muscle activity amplitude variables and the IV-RoMmax of lumbar motion segments. The effects of individual parameters, when combined, may be important when such inter-vertebral levels are considered to be sources of pain generation or targets for therapy. This is an important consideration for future non-specific low back pain (NSLBP) research, as any attempts to associate these parameters with low back pain (LBP), should also now take in to account the normal biomechanical behaviour of an individual's lumbar spine. Indeed, consideration should be given to the interactions that exists between such parameters, and they should not be considered in isolation. Multivariate investigations in larger samples are warranted to determine the relative independent contribution of these variables to the IV-RoMmax.

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This study was integrated into another ongoing study (Section 4.2.1). Therefore certain elements of the data collection process were conducted by others. The collection of data required the concurrent operation of a fluoroscope (Dr Fiona Mellor), a motion table (Professor Alan Breen) and the muscle activity recording equipment (Author). The study design, image analysis, data processing, statistical analysis and manuscript were the responsibility of the author. It should also be noted that material from this thesis has been published in open access journals (Appendices Q and R).

Chapter 1: Background

1.1 Introduction

A recent systematic review estimated the global lifetime prevalence of low back pain (LBP) to be approximately 39% (Hoy et al. 2012). According to Waddell (2005), around 85-95% of LBP falls into the non-specific category, meaning the majority of cases have no known cause (Waddell 2005), but numerous possible ones (Deyo et al. 2014; Kent 2004). In acknowledgement of the multifactorial nature of the problem, over the last few decades LBP researchers and clinicians have widely adopted Waddell's biopsychosocial model (Waddell 1987), which considers both biological and psychosocial components of the problem. Perhaps due to a perceived lack of progress in linking NSLBP to biological causes, recent years have seen an apparent emphasis placed on the latter, raising concerns that the potential importance of biological elements may be being neglected (Hancock et al. 2011). The biological component is itself multifactorial, and has been proposed to comprise of mechanical, chemical and central sensitisation constituents (Breen 2013). A challenge therefore is how to define, disaggregate and objectively differentiate those biological factors that generate and sustain non-specific LBP (NSLBP) in order to use them to facilitate clinical decision making in the care of patients. For physical therapists such as physiotherapists, osteopaths and chiropractors, there is an aspiration to determine mechanical causes of LBP, and this is reflected in their research efforts (O'sullivan 2005; Sahrman 2002; Van Dillen et al. 2003).

1.2 A requirement for enhanced functional assessment?

Even with a focus on mechanical causes, due to the large number of biomechanical factors that are co-ordinated to perform a given task, identifying the main parameters that discriminate mechanical LBP patients from healthy controls is challenging (Todorov and Jordan 2002), a task further complicated by the inherent heterogeneity in NSLBP populations, and methods of data collection (Leboeuf-Yde and Manniche 2001). This is perhaps a reason why many contradictions can be found in the literature, with mechanical factors such as spinal range of motion (ROM) (Brownhill 2010; Dankaerts 2009; Iguchi 2004; Kanemura 2009; McGregor et al. 1995, 1997; Taghipour-Darzi et al. 2012; Teyhen 2007; Triano 1987), muscle activity (Ahern et al. 1988; Alexiev 1994; Arena et al. 1991; Cassisi et al. 1993; Cram and Engstrom 1986; Dankaerts 2006, 2009; Kravitz et al. 1981; Lariviere et al. 2000; Sanchez-Zuriaga et al. 2015) muscle activity ratios (Reeves et al. 2006; Van Dieen 2003) and lordosis (Been and Kalichman 2014; Sarikaya et al.

2007; Tuzun et al. 1999) all being shown to exhibit either no association or contradictory relationships with LBP. The majority of these studies take a reductionist approach, in which emphasis is placed on specific components of the spinal system. This type of approach often fails to demonstrate how changes in specific factors affect the behaviour of the system as a whole, leading Reeves and Cholewicki (2010) to suggest that there is a “need to move away from documenting “differences” (between LBP groups and healthy controls) to “understanding” the effect these differences have on the spine” (Reeves and Cholewicki 2010). There is a perceived requirement therefore for further understanding of the integration of different components within the spinal control system. For example, Sanchez-Zuriaga et al. (2015) suggest that there are only subtle differences between mechanical NSLBP groups and healthy controls in terms of paraspinal muscle activity and regional lumbar movement (Sanchez-Zuriaga et al. 2015). This could mean that either muscle activation strategies have no effect on the motion, or that the motion differences are at individual inter-vertebral levels and remain undetected. If there is an increase in paraspinal muscle activity in recurrent LBP patient groups during a movement such as sagittal flexion, but no difference in global range of motion (RoM), then the proportional share of RoM may have shifted between levels, with the observed motor control strategy primarily influencing particular motion segments.

The primary role of the paraspinal muscles during flexion is to resist forces applied to the inter-vertebral linkage, with local and global groups activating synergistically to provide control (Bergmark 1989; Bogduk 2012). If restraint mechanisms are such that motion is restricted at a specific inter-vertebral level, it is likely that the reduced movement will be compensated for elsewhere, be this at other lumbar levels (Lee and Langrana 1984), or in the thoracic spine or hips and pelvis (McGregor 2002; Mehta et al. 2012; Rothenfluh et al. 2015). It would be beneficial therefore, when attempting to understand the relationships between functional impairments and LBP that multiple inter-vertebral levels are assessed both in terms of kinematics and associated muscle activity. This has generally not been possible for LBP clinicians, however aspirations to attribute mechanical NSLBP to symptomatic spinal segments remain (Kulig et al. 2007), leading to the development of systems that classify LBP groups in terms of motor control impairments (Dankaerts 2007; O'sullivan 2005). These approaches subgroup LBP patients based on criteria associated with the primary direction of pain provocation (e.g. pain reproduced during flexion or extension activities), without knowledge of its nociceptive source, a strategy that perhaps suffers from ambiguity of cause and effect. There

is therefore a compelling argument for wanting to know more about the factors that influence restraint at a segmental level.

If mechanical problems are thought to be a major contributing factor to LBP, then better *in vivo* objective measures are needed, to help target specific anatomy and biomechanics in management. To make progress in this area, studies should advance understanding of segment specific biomechanics. To do this, they will need to record multi-segmental biomechanical information.

1.3 Investigating spinal control

Optimal control of the lumbar spine during voluntary trunk bending requires the co-ordination of a number of muscles (Reeves 2007), and the co-ordinated participation of multiple vertebral motion segments, whose contributions are a function of their own mechanical properties (Sahrmann 2002). This dynamic control is believed to be modulated by interactions between three sub-systems, the passive (vertebrae, discs, and ligaments), the active (muscles and tendons), and the control (central nervous system and nerves) systems (Panjabi 1992a, 1992b). It is theorised that a dysfunction in any subsystem may lead to a response from another to compensate. Therefore the ability to study the contemporaneous performance of these subsystems would be of value. Investigating the interplay between sub-systems however is difficult, as the living spine is a complex structure; and a hidden kinematic chain. Several different technologies are therefore typically applied, each with its own limitations.

In order to directly investigate the passive and active sub-systems of the spine, there have been many efforts to concurrently measure spinal kinematics and muscle activity (Burnett 2004; Claus et al. 2009; Hashemirad et al. 2009; Kaigle 1998; Kim et al. 2013; Peach et al. 1998; Sanchez-Zuriaga et al. 2015). The majority of these studies have used surface electromyography combined with skin surface kinematic measurement techniques such as 3-Space Fastrak, Polhemus inc. (Burnett et al. 2004; Dankaerts et al. 2009), Isotrak, Polhemus inc. (McGill et al. 1997; Peach et al. 1998), or cameras (Sanchez-Zuriaga et al. 2015; Kim et al. 2013; Hashemirad et al. 2009). Whilst skin surface markers can be used to measure inter-vertebral motion, such measurements do not typically have good reliability due to skin movement artifacts (Cerveri et al. 2005; Zhang 2003), and are therefore generally limited to the investigation of gross spinal motion. To include segmental data therefore usually requires more invasive techniques such as x-rays (Ogston 1986a; Percy 1984b) or fluoroscopy (Ahmadi et al. 2009; Breen A.C. et al. 2012; Du Rose and Breen 2016a; Du Rose and Breen 2016b; Mellor 2014; Teyhen et al. 2007; Wong

2006), or the surgical insertion of intra-osseous pins (Kaigle et al. 1998). In this way Kaigle et al. (1998) investigated the reduction in lumbar muscular activity at the point of full flexion (Flexion Relaxation Phenomenon (FRP)) along with spine kinematics at an inter-vertebral level (Kaigle et al. 1998). However, only single motion segments were considered, and the angular ranges at different inter-vertebral levels were pooled. Electromyography (EMG) was also only recorded from one level (e.g., lumbar longissimus thoracis). Therefore, although inter-segmental kinematic data were recorded, no discriminative insight was obtained regarding multi-level interactions.

Advances in automated image motion analysis, radiation dose reduction, and digital imaging however, now enable the use of fluoroscopy to measure multi-level spine motion (Breen A.C. et al. 2012), which has been demonstrated to be an accurate and reliable 2D method (Mellor et al. 2014, Teyhen et al. 2007, Yeager et al. 2014). Whilst recent technological advances do enable the acquisition of 3D kinematic data (Aiyangar et al. 2014), it has been shown that there is only minimal axial rotation and lateral bending associated with movements in the sagittal plane (Ellingson and Nuckey 2015; Harvey et al. 2015; Harvey 1998; Pearcy 1985), and therefore no significant advantage would be gained by using technologies capable of 3D measurements, when recording movements in this plane.

1.4 Contemporaneous monitoring of inter-vertebral passive and active control systems

In order to investigate interactions between and amongst the relevant muscle, joint and osseous components, suitable variables must be identified, and a method for integrating them developed. While motor control responses to perturbation around the neutral position (Cholewicki and Van Vliet Iv 2002; Hodges et al. 2009; Macdonald 2009; Macdonald et al. 2010; Radebold et al. 2000), and the flexion relaxation phenomenon (FRP) at the limits of sagittal flexion have been widely investigated (Kaigle 1998; Kippers and Parker 1984; Luhring et al. 2015; McGorry and Lin 2012; Watson 1997), no studies have addressed these interactions throughout the entire cycle of a functional task. During a voluntary movement such as forward bending there is a continual requirement to maintain spinal integrity, and so the measurement of variables that reflect control at points between the end-ranges of movement (mid-range variables) are of value (D'hooge et al. 2013). The measurement of contemporaneous multi-level kinematic and electromyographic information throughout the motion requires synchronised recordings from two different systems. Multi-level surface electromyography fulfils these

requirements for motor control (D'hooge et al. 2013; Wolf et al. 1979) and quantitative fluoroscopy measures a range of continuous inter-vertebral motion variables (Breen A.C. et al. 2012; Breen et al. 2015; Du Rose and Breen 2016b; Mellor et al. 2014). Contemporaneous recording of these measures therefore allows an integrated assessment of the interactions between the passive and active systems of the spine.

In order to control complexity and to limit possible confounders, the study was restricted to weight-bearing flexion in the sagittal plane, in a population of young adult males using a protocol that restrained the pelvis to avoid the effects of hip flexion. The heterogeneity of IV-RoM in healthy populations (Deitz 2011), lends it to the exploration of relationships with other variables, and it also represents the function of each motion segment in terms of restraint. The amplitude of rotation for a specific motion segment is dictated by the moments exerted on the vertebrae, and the nature of its restraining structures. These moments are produced by the position and weight of the torso and head during movement, and to a smaller degree by the action of the agonist muscles (Bogduk 1995). In the lumbar spine (when adopting a protocol that restrains the pelvis), resistance to these moments is primarily provided by the activity of the antagonist muscles (paraspinals), the lumbar disc and capsular ligaments, and to a lesser extent the longitudinally orientated spinal ligaments (Adams et al. 1980). Changes in IV-RoMmax will therefore relate to changes in the factors that restrain rotation. These will include the activation strategies of the antagonist muscles, the capacity of inter-segmental passive elements to resist rotation and the shape of the spine itself, mechanisms that are further complicated by the interactions between these factors across multiple levels of the spine. This thesis investigated these interactions and relationships in healthy controls, thus demonstrating how such mechanical parameters can affect inter-vertebral angular rotation in the absence of pain.

1.5 Thesis overview

The following literature review (Chapter 2) explores in more detail the problem of NSLBP, and builds a case for the development of a protocol that is capable of measuring concurrent inter-vertebral kinematic and muscle activity information. It reviews the various technological options currently available, and the methodological considerations associated with their use. The review also considers how various mechanical parameters may be expected to influence IV-RoMmax, providing an indication of the relationships that may be expected to be found, and supporting the development of the study's hypotheses. As a foundation to this, concepts relating to spinal

control and stability, and the anatomy and function of the active, passive and neural control elements of the lumbar spine (in relation to sagittal bending) are reviewed.

Chapter three consists of three preliminary studies. The first considers the most appropriate plane of motion for the main study (i.e. sagittal or coronal), the second is an initial investigation into relationships between lordosis and IV-RoMmax, and the third examines how the sEMG signal may be influenced by sEMG electrode placement accuracy. The methodology chapter (Chapter 4) outlines the QF and EMG parameters selected for use in the study, and describes the study protocol.

Chapter five determines the reliability and agreement of the study's primary parameter measurements. The chapter is divided into two parts. The first details an investigation into the agreement and reliability of the QF measurements that are utilised in the main study (i.e. IV-RoMmax and initial attainment rate), and the second determines the agreement and reliability of the selected sEMG parameter (i.e. normalised RMS sEMG amplitude).

The main investigations of the thesis are outlined in two separate chapters. The first investigates the relationships between kinematic parameters, lordosis and IV-RoMmax during forward bending (Chapter 6) and the second introduces sEMG, to investigate the relationships between paraspinal muscle activity and IV-RoMmax during the same movement (Chapter 7). Chapter eight outlines the contributions to new knowledge that have been made, and the discussion chapter that follows (Chapter 9) provides an overview and expansion of the work in light of the broader field of spinal biomechanics. This chapter also explores the possible clinical significance of the study's findings, its limitations and some additional data analysis.

The final chapter offers an overall conclusion of the study's findings in relation to its aims and objectives, and considers some possibilities for future work (Chapter 10).

Chapter 2: Literature review

Introduction

The *in vivo* investigation of relationships between and amongst the passive and active controlling elements of the lumbar spine at an inter-vertebral level, has been limited due to the technical difficulties associated with doing so. This literature review follows on from the introduction to re-iterate why such investigations are desirable, considers the methodological options available, and outlines the study's hypotheses, aims, research question and objectives. The review is divided into seven main sections.

Section 1: Anatomy and function of the lumbar spine

This section provides a comprehensive overview of the active and passive elements of the lumbar spine and relevant surrounding structures. The review focusses on the role of such elements in terms of sagittal forward flexion of the spine¹.

Section 2: Addressing the problem of NSLBP

This section reviews the problem of chronic non-specific low back pain (CNSLBP) and the perceived current lack of research into possible biomechanical causes. It is well documented that the heterogeneity that exists in CNSLBP populations makes linking biomechanical parameters to the condition very difficult. In those studies that do claim to be able to differentiate between LBP and healthy control groups using biomechanical parameters, contradictory findings are usually found elsewhere in the literature. This section therefore reviews these studies, and explores why further investigations into the relationships between the active, passive and neural control subsystems in healthy controls would be of value.

Section 3: Spinal stability and control mechanisms

This section reviews the concept of spinal stability, and describes the active, passive and neural control elements that combine to stabilise the lumbar spine, with a focus on the structures and control required to perform forward bending in the sagittal plane.

Section 4: Sagittal lumbar curvature and interactions within the passive system

This section reviews previous research, and explores the current ideas relating to spinal system interactions. The review provides an indication as to what relationships may be found when concurrently investigating kinematic and muscle activity variables at an inter-vertebral level, and

¹ Note: This study focusses on inter-vertebral movement in the sagittal plane. The reasons why this was chosen over other planes (e.g. coronal) are discussed in detail in a preliminary study (Chapter 3, section 3.1).

leads to the development of the study hypotheses outlined in section seven. The reasons why this study focusses on the kinematic and not kinetic strains of biomechanics are also outlined.

Section 5: Measuring lumbar spine kinematics and muscle activity

This section reviews spinal kinematic measurement techniques, provides an overview of electromyography, and reviews the methodological design considerations associated with their use.

Section 6: Dynamic task standardisation

This section addresses the pros and cons of methodological standardisation.

Section 7: Summary and conclusions

Finally, a summary of the literature review is provided including a focussed systematic review of the literature. The study's hypotheses, aims, research question and objectives are also outlined.

Method of literature review

Biomedical databases were searched in order to find literature in the areas of lumbar spine biomechanics, kinematic measurement, electromyography, spinal stability, and CNSLBP diagnosis and management.

The data bases used included PubMed, Ovid, Science direct, Elsevier, CINAHL, COCHRANE, Google Scholar and a private database held at the Institute for Musculoskeletal Research and Clinical Implementation (IMRCI). Examples of specific key word searches included "lumbar AND "kinematics" OR "biomechanics", "lumbar" AND "electromyography" OR "muscle", and "lumbar" AND "kinematics" and "reliability".

The searches produced hundreds of articles which were subsequently reviewed for their relevance, and reference lists were checked to make sure no additional pertinent papers were missed. In an attempt to keep up to date with the most current literature, citation alerts were added using the aforementioned keywords, and further papers were found using the "related citations" option for articles relating to the most relevant research.

Section 1

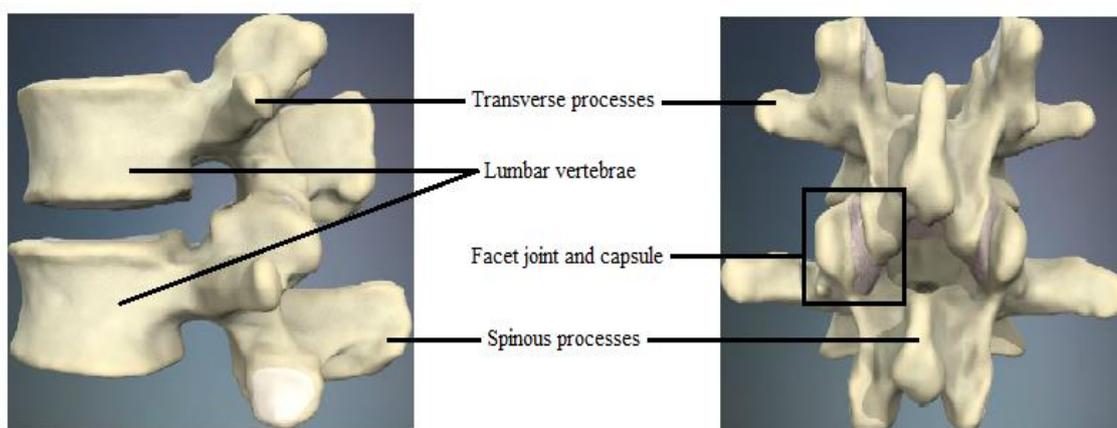
2.1 Anatomy and function of the lumbar spine

The various components of the lumbar spine can be divided into passive (vertebrae, discs and ligaments) active (muscles and tendons) and neural control elements, and all have a role in stabilising the lumbar spine during movement. Passive structures include the spinal ligaments, the capsules of facet joints, and the inter-vertebral discs, whilst the active tissues comprise of numerous muscles of the trunk, pelvis and hips. The motor control these muscles provide is regulated by inputs from various sources such as spinal stretch reflexes, cortical input and central control (i.e. adjustments based on vestibular and visual feedback). The following section provides an overview of the basic anatomy of these components of the spine, and introduces some of their functions in terms of spinal movement.

2.1.1 Passive elements

The passive elements are the spinal column's primary stabilising and load bearing structures (Chamoli et al. 2015). A functional spinal unit (FSU) or motion segment, is according to White and Panjabi (1990) "the smallest physiological motion unit of the spine to exhibit biomechanical characteristics similar to those of the entire spine" (White 1990), and consists of two adjacent vertebrae, the inter-vertebral disc between them, and the adjoining ligaments. The osseous structures of a FSU in the lumbar spine are shown in Figures 1A and 1B, and the key ligamentous structures in Figure 2.

Figure 1A and 1B: Lateral and posterior views of a lumbar motion segment. Images taken from www.anatomy.tv (04/03/2016)



2.1.1.1 Facet joints (Zygapophysial joints)

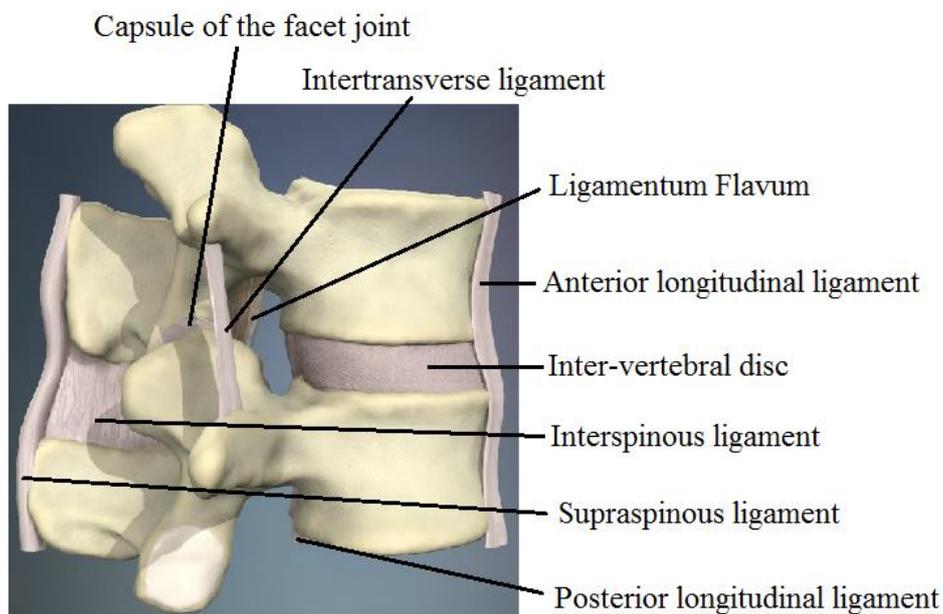
The lumbar facet joints (Figure 1B) are created by the articulations between the superior and inferior articular processes of adjacent vertebrae. The orientation of the joints determines the extent to which they can resist the anterior displacement and angular rotation associated with

forward bending. If the joint is orientated towards the coronal plane it is suited to resisting anterior translation, but not rotation. Joints orientated more in the sagittal plane will be suited to resisting rotation but provide little resistance to forward movements of the vertebrae. In addition to their anatomical positioning, it should be noted that the facet joint capsules themselves play an important role in stabilisation during dynamic movements (Adams et al. 1980).

2.1.1.2 Lumbar spinal ligaments

The ligaments found in the vertebral column (Figure 2) act in conjunction with the muscles and tendons to support the spine and protect it from injury. They are believed to have a role in neural control as sources of sensory feedback (Solomonow et al. 1998), contribute to joint stability during both rest and movement, and help prevent injury resulting from excessive movements (Sharma et al. 1995). In neutral positions (such as standing upright) ligaments provide only minimal resistance to movement, however as ligaments are stretched (i.e. during a motion such as forward bending) they become increasingly stiff and provide greater resistance to the motion (Adams 1999).

Figure 2: Lateral view of a lumbar motion segment with the key ligamentous structures attached. Image taken from www.anatomy.tv (04/06/2015)



In the lumbar spine, there are four main groups of ligaments. These include ligaments that connect vertebral bodies, ligaments that connect posterior vertebral elements, the iliolumbar ligament, and what have been described as false ligaments.

The anterior and posterior longitudinal ligaments are long bands that are found at the anterior and posterior aspects of each vertebral body and intervertebral disc respectively. Their actions are therefore closely associated with the relevant section of the annulus fibrosus. The anterior longitudinal ligament primarily resists distraction of the anterior part of vertebral column, and therefore acts to resist inter-vertebral extension. Acting on the opposite side, the main role of the posterior longitudinal ligament is to resist intervertebral flexion. Both ligaments attach caudally to the sacrum.

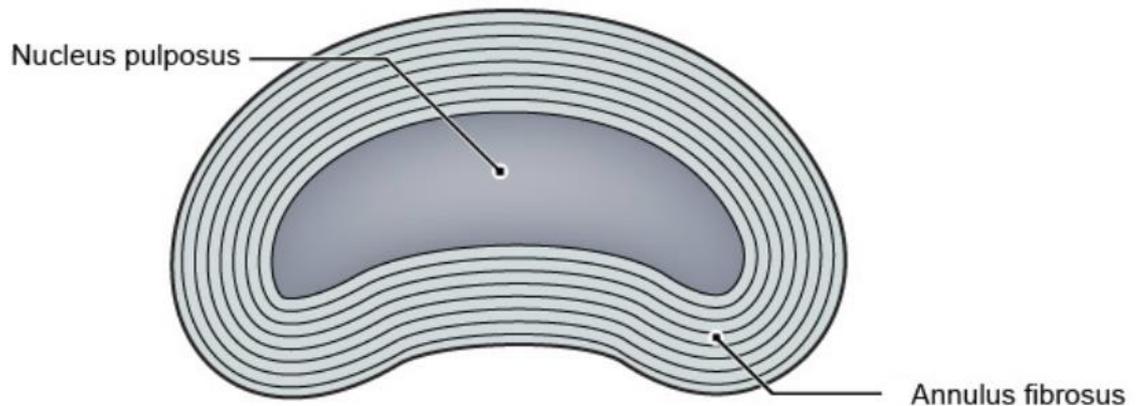
The ligaments of the posterior elements include the ligamentum flavum, the interspinous and supraspinous ligaments, and in a functional sense the ligaments of the facet capsule. Ligamentum flavum connects two adjacent vertebrae via attachments to the laminae of each bilaterally. It is unique in that it is chiefly composed of elastin, a trait believed to assist in its role in resisting separation of the laminae, but also as a mechanism to prevent buckling during approximation (Bogduk 2012). The interspinous ligaments attach between adjacent spinous processes, however they are not believed to contribute significantly to the resistance of forward bending moments, as alignment of their predominantly collagen fibres has been shown to run almost parallel to the spinous processes and not between them (Hukins et al. 1990). The supraspinous ligament attaches between the posterior edges of adjacent spinous processes, and whilst becoming thicker in the lumbar spine rarely reaches as far L4-L5 (Heylings 1978).

The iliolumbar ligaments attach to the transverse processes of L5 to the ilium bilaterally. Its primary role is to resist the forward translation of L5 on the sacrum, however it can also resist axial, sagittal and coronal rotation of the vertebra (Leong et al. 1987; Yamamoto et al. 1990). Finally, the so called 'false ligaments' are the intertransverse ligaments, the transforaminal ligaments and the mamillo-accessory ligament. The false ligaments have less biomechanical significance than those mentioned previously, and so will not be discussed in any detail.

2.1.1.3 The inter-vertebral disc

The inter-vertebral disc is found between two adjacent vertebrae (Figure 2) and consists of a central nucleus pulposus and a peripheral annulus fibrosus (Figure 3). The nucleus pulposus is a semi-fluid like substance, and as such can be readily deformed under pressure (Bogduk 2012). The annulus fibrosus on the other hand consists of fibrous rings of collagen fibres forming a tough exterior that encircles the softer inner core, and acts much like an additional spinal ligament.

Figure 3: Superior view of a lumbar inter-vertebral disc. Image taken from www.anatomy.tv (04/03/2016)



The combination of the two elements allows the disc to perform three main biomechanical functions. The first is to transfer loading through the spine without collapse, the second is to be deformable enough to allow inter-vertebral movements, and the third to be strong enough to avoid injury during such movements (Bogduk 2012). During a motion such as sagittal flexion, rotation of the superior vertebra over the inferior causes compression of the anterior annulus and stretching of the posterior side. The nucleus pulposus will respond to this anterior compression by moving backwards subsequently increasing pressure on the already stretched posterior annulus. A healthy disc can resist this combination of tension and pressure, which also influences the passive resistance characteristics of the disc during flexion.

2.1.2 Active elements

The section below provides a comprehensive review of the active components of the spinal control system. These include the muscles of the trunk, and those of the lower limb that also have a functional role in terms of lumbar stabilisation during forward bending.

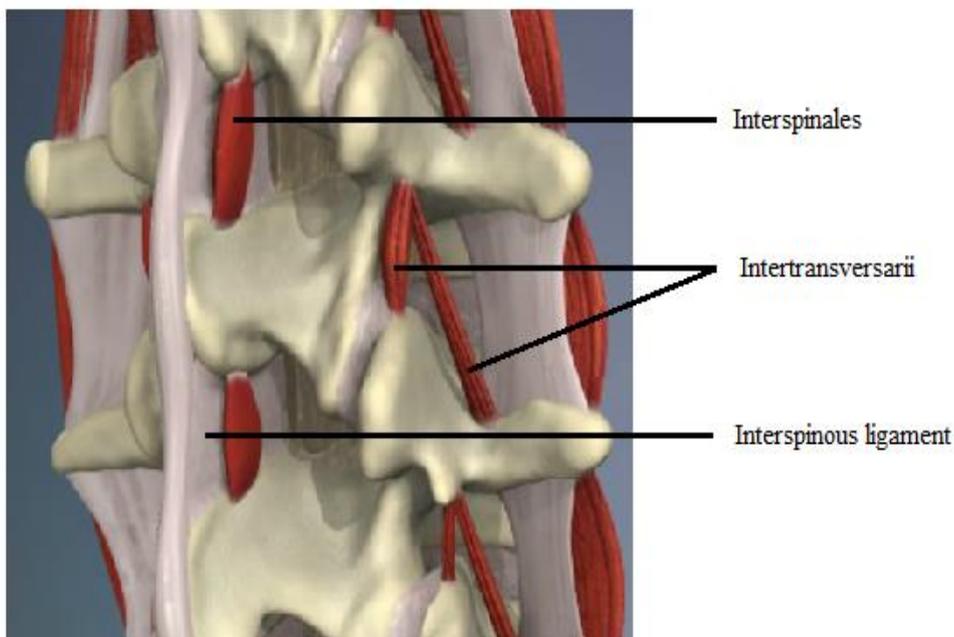
2.1.2.1 The lumbar paraspinal muscles

Anatomically, the lumbar paraspinal muscles reside behind the plane of the transverse processes, and can be divided for descriptive and morphological purposes into 3 groups. These include the short intersegmental muscles (the interspinales and intertransversarii mediales), the polysegmental muscles (the multifidus and lumbar regions of the longissimus and iliocostalis) and the long polysegmental muscles (the thoracic regions of longissimus and iliocostalis) (Bogduk 2012). In terms of function, the inter and polysegmental muscles (e.g. multifidus and longissimus thoracis pars lumborum) may be considered as locally acting, and the long polysegmental muscles (e.g. Longissimus thoracis pars thoracis) may be considered as globally acting (Bergmark 1989; O'sullivan 2000), (see also section 2.3.5).

2.1.2.2 The interspinales and intertransversarii mediales

The interspinales attach just lateral to the interspinous ligament between the spinous processes of neighbouring lumbar vertebrae, and the intertransversarii mediales attach to the accessory process, the mammillary process and the mamillo-accessory ligament inserting into the mamillary process of the segment below (Figure 4). Due to their small size and close proximity to the axis of movement, neither muscle is believed to contribute significantly to movement or indeed resisting movement. They have however been shown to have a much higher density of muscle spindles than longer polysegmental muscles, which suggests that their primary function may involve sensory feedback (Peck et al. 1984).

Figure 4: The lumbar interspinales and intertransversarii muscles (Posterior oblique view). Image taken from www.anatomy.tv (04/06/2016)



2.1.2.3 Lumbar multifidus (LMU)

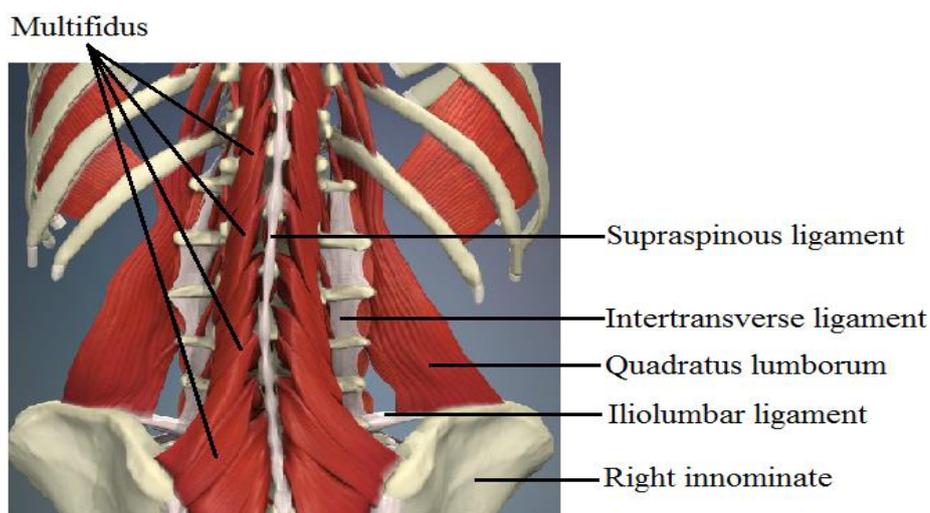
The lumbar multifidus (Figure 5) is a deep muscle that consists of a recurring sequence of fascicles that originate from the spinous processes and laminae of each lumbar vertebra. At each lumbar level, several fascicles arise from the spinous processes forming what is commonly known as the common tendon. This common tendon inserts into three separate areas of the spine, the lumbar mammillary processes, the iliac crest and the sacrum. The shorter laminar fibres insert into the mammillary processes of the vertebra two levels below or into the sacrum (for lower lumbar levels).

It is thought that the primary action of multifidus is to resist flexion and produce extension, as the force vectors of the muscle are aligned at right angles to the spinous processes they attach

to (Bogduk 2012). This makes them ideally suited to control of the lumbar spine during forward flexion. Indeed, the multifidus has been extolled as having the most suitable muscle architecture of all the paraspinal muscles to stabilise the lumbar spine during this movement (Ward et al. 2009). In their study, Ward et al. 2009 investigated the multifidus mass, sarcomere length, fibre length, physiological cross-sectional area and fibre length to muscle length ratio, finding that the muscle had an extremely high physiological cross-sectional area (greater than any other lumbar muscle). This combined with relatively short fibres (particularly at L4 and L5) means that multifidus can produce large forces over a narrow range of lengths, making them ideal for stabilisation (Ward et al. 2009). In addition, the study showed that multifidus exhibits a sarcomere length range exclusively on the ascending portion of the length tension curve (Ward et al. 2009), which suggests that the muscle will become intrinsically stronger with lumbar flexion.

It is suggested therefore that the multifidus is anatomically and biomechanically suited to control spinal movements (Macdonald et al. 2006). In vitro studies have shown that multifidus activity increases inter-vertebral stiffening in damaged motion segments (Panjabi et al. 1989), and provides as much as 2/3 of the stiffness at L4-L5 (Wilke et al. 1995). These findings suggest that the muscle has a role in the control of movement of both injured and healthy lumbar motion segments, and it is believed to have the capacity to control such motion, without constraining movement of the spine as whole (Macdonald et al. 2006).

Figure 5: The lumbar multifidus muscle and other local structures (Posterior view). Image taken from www.anatomy.tv (04/06/2015)



2.1.2.4 Longissimus thoracis pars thoracis (Thoracic erector spinae (TES))

The longissimus thoracis pars thoracis muscle (Figure 6) originates from the transverse processes and ribs of T2 to T12 and each level ultimately forms a caudal tendon that reaches the

lumbar region. The fascicle arising from the ribs and transverse processes of T2 insert at the L3 spinous process, T3 to L4 and T5 to L5 and so on. The fascicles arising from T8 to T12 however, attach to the sacrum between the spinous process of S3 and the posterior superior iliac spine. The caudal tendons of this muscle form the erector spinae aponeurosis. These fibres overlay those of the lumbar longissimus, but do not have any attachment to them.

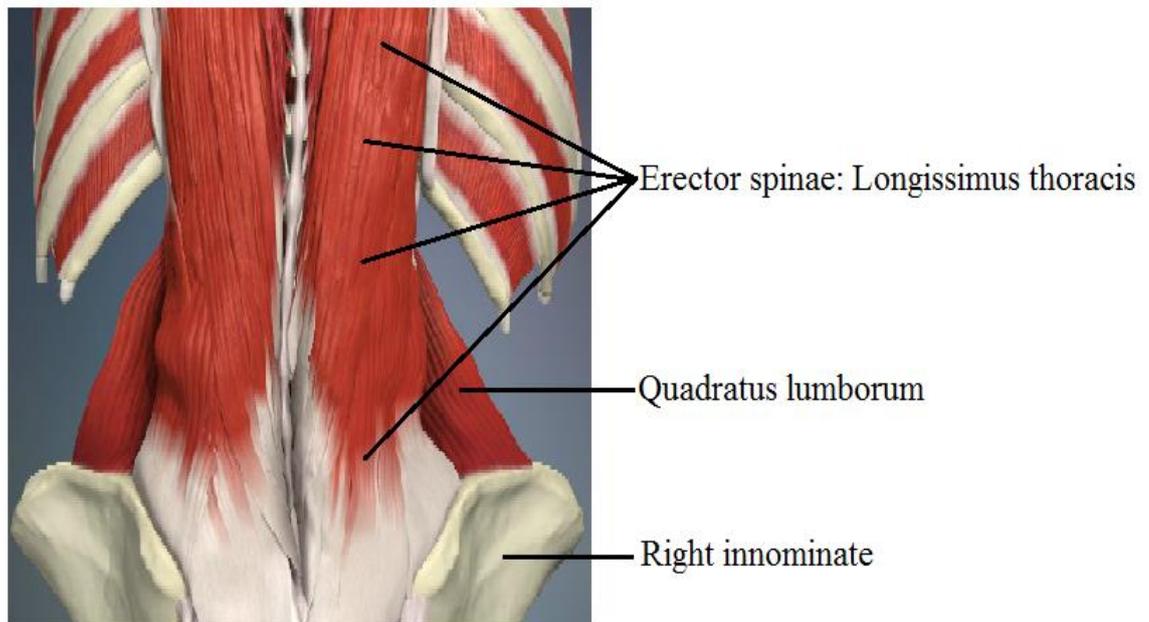
The fascicles originating between T6 and T12 traverse the whole lumbar region, and therefore through the erector spinae aponeurosis are able to influence the biomechanics of the area. The literature suggests that as this muscle spans from the ribs to the sacrum, it may be considered as “globally acting” (see section 2.3.5). The thoracic fascicles of the longissimus muscle group have been shown to contribute significantly to the extensor moment exerted on the lumbar spine (Bogduk et al. 1992), and as such they are regularly included in investigations of lumbar spinal control (Cholewicki and VanVliet 2002; Nelson Wong et al. 2010; Peach et al. 1998, Radebold et al. 2000; Van Dieen 2003).

2.1.2.5 Longissimus thoracis pars lumborum (Lumbar erector spinae (LES))

The longissimus thoracis pars lumborum muscle is composed of five fascicles, which originate from the accessory and transverse processes of the lumbar vertebrae. In the case of the lowest fascicle (L5), the fibres insert directly into the posterior iliac spine, however the remaining lumbar fascicles form tendons which converge to form the lumbar intermuscular aponeurosis. This aponeurosis attaches to the ilium lateral to the insertion of the L5 fibres. The role of this muscle is believed to differ between lumbar segmental levels due to the changes in the dominant force vectors generated by the fascicle orientations at each level. The upper lumbar spine fascicles are better suited to extending the lumbar spine, whereas the lower fascicles are more capable of resisting forward translation (Bogduk 2012).

This, combined with the fact that upper lumbar fascicles are more superficial (The more cephalic the lumbar level, the more superficial the muscle (Bogduk 2012)), is of potential importance when considering EMG recording techniques (see also section 2.5.6.1), as surface electrodes will predominantly pick up activity from the most superficial underlying muscles.

Figure 6: The longissimus thoracis muscle of the lumbar erector spinae and other local structures (Posterior view). Images taken from www.anatomy.tv (04/06/2015)



2.1.2.6 Iliocostalis lumborum (IL)

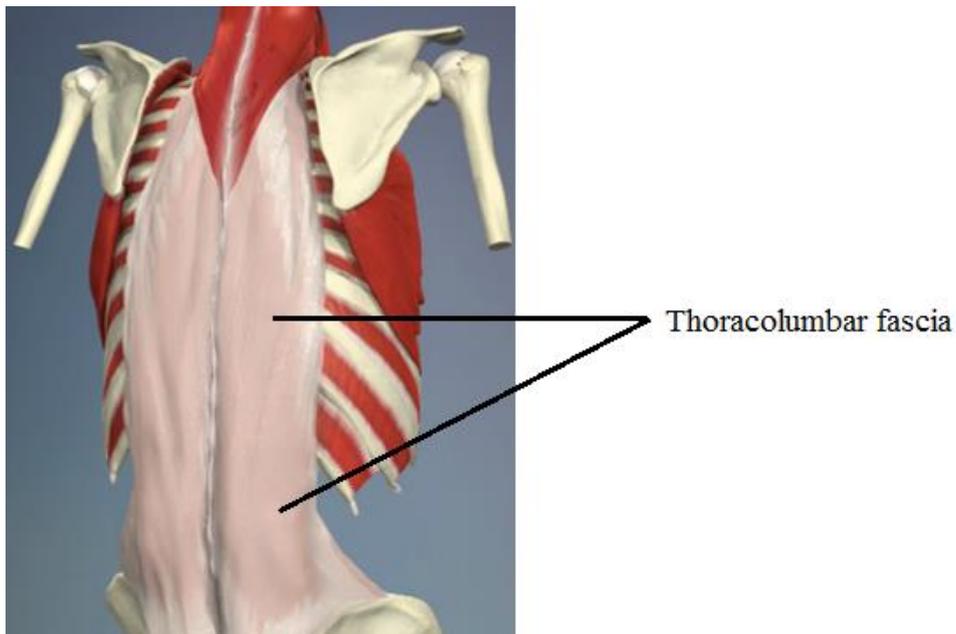
The iliocostalis lumborum consists of two separate components, iliocostalis lumborum pars lumborum and iliocostalis lumborum pars thoracis. The former consists of 4 fascicles originating from the transverse processes of L1-L4 and the thoraco lumbar fascia (TLF) lateral to them, which insert into the iliac crest. Like longissimus thoracis pars lumborum, the muscle is capable of resisting both sagittal rotation and anterior translation, but can also laterally rotate the lumbar spine. The iliocostalis lumborum pars thoracis is made up of fascicles that originate from ribs 5-12 that insert into the ilium of the pelvis and the sacrum, and therefore have no direct attachment to the lumbar spine. It can however still influence the global stability of the lumbar spine when contracting bilaterally, and invoke lateral flexion of the trunk when contracting unilaterally.

2.1.2.7 The thoracolumbar fascia (TLF)

It is not only the muscles of the trunk and lower limb that act to stabilise the lumbar spine, but a complicated partnership between these muscles, their fascia, and aponeurotic tissue that create a girde like structure around the spine (Bergmark 1989; Cholewicki 1997; Willard et al. 2012), an important component of which is the TLF. The TLF (Figure 7) consists of numerous aponeurotic and fascial levels and essentially separates the posterior abdominal muscles and the paraspinals. It consists primarily of the aponeuroses of the serratus posterior inferior and latissimus dorsi, and a retinacular sheath that surrounds the paraspinals. These layers converge towards the base of the lumbar spine and attach to the posterior superior iliac spine and the

sacrospinous ligament (Willard et al. 2012). There are several muscles that attach into the TLF, these include latissimus dorsi, gluteus maximus, biceps femoris and the muscles of the abdomen (primarily transversus abdominis (TrA)). Contraction of these muscles (as well as the paraspinals) can therefore provide a stiffening mechanism to the lumbar spine and pelvis via their action on the TLF. This increased stiffness will therefore play a role in augmenting the extensor moment during trunk flexion (Willard et al. 2012).

Figure 7: The thoracolumbar fascia (Posterior view). Images taken from www.anatomy.tv (04/03/2016)



2.1.2.8 The principal muscles of the abdomen

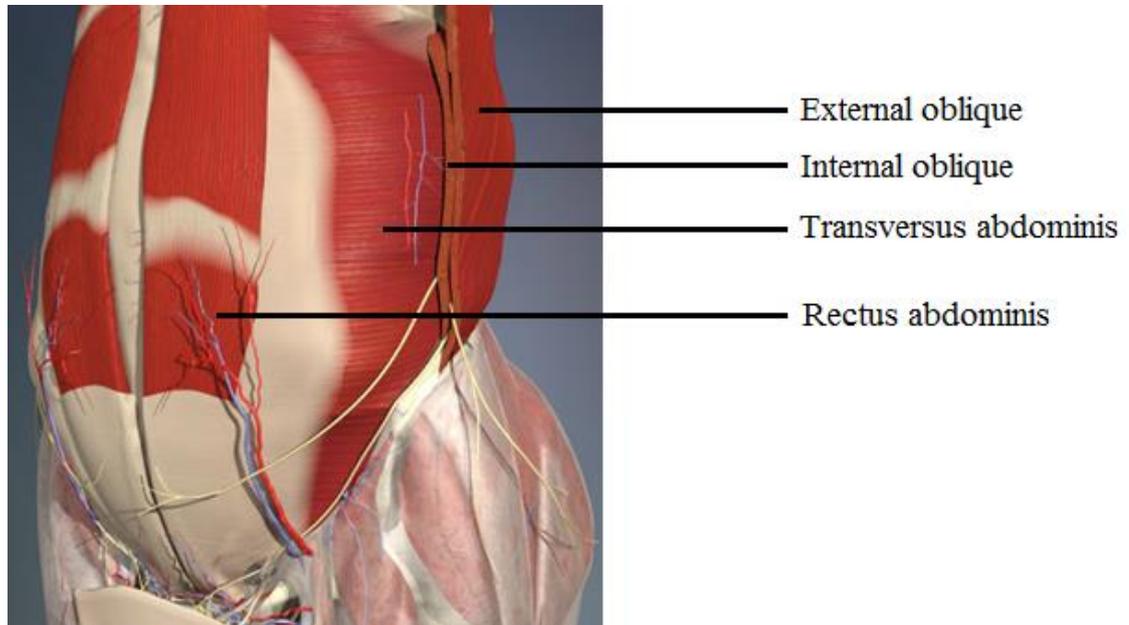
As well as their more obvious function as trunk flexors and rotators, the abdominal muscles are also believed to play an important role in trunk stabilisation through their co-contraction with the paraspinals (Cholewicki et al. 1999; Gardner-Morse 1998; Granata and Marras 2000; Granata and Orishimo 2001). The group consists of the transversus abdominis (TrA), the internal and external obliques and the rectus abdominis.

2.1.2.9 Transversus abdominis (TrA)

The TrA (Figure 8) originates from the inner surfaces of ribs 7-12, the thoracolumbar fascia (TLF), and the iliac crest and inserts into the linea alba, the pubic crest and pubis (via the conjoint tendon) (Moore et al. 2014). As an anterolateral trunk muscle, the link between TrA and the lumbar spine is a result of the anatomy of the TLF. The proposed lumbar stabilising mechanism is that when the TrA contracts, the thoracolumbar fascia tenses, subsequently raising the intra-

abdominal pressure (Cholewicki et al. 1999; Gardner-Morse 1998; Hodges et al. 2003; Kavcic 2004), and transmitting force to the spine (Barker et al. 2006).

Figure 8: The principal muscles of the abdomen (Anterior oblique view). Images taken from www.anatomy.tv (04/03/2016)



2.1.2.10 Rectus abdominis, internal and external obliques

The rectus abdominis (Figure 8) attaches to the pubic symphysis and crest and inserts into the xiphoid process and costal cartilage of ribs 5-7 (Moore et al. 2014). Its primary action is to flex the trunk, but it also acts to prevent anterior rotation of the pelvis. The internal oblique (Figure 8) originates from the TLF and the anterior iliac crest and inserts into the inferior borders of ribs 10-12, the linea alba and the pubis through the conjoint tendon. The external oblique (Figure 8) originates from the lateral surface of ribs 5-12 and inserts into the pubic tubercle, the linea alba and the anterior iliac crest. Both muscles function as trunk flexors and rotators, and compression support for the abdominal viscera (Moore et al. 2014).

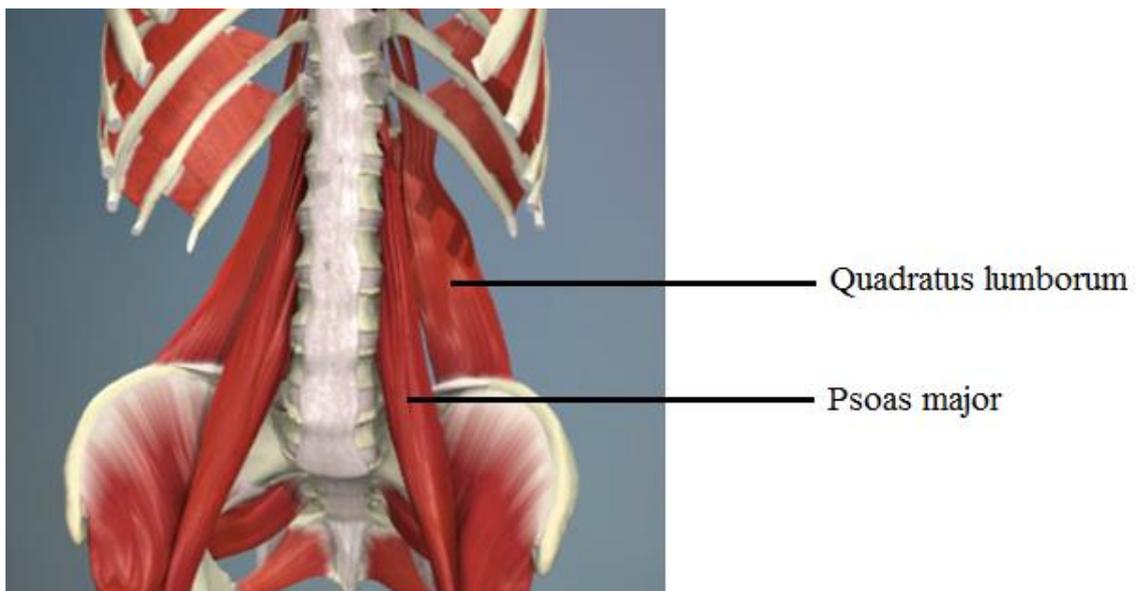
2.1.2.11 Muscles of the lower extremity, psoas major and quadratus lumborum (QL)

Anatomical texts typically categorise the lumbar spine, hips and pelvis and lower limbs as distinct entities. From a functional perspective however, trunk flexion is a combination of both lumbar spine and hip movements, and the interaction between them is known as lumbopelvic rhythm (Tafazzol et al. 2014). Indeed there are interactions between the active and passive tissues of the spine, hips, pelvis and the lower limb, and all can therefore influence the stability of the lumbar spine. The following muscles therefore also warrant discussion.

2.1.2.12 Psoas major

The psoas major muscle (Figure 9) attaches from the anterolateral lumbar spine (vertebral bodies, transverse processes and inter-vertebral discs T12 to L5) and inserts into the lesser trochanter of the femur. The muscle's primary action is to flex the hip, however the alignment of its fibres mean that it also has the capacity to flex and extend the upper and lower lumbar motion segments respectively. The muscle is not believed however to have a major role in maintaining mechanical stability of the lumbar spine (Bergmark 1989), and in agreement Bogduk et al. (1992) suggest that due to the close proximity of the fibres to the lumbar vertebrae axis of rotation, they are only capable of producing minimal moments even when maximally contracted (Bogduk et al. 1992). The vectors of action do mean however that psoas contraction can produce large compression loads, particularly on the discs in the lower lumbar region. This compression may therefore have a stabilising effect through changes made to disc stiffness.

Figure 9: The psoas major and quadratus lumborum muscles (Anterior view). Images taken from www.anatomy.tv (04/03/2016)



2.1.2.13 Quadratus lumborum (QL)

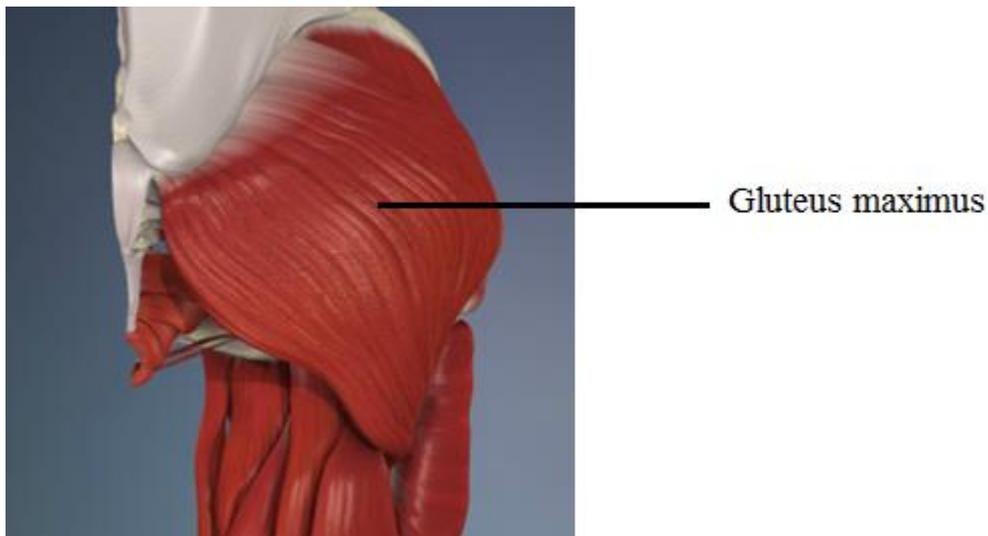
The QL (Figure 9) attaches to the lower medial half of the 12th rib, the transverse processes of L1-L4 and inserts caudally into the lip of the iliac crest. The QL is believed to have multiple functions, including the fixation of the 12th rib during respiration, ipsilateral lateral flexion, and extension of the lumbar spine. Bogduk (2012) suggests that while the muscle fibres are aligned behind the axis of sagittal rotation and therefore capable of resisting sagittal rotation, their capacity to perform this task is less than 10% of that of the paraspinal muscles (Bogduk 2012).

2.1.2.14 Gluteus maximus

The gluteus maximus muscle (Figure 10) attaches proximally to the posterior ilium, the dorsal sacrum, the coccyx and the sacrotuberous ligament, and inserts into the iliotibial tract and the gluteal tuberosity of the femur (Moore et al. 2014). The primary role of the gluteus maximus is to extend the hip, however it also has close relationships with both the lumbar paraspinals (via the TLF), and the biceps femoris (via the sacrotuberous ligament). These interactions assist with the transfer of load between the lumbar spine and the lower extremities and are believed to play an important part in lumbar stabilisation during trunk bending (Vleeming et al. 1995).

Due to their indirect influence it is more difficult to attribute activation of muscles such as gluteus maximus and biceps femoris to changes in lumbar inter-vertebral kinematics, than strong lumbar extensors such as the paraspinals.

Figure 10: The gluteus maximus muscle (posterior view). Images taken from www.anatomy.tv (04/03/2016)

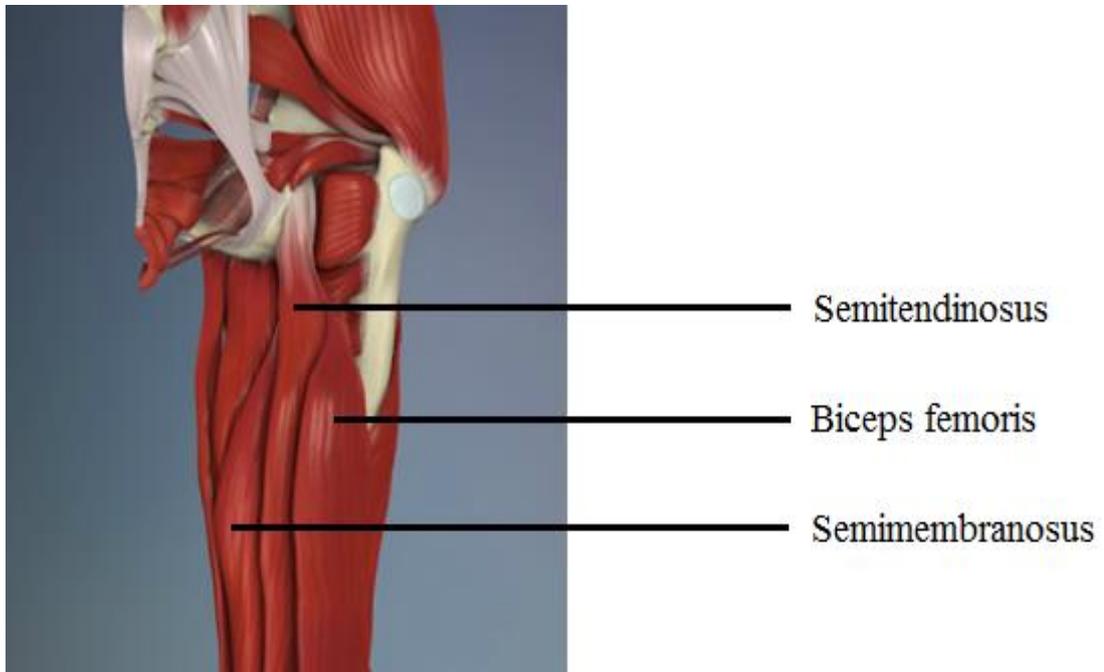


2.1.2.15 The hamstrings

The hamstrings group consists of three muscles, biceps femoris, semi-tendinosus and semi-membranosus (Figure 11). All three muscles attach proximally to the ischial tuberosity (including the long head of biceps), and insert into the lateral fibula head (biceps femoris), the superior medial tibia (semi-tendinosus) and the posterior medial condyle of the tibia (semi-membranosus) (Moore et al 2014). As the hamstrings all traverse both the hip and the knee joints, their principal actions are extension of the hip and flexion of the knee, however as in the case of gluteus maximus, they can also influence lumbopelvic rhythm via their attachments to the pelvis. Due to its attachment to the sacrotuberous ligament, biceps femoris is also believed

to indirectly influence lumbar spine stability by increasing tension in the ligament and therefore the TLF when contracting (Willard et al. 2012).

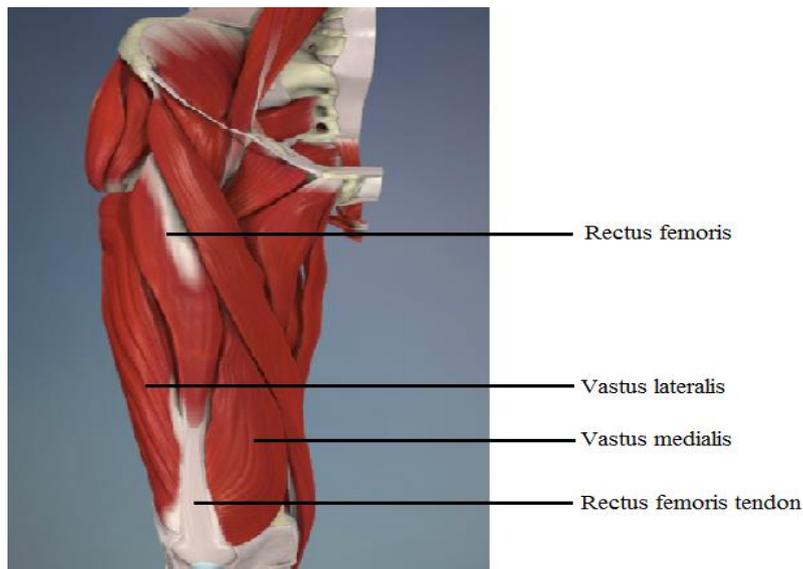
Figure 11: The hamstrings (Posterior view). Images taken from www.anatomy.tv (04/03/2016)



2.1.2.16 The quadriceps

The quadriceps group contains four muscles, the rectus femoris, and vastus medialis, intermedius and lateralis (Figure 12). The anatomy of the quadriceps muscle group means that only the rectus femoris is capable of influencing pelvic movement. The rectus femoris attaches from the anterior inferior iliac spine (AIIS) and inserts via the quadriceps tendon to the base of the patella and the tibial tuberosity (via the patellar ligament) (Moore et al. 2014). Its main functions are to extend the knee, help stabilise and assist the iliopsoas to flex the hip joint.

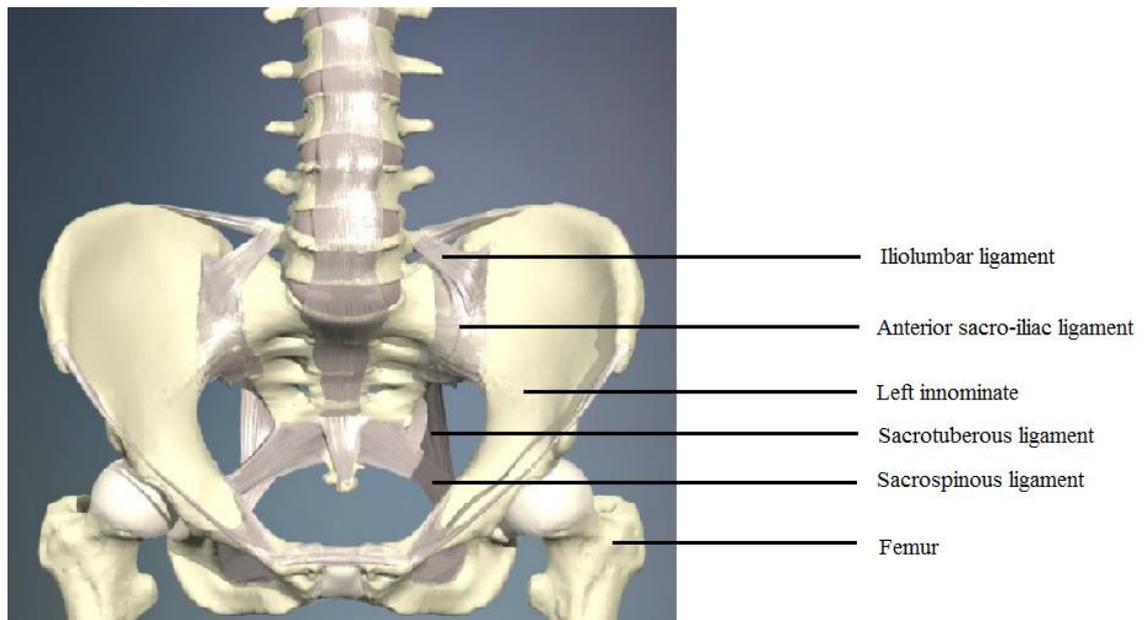
Figure 12: The quadriceps (Anterior view). Images taken from www.anatomy.tv (04/03/2016)



2.1.3 The pelvis

The pelvis consists of the two innominate bones, the sacrum and the coccyx. The ilium forms joints with the sacrum bilaterally called sacroiliac joints. This joint is stabilised by 5 main ligaments, the anterior, posterior and interosseous sacroiliac iliac ligaments, the sacrotuberous ligaments and the sacrospinous ligament (Figure 13). The pelvis's key functions in terms of biomechanics are to transfer load between the axial skeleton and the lower limbs (especially during movements), and to act as a point of attachment for the various muscles and ligaments required to perform/control such movements. In the case of forward flexion, reduced forward rotation at the pelvis may increase the requirement from the lumbar spine in order to reach a designated degree of trunk flexion.

Figure 13: The ligaments of the pelvis (Anterior view). Images taken from www.anatomy.tv (04/03/2016)



The contribution of trunk, pelvic, hip and lower limb muscles and ligaments will vary according to the task in question. In summary however, the posterior elements are best suited to resisting sagittal flexion and extending the lumbar spine, whilst the abdominal muscles primarily produce flexion and rotation. By their influence on intra-abdominal pressure, the abdominals also are also believed to have a key stabilising role, although the significance of their contribution is widely debated (see section 2.3.7). The spinal ligaments while providing increasing resistance through tension associated with length changes, also have a role in sensory feedback (see section 2.3.3).

Section 2

2.2 Addressing the problem of NSLBP

2.2.1 The 'biomechanical' components of low back pain

The diagnosis and management of NSLBP is clinically challenging, and the condition places a large socioeconomic burden on society (Bronfort et al. 2008). Prior to the emergence of the biopsychosocial model of low back pain (Waddell 1987), investigations into possible causes had predominantly focussed on the physical aspects of back pain, and there was a perceived neglect of the psychological and social components. The last 30 years however, has seen a notable divergence from a conceptual model of pathoanatomically grounded disease within the biomedical model, to a contextually grounded theory of illness within the biopsychosocial model (Weiner 2008). It is now widely recognised that low back pain is influenced by both biological and psychosocial factors, yet the balance of research over the last few decades has been largely focussed on the latter (Hancock et al. 2011). It remains unclear how effective treatments based on the biopsychosocial model actually are, and there is a perceived need for more high quality studies in the field (Chiarotto et al. 2015; Deyo et al. 2014; Karjalainen et al. 2001). In a systematic review, Guzman (2001) concluded that although intensive multidisciplinary biopsychosocial rehabilitation helps to reduce pain and improve function in low back pain patients, there were concerns over the expense and high frequency of the treatments required to do so (Guzman 2001). The concerns over cost have also been raised by Dufour et al. (2010), who when comparing multidisciplinary biopsychosocial rehabilitation to a back strengthening program found no difference in outcome between groups (Dufour et al. 2010). The effectiveness of a stratified management approach that included psychologically informed physiotherapy has also been demonstrated, however the comparative benefit compared to current best practice (physiotherapy) appeared to be only short term (Hill et al. 2011). A recent stream of research investigated structural and functional changes within the CNS (neuroplasticity) in people with a chronification of pain (Hashmi et al. 2013), but whilst a deepened knowledge of neural changes provides a novel and valuable new approach to the problem, especially in the area of NSLBP when clear signs of damaged musculoskeletal structures are rarely evident (Pelletier et al. 2015a, 2015b), it focusses on changes that are typically secondary to some kind of peripheral stimulus and it could therefore be argued that it is not addressing the root of the problem.

The dilemma therefore is how to differentiate the different causes of NSLBP to facilitate clinical decision making in the assessment and treatment of these patients. Various attempts to determine the components that would facilitate better NSLBP diagnosis and outcomes (clinical prediction rules (CPRs)), and to establish methods that distinguish subgroups of LBP from each other have been made (Childs and Cleland 2006; O'sullivan 2005; Sahrman 2002). Although promising, current evidence suggests that CPRs are not yet ready for direct clinical application (Haskins 2012; May 2009), and the value of sub-grouping methods remains widely unproven (Fritz et al. 2007; Hebert et al. 2011). There is therefore large scope for improved understanding, and a perceived requirement for further investigation into the role of biological components, including those of mechanical origin.

It is clear that low back pain remains a multi-factorial problem; however as discussed, a greater emphasis could be placed on research that considers its mechanical components. Historically the limitation of models such as the Disease Model (Virchow 1858) has been the inability to identify a causative structural lesion. In the lumbar spine for example, the visualisation of continuous inter-vertebral motion has until recently not been feasible (Weiner 2008), leaving the potential to miss potentially important biomechanical changes² that occur throughout or at specific points during the entire spinal range of movement. If the concern then is that current low back pain research neglects the biological components (Hancock et al. 2011), technological advances that provide new ways of measuring such elements, may provide a way forward for an area of research that requires a renewed focus.

2.2.2 The importance of 'normal' (the study of low back pain free participants)

The background (section 1.2) highlighted the difficulty in determining factors that can reliably distinguish CNSLBP groups from healthy controls, and that much contradiction exists between the findings of studies that investigate the same biomechanical variables. Table 1 compares a number of studies that attempt to determine mechanical links with LBP groups, and demonstrates that the findings of such investigations are sometimes contradictory to each other. The nature of NSLBP means that within a low back pain group (and low back pain free control groups for that matter), different structural, chemical and neuromuscular changes may be found (Ross et al. 2015). The array of potential dysfunctions within each sample makes it

² Note: Although biomechanical influences include those of a 'chemical' origin, the consideration of such factors are beyond the scope of this thesis.

difficult to decipher whether it is a specific biomechanical change, a combination of such changes or the influence of pain that is the primary cause of the problem.

Table 1: Contrasting findings of studies attempting to determine biomechanical variables that differentiate between LBP patients and healthy controls

Examples of biomechanical variables	Author	Study findings (Variable higher, lower or the same in LBP populations compared to healthy controls)
Regional lumbar spinal ROM during flexion (unless indicated *)	Teyhen et al. 2007	The same (regional and inter-vertebral ROM*)
	Dankaerts et al. 2009 (Using sub-grouping)	The same (for the flexion provocation group)
	Taghipour-Darzi et al. 2012 (Using sub-grouping, lumbar segmental instability (LSI) or not LSI)	The same (for full flexion), however mid-range rotation was lower in both lumbar segmental instability (LSI) and non- LSI groups
	Triano et al. 1987	Lower (during full flexion within pain limits)
	McGregor et al. 1995	Lower (during full flexion)
	McGregor et al. 1997	Lower (during full flexion for LBP group as a whole) however, higher for NSLBP group
	Sanchez-Zuriaga et al. 2015	The same
Muscle activity amplitude	Ahern et al. 1988 (sEMG recorded from lumbar paraspinals)	Lower
	Sanchez-Zuriaga et al. 2015 (sEMG recorded from the erector spinae)	Higher
	Sihvonen et al. 1991 (sEMG and fine wire electrodes recording from the lumbar paraspinals)	Higher
	Kuriyama et al. 2005 (sEMG recorded from lumbar multifidus and longissimus)	Higher
Muscle activity amplitude ratios (e.g. LES/TES ratio)	Van Dieen et al. 2003	Higher
	Van Den Horne et al. 2012	Higher
	Reeves et al. 2006	Lower
	Lariviere et al. 2002	Lower
Lordosis	Tuzun et al. 1999,	The same
	Sarikaya et al. 2007	The same
	Christie et al. 1995	Higher
	Dankaerts et al. 2009 Lordosis during usual sitting (Using sub-grouping)	Lower (for the flexion provocation group i.e. more kyphotic), and higher (for the extension provocation group i.e. more lordotic)

An additional confounder is LBP itself, as there is ongoing debate as to whether LBP (if involved at all) is actually the cause or the consequence of such biomechanical changes (Hodges et al. 2013). This lack of clarity means that investigations into possible links between LBP and mechanical factors need to consider the potential influence of pain, and indeed the fear of pain in participants, as both can affect mechanical behaviours (Asmundson et al. 1997). In an attempt to address the problem of pain (as a possible confounder), studies have investigated chronic low back pain groups in pain free periods (Sanchez-Zuriaga et al. 2015), however the problem of multiple potentially unaddressed influences remains. This has been a problem for all studies in this field, leading to the development of different strategies designed to help better distinguish between LBP and healthy control groups. Such approaches usually try to somehow narrow the focus of investigation, and include attempts to sub-group the NSLBP group in terms of movement provocation (Dankaerts 2009; Hemming et al. 2015), divide the lumbar spine into regions (Hemming et al. 2015; Pavlova et al. 2015), and investigate changes in parameters such as muscle activity (Dankaerts et al. 2009, D'hooge et al. 2013), or kinematics (Taghipour-Darzi et al. 2012) in sections throughout the motion cycle and not just at the end ranges which has been more typical (Miyasaka 2000).

Despite this, in every case, if NSLBP groups are analysed whilst in pain or if the variable under consideration is investigated in isolation, then it is not possible to say that the presumed correlated biomechanical parameter(s) are not simply variations of normal biomechanical behaviour. If we use the example of paraspinal muscle activity as a variable (Table 1), it has been suggested that activity imbalances between segmental levels, may be markers that distinguish between CNLBP patients and healthy controls (Van Dieen et al. 2003, Van Den Hoorn et al. 2012). These studies report that a relative increase in lumbar erector spinae (LES) activity compared to the thoracic erector spinae (TES) is a stabilisation strategy adopted by the LBP groups to enhance spinal stability (Van Dieen et al. 2003). Such findings are contradictory to other studies however, which have shown the complete opposite (Lariviere 2002; Reeves et al. 2006). Although these differences may partly be explained by differences in methodology, it is likely that optimal muscle recruitment strategies will vary due to the biomechanical requirements of each individual. It is not known whether such variations in strategy exist within a spectrum of normal biomechanical behaviour.

2.2.3 Is it time for a fresh approach?

The development of protocols that investigate spinal biomechanics at an inter-vertebral level in people without LBP, including the interactions that exist between such levels, may help to further understand what is normal. Indeed, in terms of kinematic and EMG variables, Peach et

al. (1998) suggested that by knowing what is 'normal' we may be able to objectively evaluate rehabilitation protocols, and classify different low back pathologies (Peach et al. 1998). This view was shared by Wong et al. (2004) who proposed that pathologic spinal motion can only be identified if 'normal' spinal motion is defined (Wong 2004), and by Teyhen (2007) who described the need to determine what 'normal' is, as a precondition to progressing the use of altered kinematics as a parameter (Teyhen et al. 2007).

Using the example of ROM, it has been suggested that traditional measurements such as total spinal ROM, in isolation, have limited clinical usefulness when attempting to distinguish LBP and healthy controls, due to inherent heterogeneity in both groups (McGregor et al. 1997). There is therefore a growing belief that research needs to move away from conventional end of range measurements, towards using mid-range (Taghipour-Darzi et al. 2012) or higher order variables such as displacement, acceleration and velocity (Lehman 2004), and that consideration should be given to the spine as a whole (i.e. a group of interacting segments) as opposed to motion segments viewed in isolation. Indeed, in a recent fluoroscopy study analysing recumbent bending, Mellor et al. (2014) claimed to be the first to demonstrate measurable biomechanical differences between NSLBP patients and healthy controls, using what they termed "combined proportional range variance (CPRV)" (Mellor et al. 2014). They found that the variation in proportional motion between lumbar vertebrae was significantly greater in NSLBP patients than in healthy controls, providing evidence that the interaction between segments may be of greater importance than inter-vertebral motion in isolation.

The present study investigates the normal biomechanics of the lumbar spine using measurements of both spinal kinematics and associated muscle activity. An improved understanding of the normal interactions that exist in healthy participants, should enhance our understanding of what is truly 'normal' and provide a basis for identifying what is 'abnormal'.

Section 3

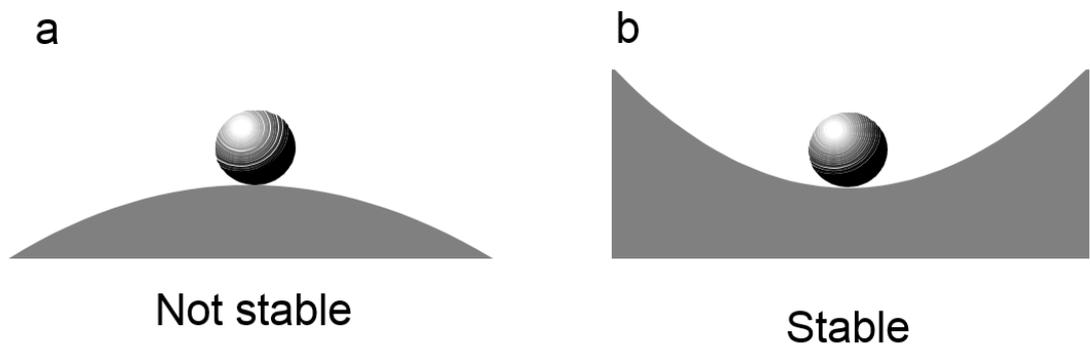
2.3 Spinal stability and control mechanisms

2.3.1 Spinal stability

The term spinal instability will mean different things to different people, as engineers for example may interpret it differently to spinal pain clinicians (Reeves and Cholewicki 2003). As clinicians, if a study participant were asked to perform a task such as forward bending, and the individual successfully performed this movement, the spinal system could be thought of as stable (i.e. controlling the movement and bearing load without injury). Whilst measuring various kinematic and muscle activity parameters provides an insight into how an individual maintains stability throughout this movement, it is perhaps incorrect to suggest that these measurements can provide an indication of the degree of stability provided by the system. Indeed, according to Reeves et al. (2011) it is important to state that a spine can by definition only be stable or unstable, and so the use of terms such as increased 'stability' are perhaps not appropriate (Reeves 2011). Instead, Reeves et al. (2007) suggest the use of the terms robustness and performance (Reeves et al. 2007)³. To explain these concepts the authors used the analogy of a ball on a surface (Figures 14-16). Figures 14a and 14b represent systems that are unstable and stable, with stability dependent on the shape of the surface on which the ball lies. In Figure 14a, any size of perturbation (movement of the ball) will result in the ball rolling away from the undisturbed position, representing an unstable system. In Figure 14b however, the raised slope of the surface either side of the ball mean that even with a reasonably strong perturbation it will eventually return to its original position, characteristic of a stable system.

³ Note: The term 'stability' is used throughout this thesis, and any reference to an increase or decrease in stability should be considered in terms of robustness and performance. The use of the generic term reflects its continued use in the wider literature.

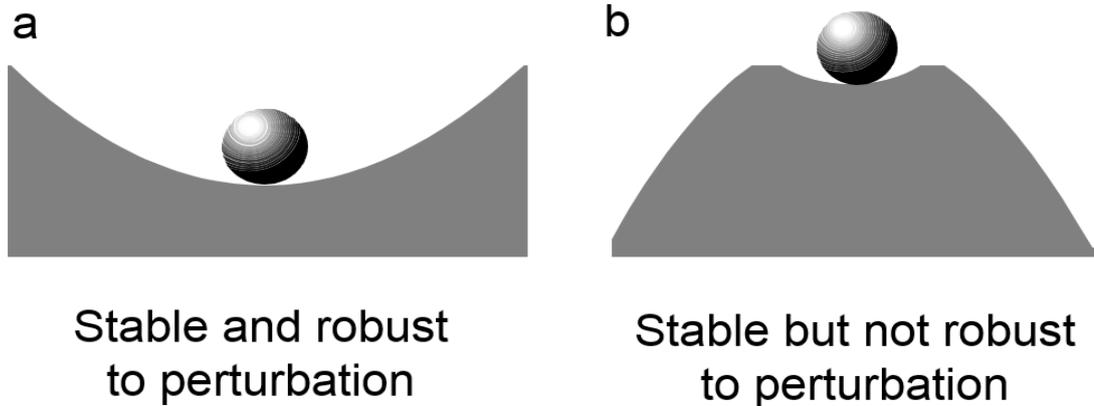
Figure 14: The concept of unstable and stable systems using the analogy of a ball on a surface. Used with permission from Professor Peter Reeves (29/03/2016)



If the system of concern is the spine, then maintenance of stability during movement and perturbation becomes a lot more complex. The state of the spine (e.g. the position of a vertebra) during any given movement is believed to be tracked using sensory feedback from many different sources (see section 2.3.3). Reeves et al. (2007) suggest that while this wealth of information contributes to producing a neural representation of the system, it is still only an estimate, therefore any uncertainty in the true system representation will have to be tolerated somehow to maintain stability (Reeves et al. 2007). The ability of the spine to remain stable under a range of perturbations is therefore an indication of its robustness. Figure 15a represents a system that is relatively more robust than that seen in Figure 15b, as the steepness of the surface walls in the former will tolerate a larger range of perturbations than the latter. Reeves et al. (2007) relate the steepness of the wall to the 'stiffness' of the spine, and a characteristic of a robust spine is that it can adapt its stiffness in accordance to the disturbance (Reeves et al. 2007).

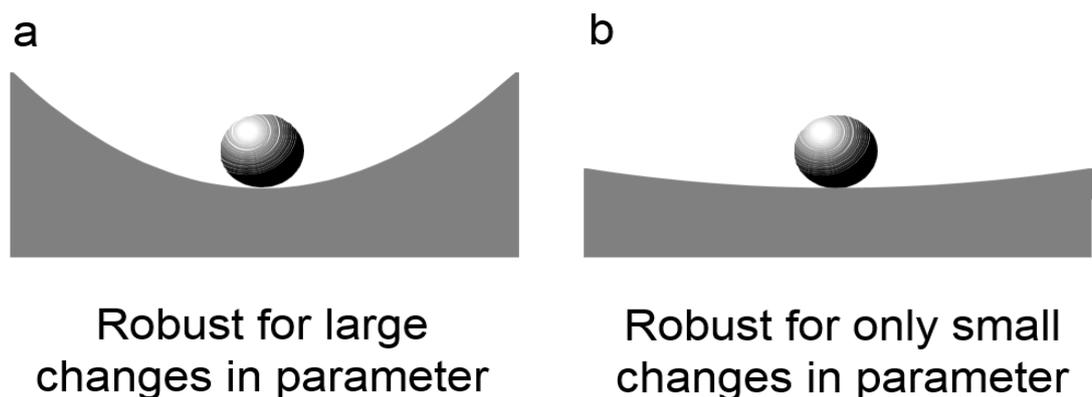
In the lumbar spine, stiffness is influenced by all three subsystems; the passive, the active and the neural control subsystems (Panjabi 1992a). Whilst recent technological advances are providing an initial insight into the control system through direct stimulation of the motor cortex (Chiou 1996; Tsao et al. 2011), understanding of neural control can also be furthered by exploring the relationships between and amongst the passive and active systems.

Figure 15: The concept of robustness in response to perturbation using the analogy of a ball on a surface. Used with permission from Professor Peter Reeves (29/03/2016)



In a stable system, Reeves et al. (2007) also explore the idea of performance (Reeves et al. 2007). Performance is the speed and accuracy at which a disturbed system can return to its initial undisturbed state. In Figures 16a and 16b, both systems may be described as stable, with the former being more robust than the latter due to the steepness of the surface walls. In terms of performance, disturbance of the ball in system 'a' will result in a quicker and more accurate return to the undisturbed position than the ball in system 'b', it is suggested therefore that the former represents higher performance.

Figure 16: The concept of robustness in response to changes in system parameters using the analogy of a ball on a surface. Used with permission from Professor Peter Reeves (29/03/2016)



It has been demonstrated *in vitro* that the ligamentous lumbar spine becomes unstable under loads of approximately 90N, and it is accepted that the activation of surrounding muscles stabilise the spine during loading that exceeds this level (Crisco 1991). Numerous trunk muscles (see section 2.1.2) contribute to spinal stabilisation during postural functions and voluntary

movements; and as such they can be considered as important components of the spinal control mechanism, due to their influence over the robustness and performance of the system (Reeves et al. 2007). Generally speaking, motor control strategies are employed to optimise the movement in terms of energy cost to the system and to protect the spine from any movement that could result in injury or pain (Cholewicki and McGill 1996; Granata and Marras 2000). In order to achieve these aims, muscle activity can change between and within muscles, and inputs may come from local and global levels of the motor control system (Bergmark 1989). These changes can have a direct impact on the mechanical behaviour of the spine, influencing spinal stiffness and modifying movement patterns.

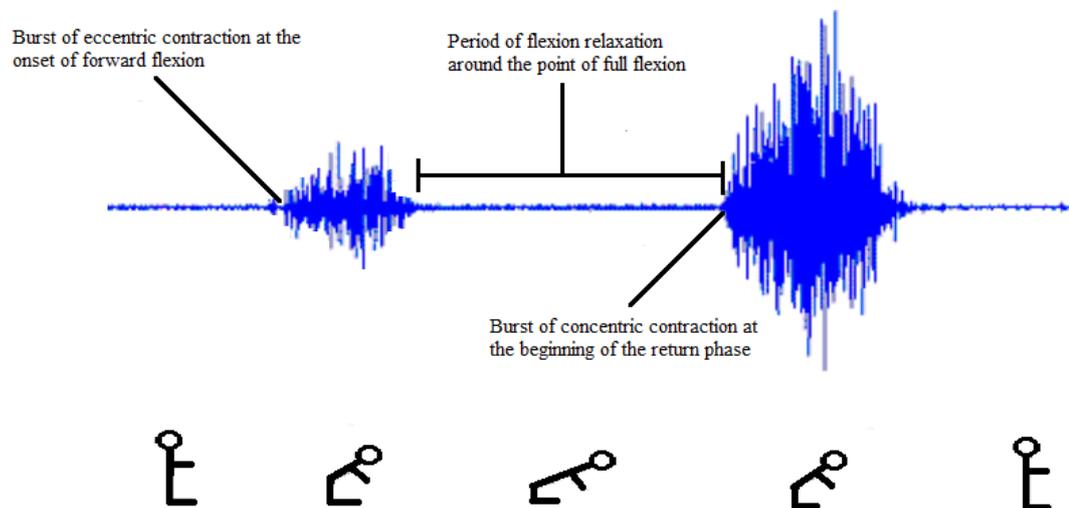
Returning to the example of a study participant performing a forward bending and return task, where they primarily bend from the lumbar spine, it can be assumed that the participant's spine was stable, if they completed the movement without injury to any spinal tissues and without having to compensate for movement by making adjustments to body position beyond the spine to maintain balance. The way in which the various active, passive and control elements combine through the neural control system to achieve stability during these movements is therefore of interest, especially as so little is currently known about such interactions at an inter-vertebral level. Investigations into spinal biomechanics at this localised level will provide a deeper insight into how interactions at an inter-vertebral level contribute to the control mechanisms of the spine.

2.3.2 Maintaining the equilibrium of the lumbar spine during forward bending

The bulk of recent research into spinal control mechanisms has concentrated on comparisons between LBP and healthy groups, and differences in muscle activity recruitment strategies related to perturbations from neutral spine positions (Cholewicki and Van Vliet 2002; Macdonald et al. 2009; 2010; Radebold et al. 2000). The findings of these studies suggest that delayed and decreased activity of the deeper paraspinal muscles and increased activation of the superficial muscles are the strategies adopted to restore or maintain the equilibrium about the neutral position, while maintenance of spinal integrity is also required during voluntary movements of the trunk through pre-determined trajectories (D'hooge et al. 2013). Using the example of sagittal flexion and return, a study by Peach et al. (1998) using Fastrak and sEMG described the general muscle activation patterns during these movements (Figure 17). Bursts in signal amplitude correlated with the onset of forward flexion, and the beginning of the return phase. Between the two bursts around the point of full flexion, there is a period of significantly reduced activity known as the flexion relaxation phenomenon (FRP) (Peach et al. 1998). Whilst many studies have investigated muscle activation patterns during trunk flexion, the majority

focus on this FRP (see Table 2), with little emphasis on changes in recruitment throughout the entire movement cycle.

Figure 17: A typical lumbar paraspinal muscle EMG trace during sagittal flexion and return



The controlling muscular contractions can be defined as concentric (i.e. there is an increase in muscle tension as it shortens) or as eccentric (i.e. there is an increase in muscle tension as it lengthens). A simplistic overview of muscular control during forward bending is that any moment produced by the head and trunk during the movement must be countered by the action of the surrounding trunk musculature (in this case predominantly the eccentrically acting paraspinal muscles). However, it is not only the active tissues that are capable of providing restraint, as the passive structures also facilitate control of the bending movement. The mechanisms that facilitate control are believed to be initiated by displacements in the equilibrium of the system, and as such they can be thought of as feedback mechanisms.

Determination of spinal stiffness relates to position feedback in the spine and is dependent on the length of the passive and active tissues, and therefore the movement between vertebrae. As well as stiffness, another component involved in resistance against moment is damping, a concept that has received far less attention in the literature (Reeves 2010). Damping is an intrinsic property of the spine dependent on the rate of change in length of the passive and active tissues (and therefore their poroelastic and viscoelastic properties), and should therefore be considered alongside stiffness when investigating dynamic movements (Reeves et al. 2007). In terms of a familiar voluntary movement such as forward bending, the degree of muscle activity will be controlled predominantly via a feed-forward governance of the motor cortex, regulating muscle stiffness and damping. This stiffness and damping subsequently influences

the restraint characteristics of the passive structures, both of which affect their control on the movements of the spine without any time delay.

A common misconception is that there is a clear distinction between open-loop control (with no sensory feedback) and closed loop control (with sensory feedback) in terms of the control of voluntary movements (Nielsen 2004). Despite voluntary bending being a well-practised movement, additional stiffness and damping will also be provided to the system via reflex muscle activity in response to any unexpected changes in the length and rate of elongation in the passive and active tissues, which can be considered a reflection of changes to the intended inter-vertebral movements (spinal feedback). This reflexive influence over control comes with a small time delay, as does the other significant contributor to control, input from visual and vestibular (supraspinal) feedback systems.

2.3.3 Spinal feedback mechanisms

Feedback mechanisms contributing to spinal control come from sensors that communicate information about the status of the whole system (Reeves et al. 2007), in particular passive and active systems are believed to interact providing a feedback control mechanism (Solomonow et al. 2010; Vleeming and Willard 2010), a synergistic relationship that provides stability and stiffness to the spine during movement (Stubbs et al. 1998). At an inter-vertebral level, these sensors provide information regarding the position and velocity of vertebral movements, and are located in both the muscles and the passive structures associated with the inter-vertebral joint. In terms of the active tissues, this information is believed to be provided by muscle spindles and Golgi tendon organs. The relative density of muscle spindles has been shown to correlate with the capacity of the muscle to provide proprioceptive feedback (Buxton and Peck 1989), and Golgi tendon organs, which are located in the muscle tendons, are believed to sense changes in muscular tension and prevent overexertion via inhibition feedback mechanisms (Brooks 1986; Windhorst 2007). In the passive tissues of the lumbar spine, mechanoreceptors are located within the ligaments, facet capsules and discs (Indahl 1997; Kiter et al. 2010; Ozaktay 1991; Roberts 1995; Yahia et al. 1992). Types of mechanoreceptor include Golgi and Rufini endings (Roberts 1995), which are believed to respond to excessive deformation of ligaments and joint capsules (Clark and Brugess 1975), and provide information regarding changes in velocity, the position, and pressure within a joint (Johansson et al. 1991). Information from sensors throughout the active and passive systems therefore can give continuous feedback via neural control, creating a system with the capacity to adapt muscle recruitment in accordance to the task.

Paraspinal muscle activity during sagittal flexion in healthy spines can typically be separated into two stages. The first is muscle activation, and can be partially explained by the concept of ligamento-muscular synergism. This theory suggests that stress and strain in the passive tissues is monitored by mechanoreceptors, which relay information to the CNS which subsequently initiates muscle activation associated with the motion segment (Solomonow et al. 1998). This mechanism is believed to control movement of the bones and protect the passive structures from injury. Feline and porcine studies have demonstrated a direct association between ligament stimulation and EMG activity in the paraspinals, with muscle excitation being shown to be strongest either at the level of stimulation (Solomonow et al. 2002), or the motion segment below (Stubbs et al. 1998). In both cases, paraspinal activity was recorded as far as 2 levels above or below the level of stimulation, which suggests a stiffening strategy that goes beyond single motion segments, but means that the precise levels at which the ligamento-muscular reflex acts remains unclear. Indeed, it is suggested that EMG discharge is often not graded with joint movement (Grigg 2001), and as studies typically only stimulate one level, therefore excluding the input and resulting interactions from other levels, the mechanism requires further clarification.

The role of stretch receptors in lumbar muscles has received somewhat less attention. Solomonow et al. (2002) suggest that stretch receptors in the paraspinal muscles do not participate in the EMG activation reflex described previously (Solomonow et al. 2002), however relationships have been shown between lumbar paraspinal muscle spindle discharge and longissimus and multifidus lengthening (Cao et al. 2009), which alludes to an important role in spinal control. Indeed Kang et al. (2002) suggest that stimuli from the medial paraspinal tissues may influence efferent activity to adjacent motion segments, and therefore directly contribute to biomechanical behaviour (Kang et al. 2002). They acknowledge however that the function of such inter-segmental reflex pathways again requires further clarification (Kang 2002).

The second stage that normally occurs during forward bending is a myoelectrical deactivation of the erector spinae, a muscle activity decrease that goes on until the start of the relaxation phase (i.e. the onset of the FRP) (Sanchez-Zuriaga et al. 2015), associated with a shift of moment from the active to the passive structures (Floyd and Silver 1955). This transferral is also likely to be linked with a redistribution of muscle activity to muscles less frequently recorded, (e.g. Quadratus lumborum, and deep erector spinae) (Andersson et al. 1996), and elastic resistance from the myoelectrically silent erectors (Watson et al. 1997). The mechanism for silencing these muscles during flexion is also believed to be the result of stretch receptor stimulation in the

posterior passive tissues, serving to reflexogenically inhibit paraspinal muscle activity (Kippers and Parker 1984).

2.3.4 Muscle recruitment strategies

Investigations into the function of trunk muscles have commonly used surface electromyography (sEMG) amplitude measurements to compare LBP subjects with healthy controls (Dankaerts et al. 2006). However, as described in Chapter 1, there are often contradictions in study findings, with examples of LBP groups demonstrating increased, decreased or indeed similar activity when compared to controls (Dankaerts et al. 2009; Arena et al. 1991; Alexiev et al. 1994; Cram and Engrstom 1986; Ahern et al. 1988; Cassisi et al. 1993; Kravitz et al. 1981; Cohen et al. 1986; Dankaerts et al. 2006; Sanchez-Zuriaga et al. 2015). These contradictory results are postulated to result from the heterogeneity of NSLBP groups, which is proposed to conceal subgroups (Dankaerts et al. 2006). In those that do find an objective difference, conclusions are typically based on an assumed lumbar stabilisation strategy in the LBP group. For example, Sanchez-Zuriaga et al. (2015) investigated normalised sEMG amplitudes of the erector spinae, and showed that activity during flexion was greater in the LBP group, suggesting a stabilisation strategy compensating for deficient generation of extensor moment by impaired lumbar structures (Sanchez-Zuriaga et al. 2015). The study found no difference between groups in gross lumbar ROM however, which suggests that if such activity is related to kinematics, then regional kinematic measurements are not adequate to highlight their relevance.

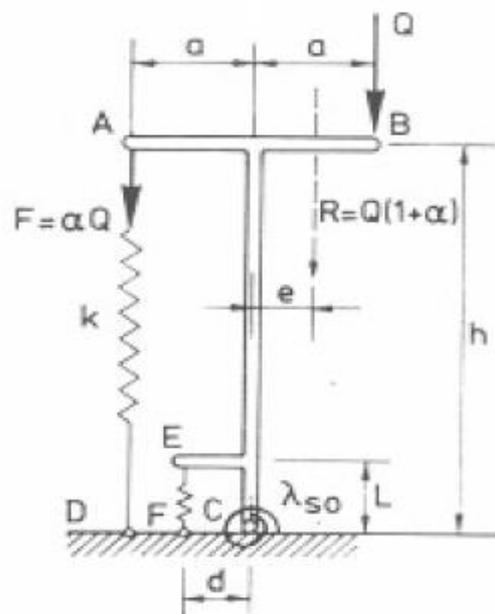
The type of strategy employed by the neural control system to maintain stability of the spine will depend on the task. If large perturbations are expected, then a strategy such as muscle co-contraction can be employed to maximise stability (Gardner-Morse 1998; Granata and Marras 2000; Oomen et al. 2015), however it can be costly both in terms of the energy required, and mechanically (Marras and Mirka 1990). Co-contraction involves the activation of both agonist and antagonist muscle groups, it has been shown to increase inter-vertebral stiffness (Stokes 2002), and is typically pre-emptive therefore reducing the likelihood of the need for reflexive inputs (Reeves et al. 2007). Tasks requiring a finer control however, may be better suited to a strategy more reliant on reflex pathways, as it may be more efficient in terms of stabilising during that specific task (Reeves et al. 2007). The movement of forward bending is likely to involve a mixture of strategies, although it has been demonstrated that there is little requirement for the activation of trunk agonists during the flexion phase without external

loading (Peach et al. 1998). Reeves et al. (2007) conclude that for any given task there is likely to be an optimal control strategy that maximises performance at the minimum metabolic cost (Reeves et al. 2007). Given the heterogeneity in each individual’s biomechanical make up (e.g. differences in the restraining properties of passive and active tissues and proprioceptive capacities within them), the strategies utilised are likely to be determined to some degree by the variation in these parameters. An insight into these strategies can be gained through examining relationships between and amongst active and passive system parameters.

2.3.5 Locally and globally acting muscles (stability)

As well as the co-contraction of agonist and antagonist muscles, another previously described strategy purported to increase spinal stiffness, is the preferential recruitment of locally acting muscles over synergistically acting global muscles (Van Dieen et al. 2003, Bergmark 1989). Bergmark suggests that globally and locally acting muscles will have fundamentally different roles (Figure 18). Globally acting muscles will balance the outer load in order for the force being transferred to the lumbar spine to be manageable for the locally acting muscles. This system means that large differences in outer load distribution result in only minor variations in resulting load on the lumbar spine (Bergmark 1989).

Figure 18: A simple spinal system viewed in the sagittal plane taken from Bergmark (1989)



Stability in the sagittal plane is maintained “when the total sagittal torque stiffness (λ_s) at the joint C, constituted by the passive torque stiffness in flexion extension and the local and global systems is greater than the critical value λ_{crit} .” (Bergmark 1989).

They go on to suggest that the locally acting control system is effectively dependent on the size of the outer load and the curvature of the spine. In the lumbar spine therefore the degree of lordosis is likely to be a key component (see section 2.4.1).

Bergmark's definition of locally and globally acting muscles is broadly similar to that of Bogduk (section 2.1.2.1). Any muscles (with the exception of psoas) with their origin or insertion at the vertebrae are defined as locally acting, and those that span between the thoracic cage and the pelvis as globally acting. Of the muscles described (section 2.1.2), the longissimus thoracis pars thoracis and iliocostalis group traverse the entire back and may be considered to act globally, as can the internal and external obliques, the rectus abdominis and the lateral quadratus lumborum. The longissimus thoracis pars lumborum traverses the lumbar spine, with upper lumbar region fibres that are particularly suited to extending the spine (or resisting flexion), and along with the lumbar multifidus that can also resist flexion and produce extension, may be considered to act at a (local) inter-segmental level.

An investigation into the relative effects of eccentric versus concentric contractions on the function of lumbar paraspinals, showed that repeated eccentric contractions (moving from 10° of extension to 40° of sagittal flexion) results in a higher level of multifidus activity required to produce a given level of torque production, suggesting that the multifidus become less efficient with repetitive loading (Herman and Barnes 2001). This was not the case for the iliocostalis lumborum muscle that appeared to be unaffected by the task. The reasons for such different characteristics between these muscles are of interest, as they provide an insight into the roles of each in terms of the task. Skeletal muscle consists of varying proportions of three fibre types. Type I fibres (also known as 'slow twitch', type IIA (also known as 'fast twitch oxidative') and type IIX (also known as 'fast twitch glycolytic') (Mannion 1999b). Generally speaking, type I fibres are more resistant to fatigue and are believed to be suited to postural functions, whilst types IIA and IIX being more prone to fatigue are better suited to fast, strong contractions. Whilst it has been suggested that muscle fibre type is unlikely to be the primary reason for the differences in muscle characteristics described above (Thorstensson and Carlson 1987), the inter-subject variation in paraspinal fibre type composition (Mannion 1999b) means that the extent of its influence cannot be fully known. As previously described the two muscles are believed to have different functional roles, multifidus as a local segmentally acting stabiliser, and iliocostalis acting more globally. Herman and Barnes (2001) suggest that the multifidus therefore may have to work harder and undergo relatively greater excursion than iliocostalis during the eccentric contraction required to bend through a relatively small range (Herman and Barnes 2001).

The sensory mechanisms outlined previously (section 2.3.3) afford an internal representation of the outside world, providing the information necessary to guide the movement. To an onlooker, the movement of bending forward appears to be a relatively straight forward task, which is also reflected in the relatively small size of the motor cortex believed to be involved in the control of the trunk musculature (Penfield and Rasmussen 1950). Movement control mechanisms are however to the contrary extremely complex, and using the example of bending forward, control of the spine involves numerous independently contracting muscles that insert on to vertebrae at multiple levels, with the capability of adjusting the moments produced at each. In view of the fact that several different muscle fascicles attach to each vertebra, it would seem logical that each fascicle may have the capacity to affect control at specific levels. A recent investigation using transcranial magnetic stimulation investigated the changes in motor cortical representation of different lumbar paraspinal muscle fascicles in people with LBP (Tsao et al. 2011). The study showed that in healthy controls there was a discrete cortical organisation of inputs to the LES and LMU muscles, which was lost in participants with LBP. In healthy individuals therefore, it is likely that the control system can utilise different muscle fascicles to optimise stability dependent on the conditions. The degree to which this can occur remains unclear however, as although individual fascicles can function independently of each other in a purely mechanical sense, connections that exist with extramuscular connective tissues, and indeed other muscles, shroud the relative contributions of each to the control system (Huijing 2003).

In terms of spinal motor control mechanisms, at an inter-vertebral level, current understanding is lacking. In a recent update to a text concerning musculoskeletal rehabilitation (Magee et al. 2015), state that locally acting muscles (e.g. multifidus) play an essential role in spinal control at an inter-vertebral level. To back up this statement they refer to an *in vitro* study that investigated the influence of different muscles on L4-L5 motion during flexion/extension (Wilke et al. 1995), and an *in vivo* cross-sectional study investigating deep and superficial multifidus activity during arm movements (Moseley et al. 2002). The former has limitations as an *in vitro* study, and the investigation concerned only a single inter-vertebral level. The latter study however, did not include the measurement of spinal movement at any level, and so any conclusions made regarding the role of multifidus in terms of inter-segmental control remain theoretical. Therefore, based on these studies, the suggestion that locally acting muscles are of key importance to control at an inter-vertebral level is so far unfounded. Further *in vivo* information is therefore required, and a capacity to collect concurrent inter-vertebral and multi-muscle data would be of value.

2.3.6 Muscle deactivation and the flexion relaxation phenomenon (FRP)

The flexion relaxation phenomenon is a temporal deactivation pattern that has received a great deal of attention in the literature (Dickey 2003; Kaigle 1998; Kippers and Parker 1984; Mathieu and Fortin 2000; McGill and Kippers 1994; McGorry et al. 2001; McGorry and Lin 2012; Neblett 2003; O'sullivan et al. 2006c; Olson et al. 2004; Sarti et al. 2001; Sihvonen et al. 1991; Steventon and Ng 1995). During sagittal flexion from a neutral standing position, the activity of the eccentrically contracting paraspinal muscles will increase in order to match the increasing moment arm of the head and torso. In the majority of healthy (back pain free) participants, it is believed that this muscle activity will increase until a point is reached at which an adequate extensor moment can be provided by the passive elements (Floyd and Silver 1955; Sihvonen et al. 1991), by other synergistically acting muscles (Andersson et al. 1996), and passive resistance from myoelectrically silent stretched muscles (Adams et al. 1980). This deactivation of paraspinal muscles is thought to be invoked by a stretch inhibition reflex (Kippers and Parker 1984), and in a review of spinal muscle activity literature, Demoulin et al. (2007) reported that sEMG in chronic LBP patients often shows an abolition of the FRP (Demoulin et al. 2007). Persistent muscle activation may therefore be a mechanism that the neuromuscular control system employs to protect diseased or damaged structures from reaching a point that would cause pain or further damage. As such, it has also been suggested that assessment of the FRP can be used as a clinical tool to assist in the diagnosis and treatment of LBP patients (Colloca and Hinrichs 2005). This theorised protection mechanism (i.e. persistent muscle activation) is also an indication that the function of paraspinal muscle activation in healthy individuals is primarily to restrain inter-vertebral movement.

The FRP has been shown to occur less frequently at more cephalad muscle sites. For example many studies analysing EMG recorded from sites lateral to T9 have shown that the FRP is absent in a large proportion of participants (Dolan and Adams 1993; McGill and Kippers 1994; McGorry et al. 2001; Toussaint 1995). A criticism of all of these studies is that they all measure gross kinematics (i.e. lumbar ROM) well below the level of EMG recording (i.e. around T9 spinous process), and so no insight into why this absence may arise (especially in terms of stretch reflex inhibition) can be found within them. Whilst the FRP is well researched, the synergistic activation timings (i.e onset) of paraspinals from different levels of the spine (i.e caudal to cephalid) are not so well described in the literature. This is likely due to the fact that a degree of underlying muscle activity acts as a postural control mechanism during standing (O'sullivan et al. 2002), and so calculations of onset times can be difficult.

A general criticism of all FRP studies that use gross lumbar measurements and record from a single electrode site, is that the results cannot account for inter-participant differences in terms of individual inter-segmental kinematics within the lumbar spine. At this point little is known about the capacity of muscles acting over multiple levels to control inter-vertebral movement at specific levels throughout the lumbar spine, and it is likely that for this reason the majority of studies choose an electrode location lateral to L3 (i.e. in the middle of the lumbar spine) as a site most likely to provide the broadest representation of activity (Sarti et al. 2001; Steventon and Ng 1995). The fact remains however that some individuals may bend differently to others (i.e. predominance of upper or lower lumbar or of specific inter-vertebral levels), and such movement patterns are likely to relate to the concurrent motor control patterns. To gain further insight, an approach incorporating multiple muscles and at an inter-segmental level is required.

There is much debate over the order of inter-vertebral movement during forward bending. In a small study (n = 8) using cineradiography, Kanayama et al. (1996) demonstrated that segmental motion was sequential, beginning in the upper levels and exhibiting a phase lag (a delay in movement) before the movement of the segment below (Kanayama et al. 1996). If this cascade of movement is representative of a larger population, then it is logical to assume that upper lumbar segments will complete rotation earlier than the lower levels and that motor control patterns will mirror such changes. Indeed, McGorry et al. (2001) showed that muscle activity at sites around L5 ended (i.e. FRP began) later than at sites around L2 and T12 (McGorry et al. 2001). In one of the very few studies to investigate FRP at an inter-segmental level, Kaigle et al. (1998) concluded that muscle deactivation occurs concurrently with the completion of segmental rotation (Kaigle et al. 1998), a finding that would appear to explain McGorry's finding if a cascading movement pattern is assumed. There is however much debate over how the spine moves during bending and a great deal of natural heterogeneity is likely, therefore no consensus has yet been reached.

There is also some confusion in the literature regarding the true definition of FRP. The majority consider FRP to be a pre-defined level of relative myoelectrical silence (MS) (Kippers and Parker 1984; Sarti et al. 2001; Dickey et al. 2003; Descarreaux et al. 2010), whereas some relate the phenomenon to the point at which deactivation begins (i.e. after which muscle activity declines towards total shut-off), and refer to this as the critical point (CP) (Kaigle et al. 1998; Sarti et al. 2001; Steventon and Ng 1995). O'Sullivan et al. 2006 reviewed the common ways by which FRP is determined. These included visual inspection (Kippers and Parker 1984), and a drop in the MVIC by a specified percentage (McGill and Kippers 1994). The obvious advantage of using

deactivation onset times (CP) as a variable, is that a methodology involving full sagittal flexion is not necessarily required. Kaigle et al. (1998) for example simply use the peak RMS EMG during the flexion phase of the cycle as the point after which RMS EMG will be in decline, and is therefore representative of the CP (Kaigle et al. 1998).

Kaigle et al's study concludes that the onset of muscle deactivation occurs when segmental rotation reaches 80% of its full rotation. Unfortunately due to the limited sample size, the multi-level inter-vertebral data were combined and so no conclusions regarding specific levels can be made. This is an important limitation, as combining the data effectively means that the study was not insightful about inter-vertebral relationships. Indeed to combine the levels is perhaps a major weakness, as the proportion of motion segment movement relative to total ROM is not known. FRP studies focus on the point at which deactivation begins or completes, but they do not provide any information about the relative size of deactivation. Although the most widely held view is that lumbar paraspinal deactivation begins at a point when most sagittal rotation has been completed, there are many gaps in the literature in terms of inter-segmental information, the feedback mechanisms involved and the specific muscles that become deactivated. Indeed the literature would also seem to suggest that as the paraspinal muscles primarily act as extensors, their principal role is to restrict sagittal rotation during bending, which would indicate that deactivation would have to occur before or concurrently with vertebral motion. If this were the case, then the size of the deactivation could feasibly be linked to the subsequent range of the movement. Bergmark's (1989) theories concerning the distinct roles of local and global muscles, provide a possible mechanism in terms of shared responsibility between muscle groups (Bergmark 1989), but a lack of current multi-level, multi-muscle information, warrants further investigation (McGill and Cholewicki 2001).

2.3.7 Selecting muscles for investigation

Due to methodological limitations (e.g. availability of specialist EMG equipment and expertise), investigations into spinal control mechanisms are typically restricted in terms of the number of muscles that can be analysed. Selecting the most appropriate muscles as a focus for study is not straight forward however, as motor control of the lumbar spine is accomplished through the co-activation of many different muscles, each with their own force capacities, geometries, and lines of action (Crommert et al. 2011). The action of muscles is task specific, and so the decision to investigate movement in the sagittal plane evidently affects the choice of recording sites. The paraspinals are an obvious choice due to their established role in the control of forward bending, and are included in the majority of studies investigating motor activity during this movement

(D'Hooge et al. 2013; Hashemirad et al. 2009; McGorry et al. 2001; Neblett et al. 2003; Reeves et al. 2006, Van Dieen et al. 2003), however other groups also warrant consideration.

The TrA for example has received special attention in the literature, due to its perceived role in spinal stabilisation (Barker et al. 2006; Crommert et al. 2011; Hodges et al. 2003). It has been shown to have an influence over the control of sagittal rotation during flexion (Barker et al. 2006; Hodges et al. 2003; Tesh 1987), with Barker et al. (2006) showing that it contributes to the size of the neutral zone, by increasing neutral zone stiffness during flexion (Barker et al. 2006). It is also believed to have a degree of direct control over trunk movements, being most active during flexion period, but demonstrating no sharp changes between flexion and extension phases (Cresswell et al. 1992).

Two of the aforementioned studies (Barker et al. 2006; Tesh 1987) were *in vitro* investigations, and control mechanisms provided by other muscles and passive tissues, combined with the effects of gravity and loading may well result in different findings *in vivo*. Hodges (2003) used sedated domestic pigs to electrically stimulate the TrA and provided the first *in vivo* evidence that TrA activity increases inter-vertebral stiffness (in combination with raised intra-abdominal pressure and diaphragm activity) (Hodges et al. 2003), however it has been shown that during full sagittal flexion intra-abdominal pressure is in fact zero (Hutton et al. 1979 the compressive strength of lumbar vertebrae). Bartelink (1957) suggested that intra-abdominal pressure could aid resistance to flexion moments via a "intra-abdominal balloon mechanism" (Bartelink 1957), but this idea has received criticism as the pressure required would likely exceed the capabilities of the abdominal muscles, be so high it would actually obstruct the aorta, and the required contractions would be so strong, they would actually increase the flexion moment due to the anatomical position of the muscles (Bogduk 1997). The potential for relationships between TrA and lumbar spine kinematics should therefore be recognised, but both practically and in consideration of more recent literature, its inclusion warrants careful deliberation.

Firstly, whilst being a muscle of potential relevance to lumbar spinal control, the TrA is the deepest abdominal muscle, and its assessment typically requires the use of fine wire intra-muscle electrodes (Crommert et al. 2011; Hodges 1998; Hodges and Richardson 1999). The typical insertion of the electrodes is also through the anterolateral trunk, (guided into place by real time ultrasound) creating additional time, ethical, training and equipment considerations.

Secondly, attention should also be given to the size and relative importance of the muscle. As the smallest of all the abdominal wall muscles, its relative contribution to the stabilisation of the

spine compared to that of the paraspinals and the obliques is minimal (Kavcic et al. 2004). The perceived importance of the muscle to researchers and clinicians alike has a foundation in studies reporting a delay in its activation during perturbation tasks in low back pain groups (Hodges et al. 1996), making it a potentially useful pathology marker and a target for rehabilitation. However since Hodges small study (n = 15) was conducted, a larger investigation (n = 96) using a similar methodology (i.e. rapid shoulder flexion) failed to demonstrate a significant delay in either control or LBP populations (Gubler et al. 2010), raising questions over conclusions regarding TrA's importance in lumbar spine stabilisation. Indeed in another similar study design, Silfies et al. (2009) reported on numerous trunk muscles and showed that the LBP group as a whole had delays in the activation of multifidus, erector spinae and external oblique, but not in the TrA. When subgrouping the LBP group into stable or non-stable (decided by an orthopaedic spine surgeon using discography and degeneration criteria consistent with segmental hypermobility), delays occurred most frequently in the non-stable group (Silfies et al. 2009). What these studies have in common is that they demonstrate that not all LBP patients demonstrate a delay in muscle activation in response to perturbation, and that the specific muscles showing a delay, may vary between individuals.

Perturbation methods such as exposure to rapid raising of limbs (Gubler et al. 2010; Hodges 1996; Silfies et al. 2009) or sudden loading (Cholewicki and Van Vliet 2002) have a valuable place in stability research, but they are not ordinarily exercises performed in everyday life. The notion that groups of individuals with and without low back pain both demonstrate varying degrees of muscle activation delay and in a number of different muscles, raises questions over the importance of their role (at least in isolation) in maintaining stability. Differences in muscle activation patterns between individuals are also going to be evident throughout contraction periods, and it could be argued that muscle activity changes that are more readily observable, and measurable during more common everyday tasks (such as forward bending) may also be of importance, especially if the accompanying inter-vertebral kinematics can be adequately recorded.

When there is discrepancy between the results of different study populations, there are two main possibilities for differences. The first is that the methodology is different producing distinct results (e.g. subtle differences in electrode application sites), and the second that there are additional differences between the populations studied, so that individuals in both LBP and control groups may respond uniquely as an individual, irrespective of their symptomatic group. It is feasible therefore that for many individuals experiencing LBP, the pain mechanism may not

be the main influence on their biomechanical behaviour during tasks. To understand other possible mechanisms better, it is first necessary to consider normal biomechanical interactions in the absence of pain.

2.3.8 The other abdominal muscles

While the abdominal muscles are believed to be co-activated in extension, lateral flexion and axial rotation movements (Thelen et al. 1995), they are not believed to have an important function in terms of producing sagittal flexion from a standing position (Olson et al. 2006). In fact the abdominal muscles have been shown to remain relatively inactive during the majority of the flexion phase, and would not be expected to demonstrate significant activity during flexion and return (Peach et al. 1998). Indeed Peach et al. (1998) demonstrated that Rectus abdominus, and the external and internal obliques produce less than 10% MVC during the flexion and return cycle (Peach et al. 1998). During a controlled bending task that does not reach full flexion therefore, it is unlikely that these muscles would demonstrate any significant activity, and subsequently would be less likely to be associated strongly with kinematic changes. Muscle selection also depends on the methodology chosen. In studies that restrain the pelvis for example, the recording of other potentially important controlling musculature such as the gluteus and hamstring groups (Nelson-Wong and Callaghan 2010) becomes less desirable, as their natural function is inhibited (see also section 9.11.3).

Section 4

2.4 Sagittal lumbar curvature and interactions within the passive spinal system

2.4.1 Lordosis

The curvature of the lumbar spine is designed to help maintain a stable posture with minimal energy cost, absorb spinal load and augment the efficiency of the surrounding musculature (Kim et al. 2006). It has been suggested that the shape and orientation of the lumbar vertebrae are intimately related, and can therefore influence the behaviour of those adjacent to them (Kim et al. 2006), and that the intrinsic shape of the lumbar spine can physically change throughout activities such as forward flexion (Pavlova et al. 2014).

In individuals with and without low back pain, there is much variation in the degree of lordotic curvature, which would suggest that biomechanical behaviour will also vary in accordance. Meakin and Aspden (2012) suggest that forces required to yield a follower load (i.e. “a

compressive load applied along a path that approximates the tangent to the curve of the column”) are increased as the lordosis increases and becomes more evenly distributed (Meakin and Aspden 2012), and is controlled by the local segmentally acting musculature (Patwardhan et al. 1999). Patwardhan et al. also suggest that in a normal spine, this follower load protects the spine from damaging fluctuations in curvature, and provides an increased capacity for load bearing (Patwardhan et al. 1999). The degree of initial lordosis will therefore influence subsequent biomechanical behaviours during forward bending, which will require different control mechanisms to optimise the movement. It seems logical therefore that biomechanical adaptation to variations in curvature may sometimes be sub-optimal and relate in some way to LBP. In a study analysing radiological parameters including lumbar lordosis in LBP patients and healthy controls, Tuzun et al. (1999) found no significant difference in lumbar lordosis between the groups (Tuzun et al. 1999). This was in agreement with Sarikaya et al. 2007 who when investigating the incidence of LBP in coal miners, similarly did not find any significant relationship (Sarikaya et al. 2007). Current understanding of the potential mechanisms involved is limited, and evidence for any direct association between LBP and lordosis remains inconclusive and the ideal lordotic range unknown (Been and Kalichman 2014). Its implications for intervertebral loading are intuitively powerful however, and its accessibility to measurement makes it attractive for inclusion in biomechanical studies⁴.

2.4.2 Sagittal Balance

The normal spine has lordotic curves in the cervical and lumbar regions and a kyphotic thoracic curve in between, allowing for the even distribution of forces throughout the spine (Roussouly and Nnadi 2010). Disruption to this state of equilibrium is referred to as sagittal imbalance, and has been widely linked with LBP. There are numerous causes of sagittal imbalance, however it is commonly associated with a loss of lumbar lordosis (Glassman et al. 2005; Jackson 1994; Le Huec et al. 2015), indeed, some believe that a loss of lordosis is the initial morphological change that leads to the development of sagittal imbalance (Le Huec et al. 2011). Normal sagittal alignment is typically taken as a plumb line descending from the centre of the C7 vertebral body to the posterosuperior corner of S1 (Jackson et al. 1994). In a retrospective review of LBP patients with adult spinal deformity, Glassman et al. (2005) found that a positive sagittal balance (i.e. anterior deviation of the C7 plumb line) was directly associated the severity of symptoms,

⁴ Note: In order to gain an initial insight into the influence of lordosis on segmental movement and to support the development of one of the study hypotheses (section 2.7.2.2), a preliminary study was conducted (Chapter 3, section 3.2).

and that kyphotic lumbar spines were associated with poor disability scores (Glassman et al. 2005). Therefore although lordosis in isolation has not been irrefutably linked to CNSLBP, it is intrinsically related to sagittal balance, and so relationships between lordosis and the inter-vertebral movement behaviours within it warrant further investigation.

2.4.3 Regions within the lumbar spine

More recently attention has been focussed on regions within the lumbar spine. It is well established that the lower lumbar region is not only the most common site of pain, but segments in this region also typically exhibit greater degenerative changes than those in the upper (Beattie et al. 2000; Biering-Sorensen 1983; Quack et al. 2007), therefore considering the lumbar spine as a homogenous region may provide little useful information about pain and function (Mitchell et al. 2008). It has been shown in cyclists for example, that LBP groups can display comparatively less multifidus activity and greater flexion at the lower lumbar levels than healthy controls (Burnett 2004), suggesting that flexion strain and excessive IV-RoM may be possible pain generators. In a study investigating regional differences in lumbar posture in a group of nursing students with and without LBP, Mitchell et al. (2008) used the 3-Space Fastrak to calculate sagittal angles between T12 and L3 (i.e. the upper lumbar spine) and L3 a S2 (i.e. the lower lumbar spine). They found that although LBP was not associated with regional lumbar spine angles or ROM, an inverse correlation (-0.422, p value <0.001) existed between upper and lower lumbar angle ranges from standing to full flexion, suggestive of a compensatory function existing between regions (Mitchell et al. 2008).

2.4.4 Sub-groups

Although the findings of Mitchell et al. were not found to be related to LBP, the problem of heterogeneity within NSLBP groups, means that such relationships cannot be dismissed completely. In an attempt to sub-group NSLBP patients, O'Sullivan et al. (2005) devised a strategy to classify patients based on the direction of pain provoking movement (O'Sullivan et al. 2005). Perhaps the most widely studied sub-group (due to the frequency of occurrence) is a group whose pain is reproduced in tasks involving sagittal flexion. O'Sullivan et al. (2006) showed that in 'normal' sitting, this particular group postured their lumbar spines significantly closer to their end-range than health controls (O'sullivan et al. 2006a), suggesting that increased ROM may be a predisposing factor.

In agreement, Dankaerts et al. (2006) found that when sub-grouped in such a way, differences could be found between LBP patients and healthy controls in sitting postures, when considering upper and lower lumbar angles as distinct entities, relating flexion-provoked pain to an increased kyphosis in the lower lumbar (Dankaerts et al. 2006). Hemming et al (2015) also

looked at differences in regional curvature of the spine during tasks using O'Sullivan's subgroups. Their results showed no difference between groups in the lower lumbar curvature, but the upper lumbar and lower thoracic sections demonstrated greater flexion compared to controls (Hemming et al. 2015). The contrast between these findings may be explained by differences in methodology (e.g. sitting versus standing during tasks performed), however, due to the problems associated with skin movement, concerns are also raised over the reliability of surface marker measurements of regional areas, especially in the lower lumbar. Studies have therefore established that differences in regional lumbar spine kinematics can relate to specific sub-groups of LBP patient, but this still does not provide a clear biomechanical explanation as to why this is the case. Indeed Mitchell et al. (2008) state that global lumbar spine kinematics (i.e. movement of the entire lumbar spine) do not reflect regional lumbar spine kinematics (i.e. movements of regions within the lumbar spine) (Mitchell et al. 2008), and taking this a step further, it is also likely that regional kinematics do not reflect inter-vertebral kinematics. A more complete understanding will require even more detailed information, and so inter-vertebral level data in the absence of pain seems like a logical progression.

2.4.5 Inter-segmental interactions

The effects of changes to stiffness and restraint at an inter-vertebral level have been considered in the spinal surgery literature. A lumbar spinal fusion will aim to stabilise one or more motion segments, however whilst the aim of the surgery is usually to stabilise (i.e. increase stiffening) it will also have a biomechanical impact on the segments adjacent to them, effectively necessitating a redistribution of mobility within the lumbar spine (Lee and Langrana 1984). It is proposed that juxta-fused motion segments will become more mobile in way of compensation, which subsequently leads to degenerative changes within them (Chow et al. 1996; Lee 1988; Scannell and McGill 2003; Untch 2004; Xia et al. 2013). In a study investigating risk factors for the development of such adjacent segment disease (ASD) post spinal fusion surgery, Rothenfluh et al. (2015) identify a combined high pelvic incidence and flattened lordosis as predisposing factors in the development of the condition (Rothenfluh et al. 2015). Although Rothenfluh et al (2015) do not report on specific levels of fusion, as discussed previously, it is widely accepted that spinal fusion is most often performed at the lower lumbar levels (e.g. L4-L5, L5-S1) (Le Huec et al. 2015). If we assume this to be the case, then a shallow lordosis and stiffening of the lower lumbar segments is compensated for by an increased ROM in superior motion segments. In order to maintain stability and perform movements optimally, it is likely these kinds of compensation mechanisms will also exist in healthy spines; however there is currently limited research in this area.

Cadaveric studies have also demonstrated that the invasive nature of lumbar spinal surgery can affect the integrity of passive system structures (e.g. ligaments and discs), with additional (and opposing) consequences for spinal stability (Chamoli et al. 2015). In their study, Chamoli et al. (2015) investigated the impact of interspinous and supraspinous ligament transection and bilateral facetectomy (representing a graduated decrease in the capacity to restrain inter-vertebral movement) at L4-5 on the sagittal motion of this segment and both caudal and cephalad segments. The results showed that sagittal ROM increased at the damaged segment, but was decreased in adjacent segments, suggestive of a multi-segmental compensation mechanism to perform the overall movement (Chamoli et al. 2015). Indeed the study also found that the changes in passive structure integrity had no significant impact on global sagittal kinematics, highlighting the compensation mechanisms within the lumbar spine, and also raising questions over the suitability of using gross lumbar measurements as a kinematic parameter in general.

The research therefore shows that an increase or decrease in motion segment stiffness as a consequence of lumbar spinal surgery can influence the behaviour of neighbouring segments. The heterogeneity seen in IV-RoM at different inter-vertebral levels (Deitz 2011) in healthy controls, may therefore represent the capacity of individuals to compensate for variations in stiffness naturally occurring within their spines. In terms of IV-RoM, it may be expected that a relatively large IV-RoMmax at one level of the lumbar spine will be compensated for by relatively reduced movement at another. If there is laxity within a motion segment, it may be expected that the IV-RoMmax at that level will be relatively large in accordance; however the IV-RoMmax at other levels may be expected to be reduced as a compensatory mechanism. Such interactions between variables that have influence over inter-vertebral restraint have never been examined in healthy controls, and are likely not to be restricted to the influence of adjacent levels but interactions throughout the whole lumbar spine.

Of course increased IV-RoM (e.g. hypermobility) is not the only proposed risk factor for developing conditions such as ASD. Others include age, gender and patient weight (Park et al. 2004), pre-existing degenerative changes (Lee et al. 2009), the number of segments fused (Gillet 2003), and post-operative disc height (Kaito et al. 2010) (i.e. induction of early degeneration of the adjacent segment after posterior lumbar interbody fusion). These factors are all potentially confounding, and the complexity of possible biomechanical interactions can lead to contradictions in the findings of studies (Rothenfluh et al. 2015), especially in diseased or surgically altered spines. Investigating interactions in a healthy population will minimise the

impact of many of these variables, and provide a clearer insight into underlying interactions with IV-RoMmax.

2.4.6 Should researchers persist with the investigation of IV-RoM?

How best to determine abnormal motion or instability is a subject of continuing debate, with a deficiency of evidence for methods that profess how to do so (Hicks et al. 2003; Steiger et al. 2014). However, intuitively, increases in IV-RoM and translation should relate to instability, leading to attempts to determine abnormal limits in these parameters, typically with the use of functional radiographs (Abbott et al. 2006; Boden 1990; Bridwell et al. 1993; Dvorak 1991; Hayes 1989; Wood 1994). Bridwell et al. (1993) for example defined instability as a difference of ≥ 3 mm translation or $>10^\circ$ angle difference between flexion and extension radiographs (Bridwell 1993). The usefulness of such diagnostic criteria can be questioned however due to the significant heterogeneity that exists in populations without LBP (Hayes 1989; Panjabi 1994), and have led to recommendations to explore characteristics such as motor control and the neutral zone, as alternative indicators of instability (Hicks et al. 2003).

Perhaps unsurprisingly therefore, the value of ranges of spinal motion as a means of differentiating between the kinematics of healthy controls and LBP patients has also been questioned (Lehman et al. 2004; McGregor et al. 1997), its limited capacity to differentiate, again at least explained in part, by the large inter-and intra-subject variation found in such groups. A large range of normal population values does however make ROM appealing in terms of the investigation of normal lumbar biomechanics, as interactions and relationships are likely to be clearer when there is greater variation in the outcome measure. If we use the example of ROM and muscle activity during sagittal flexion, eccentric contraction of the paraspinals will provide much of the extension moment required to control forward momentum. At an inter-vertebral level, as ROM increases it is logical that concurrent muscle activity will decrease in order for the inter-vertebral rotation to occur. The opposite may be the case for a decrease in ROM as muscle activity increases in association with the relative lack of movement (Kuriyama and Ito 2005). This knowledge regarding control mechanisms of particular muscle groups, compensation mechanisms between muscles and the influence of activity at specific inter-vertebral levels is lacking in normal populations.

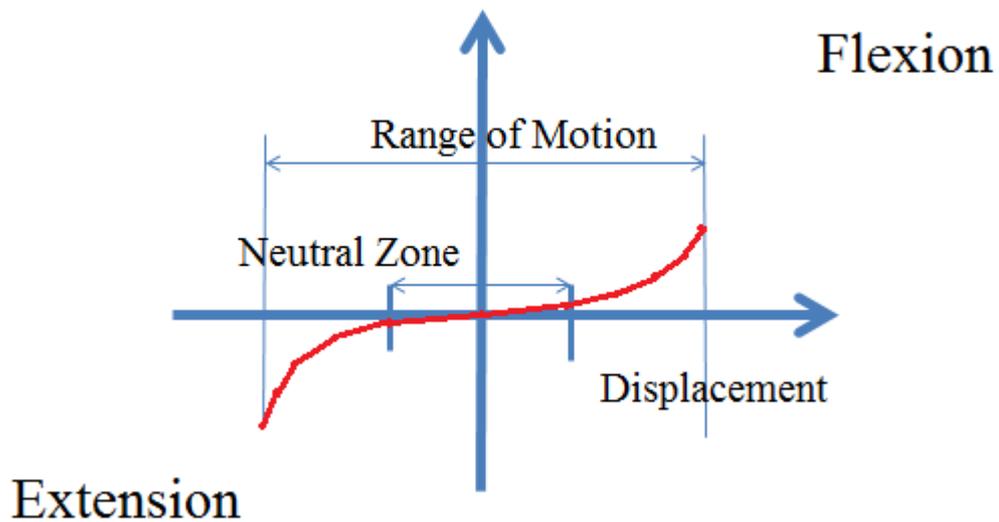
ROM is frequently used as an outcome measure in biomechanical studies, and as a value for direct comparison between groups. What is not typically investigated are numerous factors that are associated with angular range, that themselves may be useful indicators of dysfunction. To date, there have been relatively few attempts to investigate these links, which needs to be done

first in healthy controls. ROM's perceived lack of objective use, has however lead researchers to consider the use of other variables. Mieritz et al. (2012), suggests that measuring parameters of higher order kinematic motion (e.g. acceleration and velocity) may be relevant (Mieritz et al. 2012). In order to do so, continuous measurement techniques are typically required, and so technologies such fluoroscopy are increasingly being used, and so called higher order parameters are being developed (Breen et al. 2015; Mellor et al. 2014; Teyhen et al. 2005; Wong et al. 2006).

2.4.7 Neutral zone, initial attainment rate and spinal RoM

It has been proposed that total IV-ROM comprises of a neutral zone (Figure 19) and an elastic zone (Panjabi 1992a; Panjabi 1992b). The neutral zone is purported as the flexible section of total ROM, where there is minimal resistance to motion provided by the passive structures, and the elastic zone is believed to be a section closer to the end of total ROM, where there is significant resistance provided by the passive structures (Hicks et al. 2003). In terms of spinal stability therefore, decreased motion segment stiffness may lead to relatively larger angular ranges, and increased stiffness may result in reduced rotation required for given moments (Hodges et al. 2013). This idea is supported by research that examines the effect of spinal degeneration on spinal stiffness. Kirkaldy-Willis and Farfan (1982) divide the spectrum of lumbar spinal degenerative change into three stages, (1) temporary dysfunction, (2) unstable phase and (3) stabilisation (Kirkaldy-Willis and Farfan 1982). In the unstable phase it is assumed that damage to the structures of the passive system (i.e. discs, ligaments and facet capsules) will result in a decrease in inter-vertebral stiffness. This is supported by the work of Panjabi et al. (1984) who demonstrated that by purposely injuring discs, IV-RoM was increased for a given moment (Panjabi et al. 1984), and by investigations into biological disc degeneration, that in addition to increased IV-RoM, also demonstrated a decreased stiffness and an increased neutral zone associated with the progression of degeneration (Gay et al. 2008; Muriuki et al. 2016). However McGregor et al. (1997), analysing the relationship between degenerative disc disease (DDD) and lumbar motion characteristics, found that LBP patients with signs of DDD, actually showed a reduction in lumbar ROM of motion compared to controls (McGregor et al. 1997). This study used regional measurements however, and the level of disc degeneration was not described. It may be therefore that specific levels of hypermobility were compensated for by increased restraint at other levels within the lumbar spine, or that the DDD was in a late stage in the majority of participants (i.e. stabilisation stage (Kirkaldy-Willis and Farfan 1982)), which would also explain such movement behaviours.

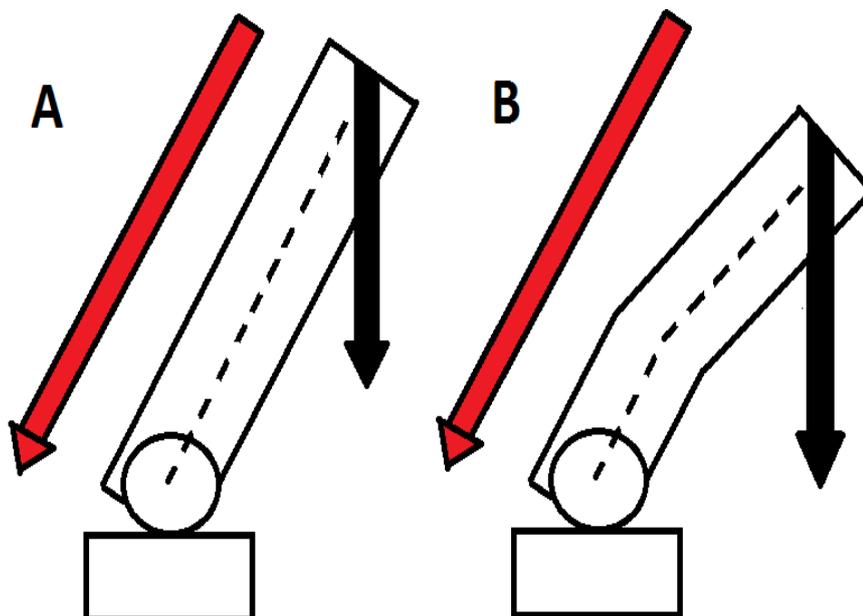
Figure 19: Panjabi's Neutral Zone (NZ) adapted from Panjabi et al. (1994)



Panjabi showed that motion segments exhibit non-linear load displacement curves. This suggests that there is a changing relationship between the applied load and the displacements produced.

It has been suggested that most likely an approximate motor control strategy (i.e. the exact vertebral positions are not vitally important as the system can account for substantial margins) of spinal equilibrium is used (Cholewicki and McGill 1996; Kingma et al. 2007), and therefore decreased stiffness at a particular inter-vertebral joint will result in increased angular rotation before equilibrium is reached (Hodges et al. 2013). In a degenerative spine this could lead to disproportionate rotation and what could be termed instability, and in a healthy spine, this could equally lead to relatively larger IV-RoMs at a particular segment (Mahato 2013), complicating optimal spinal control strategies. Indeed approximate control may deal with homogenous change (changes in stiffness throughout the lumbar spine) quite well, however heterogenous changes (i.e. decreased stiffness at a particular level) may lead to further increases in range, as bending moments would be focussed at that level (Hodges et al. 2013) (Figure 20). This relates to the previously described theories regarding lordosis and follower load, as if the force vector established by the follower load alters from its optimum course, then the local motor control strategy will have to adapt to maintain stability (Preuss et al. 2005).

Figure 20: Loading of a crane (analogous to the loading of a spine) (adapted from Hodges et al. 2013)



Loading of a crane is analogous to lumbar sagittal bending. The black arrow represents loading as a result of gravity. The red arrow represents the muscle activity force required to maintain equilibrium around the centre of rotation (i.e. the circle). The crane's beam is loaded by a bending moment resulting in a bend to the right. If there is an area in the beam with reduced stiffness, it will buckle (B). The bending moment of the structure will be greatest at the point of buckling, and subsequently buckling will tend to increase further at this point.

Attainment rate, defined as the ratio of the gradient of the motion segment over 10° increments (using the change in lordosis as the global ROM) (Teyhen et al. 2007), and initial attainment rate, defined as 'the ratio of the initial gradient of the segment over the first 10° of rotation' (Mellor 2014) have been investigated *in vivo*, and are suggested to represent laxity within the motion segment. Indeed, initial attainment rate has been shown in frontal plane movements to be representative of the neutral zone (Breen et al. 2015). Although the calculation of initial attainment rate has been used in recumbent protocols (Mellor 2014), it cannot be interpreted the same way in a weight-bearing protocol due to the addition of muscle activity and load bearing. Muscle activity will be present from the onset of flexion during weight-bearing protocols, and the only previous study to describe attainment rate doing so, failed to consider the likely influence of muscle activity on this variable in their discussion (Teyhen et al. 2007). Indeed, the neutral zone is a concept traditionally associated with the *in vitro* spine (i.e. with no contribution of muscle control) where details of forces and loading can be identified. Investigations into spinal movement during weight bearing tasks, may however consider initial attainment rate as an indication of laxity in the presence of muscle activity. In terms of stability, it can be considered a representation of damping at an inter-vertebral level. As such, an

increased initial attainment rate would relate to increased damping in both the active and passive subsystems, which may have a positive influence on inter-vertebral stiffness. This is counterintuitive however as discussed previously, a relatively large neutral zone is believed to relate to instability and LBP. Some authors suggest that this can be managed through trunk muscle training (Sun et al. 2006), which in itself suggests that damping has only a negligible effect on the overall restraint properties of the spine. However, in their study, Sun et al. (2006) used an RCT to examine the effectiveness of a training programme (designed to control the lumbar neutral zone) on LBP outcomes, but they did not physically measure the neutral zone (Sun et al. 2006). Their conclusions therefore are based on the theoretical impact of their exercise, which feasibly could also influence factors such as IV-RoM. That said, the neutral zone (Panjabi 2003) and attainment rate (Teyhen et al. 2007) have both been shown to be more sensitive than IV-RoM as indicators of LBP, and the neutral zone is purported to be more sensitive in terms of stability (O'sullivan 2000; Oxland 1992). At this time, there do not appear to be any studies that have looked directly at how a change in such laxity parameters may influence IV-ROM either amongst or between levels. If such interactions can be found, they would provide new insight into the possible importance of both factors (i.e. laxity and IV-RoM). A simple hypothesis would be that the greater the laxity the larger the IV-RoM at the associated motion segment, with extremes compensated for by variations in laxity and IV-RoM elsewhere.

The true value of IV-ROM and its associations with other mechanical parameters has yet to be fully examined, in order to measure ROM and other parameters associated with it, numerous technologies can be used. Section 2.5 provides an overview of the techniques currently available.

2.4.8 Force deformation

Forces acting on the spine generally derive from muscle contraction and gravity. According to Newton's third law, such forces will be opposed by equal and opposite forces, which in the case of the spine, are provided by its active and passive elements. These loads and the resulting displacements within the spine can be thought of as the stress (defined as force per unit area) and strain (defined as the percentage change in length of a material relative to its original length) of the system respectively (Chang et al. 2011). These force deformation characteristics are however difficult to measure *in vivo* (Shirazi-Adl 1986), and therefore investigations typically rely on either *in vitro* studies (Panjabi et al. 1989), or estimations from computer based modelling (Shirazi-Adl et al. 2005; Wong et al. 2011; Zander et al. 2001). This means that studies investigating spinal biomechanics *in vivo* are typically limited in terms of kinetic data, and will tend to focus on kinematics. This type of information is still of value however, and initial

attainment rate is an example of such a kinematic measurement that can be considered to be affected by the strain on a motion segment.

2.4.9 Spinal modelling: reductionist and systems approaches

Spinal modelling is typically used to provide estimates regarding spinal kinetic information when this is not feasible to obtain through *in vivo* investigation. Two commonly used methods are the reductionist and systems approaches. In terms of lumbar spine pathomechanics, the major problem with the use of reductionist models is the process of reduction itself. Reductionism discounts component-component interactions and their subsequent dynamics (Ahn et al. 2006), and so the complexity of the lumbar spine more naturally lends itself to a systems based approach when studied. The reductionist approach does however have its advantages, particularly when only one or two components have a large impact on the system's behaviour. If such elements could be found then a focussed and appropriate response could be applied. This may however lead to treatments based on single factors, and as has been shown, an understanding of the consequences of changing individual parameters is necessary, due to the multi-level interactions that take place in the lumbar spine.

Systems approaches are perhaps more suited to the study of complex problems, as they are not only capable of incorporating numerous different elements, but provide insight into how such different parts interact, and affect the behaviour of the entire system (Ahn et al. 2006). An advantage would be gained therefore if data collected could include information from multiple segments. If this data were to be used in systems modelling for example, it would represent a shift to more complex models. The use of multi-variable, multi-level and multi-muscle data would be very complicated, however this information would lead to improved accuracy of such models, and move away from the simpler 'single hinge' type inputs. Indeed, in terms of kinematic inputs, more detailed modelling using fluoroscopic inter-vertebral data is already underway (Putzer et al. 2016), however such studies are still constrained by limited inter-vertebral information (e.g. regarding stiffness) and rely on inputs extrapolated from a narrow previous work base.

Section 5

2.5 Measuring lumbar spine kinematics and muscle activity

2.5.1 Surface measures

There are many different methods available to measure the kinematics of the lumbar spine. The least invasive techniques usually involve apparatus applied to the skin surface, including the use of flexible rulers (Stokes 1987), goniometers (Boocock 1994; Natrass 1999) and reflective markers with photography (Straker et al. 2009). Technological advances have seen the introduction of motion analysis systems that enable the assessment of all three planes such as Fastrak (Abdoli-E and Stevenson 2008; Burnett 2004), Flock of birds (Bull and McGregor 2000; Butler et al. 2009; Hsu et al. 2008), Optotrak (Nelson-Wong et al. 2012; O'shaughnessy et al. 2013), and multi-camera systems (Buchecker et al. 2013; Preuss and Fung 2008). These along with novel methods including strain gauges (O'sullivan et al. 2012; Van Hoof et al. 2012) and inertial sensors (Williams et al. 2013a), all measure movement over a region of the spine, but have limitations when investigating segmental inter-vertebral motion. Even low-tech clinical measurements such as Schober's test can be used to assess lumbar range of motion (Steele et al. 2013), but their correlation has been shown to be poor when compared to radiographical analysis (Rezvani et al. 2012).

2.5.2 Inter-vertebral measures

An ability to measure inter-vertebral motion provides an essential means of furthering our understanding of spinal biomechanics both in terms of IV-RoM but also higher order kinematic variables such as attainment rate (Lehman et al. 2004). Historically, the more accurate the required measurement, the more invasive the technique. The use of radiographs in healthy participants has been approved in the past (Ogston 1986b), but are limited due to the associated radiation dose exposure, and typically only provide data from particular sections of the motion sequence such as neutral, full flexion or full extension (Pieper et al. 2013). Uni-planar continuous radiographic techniques have also been used (Ahmadi et al. 2009; Harada 2000; Kanayama 1998; Okawa 1998; Wong 2006), but are typically limited by the same issue. Bi-planar techniques have been used to retrieve kinematic data from all 3 planes of movement (Appendix B) (Li 2009; Passias 2011; Percy 1984a), and whilst early techniques were associated with high measurement error, a technique called Roentgen Stereophotogrammetry that required the insertion of tiny metal spheres into the vertebrae, reported precision of $<3^\circ$ in all planes (Olsson 1977). Recent advances using dynamic stereo x-ray imaging are showing considerable promise (Aiyangar et al. 2014; Anderst et al. 2008; Wu et al. 2014), especially when investigating planes of motion where significant out of plane or coupled movements are anticipated. The highly

invasive surgical insertion of intraosseous pins into a participant's spinous processes (Kaigle 1998; Steffen 1997) whilst providing a solution, is ethically questionable and never likely to be widely incorporated. Computed tomography (CT) and magnetic resonance imaging (MRI) have also been used to measure lumbar spinal kinematics (McGregor et al. 2002; Ochia 2006; Xia 2009). CT scans can only currently be taken in the recumbent position and require an even greater radiation dose than traditional radiographic techniques. They therefore have limited use in terms of measuring large sagittal or coronal plane movements, however they are particularly useful when detailed information about axial rotation is required (Ochia 2006; Rogers 2005; Singer 1989; Zuhlke 2009). MRI has also traditionally been limited to recumbent imaging and is therefore well suited to axial plane measurements, however the increasing availability of open MRI scanners means that weight-bearing studies are becoming more common in other planes (Beneck et al. 2005; McGregor et al. 2002; Rodriguez-Soto et al. 2013). These studies are all still limited in terms of cost however and are still too slow to provide continuous images, and therefore are also restricted to pre-determined sections of a motion sequence or movement. Advanced MRI techniques have demonstrated that image acquisition times can be reduced to fractions of a second (Uecker 2010), but these are unlikely to be widely available in the foreseeable future. The use of ultrasound as a measurement tool has been investigated (Chleboun et al. 2012), but whilst providing a relatively inexpensive and widely available alternative, it again does not provide continuous data throughout the motion sequence.

Quantitative fluoroscopy (QF) is a technique using x-rays to obtain continuous moving images, and is therefore well suited to the investigation of spinal motion (Du Rose and Breen 2016a; Du Rose and Breen 2016b; Harvey et al. 2015; Mellor 2009; Mellor F.E. et al. 2014; Teyhen 2005; Teyhen et al. 2007; Wong 2006). Fluoroscopy has traditionally been hindered by high radiation levels. Advances in image intensification, digital magnification, automatic dose control and pulsing of the beam in synchrony with the camera, have however all contributed to reducing radiation dose levels to the extent that a complete motion sequence of the lumbar spine now requires less radiation than a single traditional radiograph (Mellor et al. 2014). Improvements in QF have also been made by addressing common sources of error such as subject positioning, the digitisation process, image distortion and movement coupling. The standardisation of many of these elements has been outlined (Breen A.C. et al. 2012).

2.5.3 Strengths and weaknesses of lumbar kinematic measures

Each technique has its own limitations, be it cost, invasiveness, complexity of methodology or analysis, not providing continuous data, or providing global and not segmental information. In order to improve understanding of spinal biomechanics at a segmental level, numerous methods are currently used as research tools. The array of techniques causes a problem when comparing values between them. This problem was identified by Mannion (1999), who suggested that a comparison of values with a 'gold standard' is required in order to see which device best reflects true vertebral movement (Mannion 1999a). Whilst there are currently questions regarding how well QF represents true movement (as the protocol restrains the pelvis), it is feasible that this modality could meet this requirement. QF is relatively low cost, low radiation dose and provides an ability to analyse continual inter-vertebral movement, and therefore able to extract information from the mid-range (Taghipour-Darzi et al. 2012), the end range, and higher order variables (Lehman 2004), satisfying the requirements outlined previously. QF has never been used concurrently with sEMG before, and the combination of the two would provide unique insights into the relationships between the lumbar spine's kinematics and muscular activity.

2.5.4 Agreement, reliability and accuracy of spinal movement measurements

The development of computer assisted inter-vertebral measurement has led to the improvement in reliability of such measurements when compared to manual techniques (Pearson et al. 2011). However, although many modern kinematic (and EMG for that matter) measurements do now utilise computer software programmes, the processing cannot be considered error free, especially when there are human controlled inputs. It is therefore important to assess the agreement and reliability of such measurements, terms which have historically, and incorrectly, been used interchangeably (De Vet 2006). This confusion was addressed by Kottner and Streiner (2011) who defined agreement as whether measurements are identical or similar, or the degree to which they differ, and reliability as the ratio of variability between measurements in the same subjects, to the total variability of all measurements in the sample (Kottner and Streiner 2011). Reliability coefficients give an indication of how well individuals can be differentiated from each other, while incorporating the measurement error in their calculation.

RoM is a commonly investigated variable and has been tested for its reliability when measured by many different modalities. Mieritz et al. (2014) for example examined the reliability of regional lumbar sagittal movements using the CA6000 spine motion analyser (i.e. skin surface measurements) and despite finding reasonable reliability (ICC 1,1 = 0.51-0.70), the limits of agreement (LOA) were considered too large for the comparison of individuals (Mieritz et al.

2014). The problem with using instruments attached to the skin is that variation may arise from the measurement device itself, the participant, the examiner and the interface between the instrument and the participant (Mayer 1997). Despite this, the majority of studies report good accuracy and precision (McGregor et al. 1995; Schuit 1997; Troke 1996), however there are also examples where poor accuracy has been reported (Christensen 1999). Schuit et al (1997), demonstrated that intra-examiner reliability of regional lumbar flexion measurements was excellent (ICC 2,1 = 0.875-0.966), and despite agreement being poor (SEM = 3.7°) concluded that the measurement method had acceptable validity when compared to a radiographic technique (Schuit 1997). Such large errors may be acceptable when comparing regional measurements, but they are not be adequate when investigating inter-vertebral ranges. Indeed, whilst it has been suggested that the use of skin surface markers for inter-vertebral measurements provide a reasonable reflection of lumbar inter-vertebral motion (Gracovetsky 1995), others disagree and suggest that such measurements are significantly different to methods that can measure internal vertebral kinematics (Zhang 2003). Indeed, it has been shown that in terms of inter-vertebral measurements, skin surface devices have comparably lower reliability (Mannion et al. 2004).

Fluoroscopy has been demonstrated as an accurate and reliable technique to measure continuous lumbar inter-vertebral motion (i.e. IV-RoM) (Ahmadi et al. 2009; Breen et al. 2006; Mellor F.E. et al. 2014; Teyhen 2005; Yeager et al. 2014). There are however areas where reliability has yet to be shown. For example, studies that have previously investigated agreement and reliability of these measurements have either pooled inter-segmental levels (Yeager et al. 2014), have not included all the lumbar inter-segmental levels (Mellor F.E. et al. 2014; Teyhen 2005), have not used continuous data to find the IV-RoMmax (Ahmadi et al. 2009; Teyhen 2005) or have not conducted inter-examiner studies (Ahmadi et al. 2009; Teyhen 2005). This information is also lacking for mid-range variables such as attainment rate. Indeed, the agreement and reliability of lumbar weight-bearing initial attainment rate measurements has never been investigated. Initial attainment rate agreement and reliability has been investigated previously in the cervical spine (Branney 2014), and in the recumbent lumbar spine (Mellor 2009), but never at the level of L5-S1.

2.5.5 Electromyography (EMG)

"Electromyography (EMG) is a technique concerned with the development, recording and analysis of myoelectric signals. Myoelectric signals are formed by physiological variations in the state of muscle fibre membranes." (Basmajian and De Luca 1985)

The degree of electrical excitation is believed to be highly correlated with the size of the muscle contraction (Merletti et al. 1992), and can therefore be used to interpret muscle activity and fatigue. As such electromyography can be used to investigate the role of active tissues in lumbar function, through examination of the various myoelectric signal components. These can be separated into amplitude (Butler et al. 2009; De Nooij et al. 2009; Van Dieen 2003), frequency (Abboud et al. 2016; Lariviere et al. 2001; Mannion and Dolan 1994) and timing related parameters (Hodges and Bui 1996; Kuriki et al. 2011; Williams et al. 2013b).

2.5.5.1 EMG amplitude

The EMG signal is most commonly analysed using amplitude parameters (Kollmitzer et al. 1999). Due to the bipolar nature of raw EMG signal data, raw EMG has a mean value of zero, and in order for the amplitude of a signal to be interpreted, a process of rectification is required that converts all negative amplitudes to positive. This is a precondition to EMG amplitude parameter calculations, which include mean, peak, area and slope. As described above, EMG amplitude relates to the force a muscle generates and therefore provides a quantifiable insight into muscle function. Of the amplitude parameters, the mean amplitude value is widely regarded as the most useful, as it is less sensitive to duration differences in analysis periods. It also provides the most reliable reflection of muscle activity for a given task, and is recommended for use when comparisons between individuals are required (Konrad 2006).

When a muscle activates, motor unit action potentials will superimpose in an arbitrary manner, and so no two bursts of EMG will ever be the same. As a solution to this problem, smoothing algorithms that outline the mean trend of signal development can be applied (Konrad 2006), and of all the options available, root mean square (RMS) EMG has been demonstrated to be the most reliable (Basmajian and De Luca 1985).

2.5.5.2 EMG signal frequency

Modern EMG analysis software also makes it possible to analyse the frequency content of a signal (Lariviere et al. 2000). Fast fourier transformations (FFT) can be used to analyse the frequency content of EMG signals (Konrad 2006), changes in the distribution of which can be

helpful when evaluating the level of fatigue developed during muscle contraction (Abboud et al. 2016; Mannion and Dolan 1994). Generally speaking, a fatiguing muscle will correlate with EMG signal recordings that have shifted towards lower frequencies (Merletti et al. 1992). The most commonly used frequency parameters are known as the mean and median frequency, which represent the mathematical mean of the spectrum curve and the division of Total Power (the integral under the spectrum curve) area into two uniform sections (Konrad 2006).

2.5.5.3 Temporal activation patterns: EMG signal onset and offset

It is also possible to determine temporal parameters such as the point of signal onset and offset (Hodges and Bui 1996). There is much debate over how best to determine precise muscle activation onset and offset times, but it can be important if dealing with narrow time differences when comparing muscles, subjects or subject groups. In the majority of early studies evaluating temporal parameters of EMG there was usually no description of how EMG onset/offset was determined, and when the method was described it was usually performed by a visual evaluation of the EMG trace with no mention of the criteria by which this decision was made (Hodges and Bui 1996). Typically a threshold value will be determined e.g. 1, 2, or 3, standard deviations above the mean baseline activity, or 15-20% of the peak EMG, often associated with a minimum time duration during which the signal must remain over the threshold (Hug 2011). Leinonen et al. (2000) for example defined onset as the minimum of a +10 μ v deviation from the baseline (Leinonen et al. 2000). Another simple method was demonstrated by Li and Caldwell (1998) who used a threshold of 25% of maximum amplitude during a cycle to indicate muscle onset (Li and Caldwell 1998). The 'correct' threshold to be employed however remains an area of contention amongst investigators (Hodges and Bui 1996). It should be acknowledged however that with a sufficiently clean sEMG signal, traditional approaches can achieve comparable results to the more complex (Zhang and Zhou 2012). In such an environment it is possible to determine the onset and termination of muscle activity by using the on/off methodology by visual interpretation (Sanchez-Zuriaga et al. 2010; Worsley et al. 2013). In terms of muscle activation during flexion from standing, detecting paraspinal muscle activity onset can be problematic due to the background firing of these muscles during neutral standing. Signal offset can be classed as complete myoelectrical silence (MS) using a method as described previously, or the point at which deactivation begins (i.e. EMG amplitude peak), described in the literature as critical point (CP) (Kaigle 1998; Sarti et al. 2001; Steventon and Ng 1995).

2.5.6 EMG methodology design considerations

Methodological differences between studies that investigate both spinal kinematics and muscle activity can make it difficult to compare and interpret their findings. Whilst QF protocols have

already been subject to extensive standardisation (Breen A.C. et al. 2012), the wider and less regulated use of sEMG makes this difficult, and so consideration is required in terms of the protocols's sEMG components. The sEMG signal can be influenced by various decisions relating to electrode placement, normalisation techniques, and signal processing. The following section reviews the key sEMG considerations in relation to the development of an appropriate data collection protocol.

2.5.6.1 Surface or fine wire intra-muscular electrodes?

EMG can be recorded using surface electrodes or intra-muscular needles; the former is a non-invasive technique referred to as surface electromyography (sEMG), and is typically used for analysis of more superficially located muscles. The latter involves the placement of fine wire needles directly into the muscle, and therefore suited for investigations of the deeper musculature. There are advantages and disadvantages associated with both electrode types. In terms of reliability, studies that have compared EMG data collected simultaneously with fine wire and surface electrodes have demonstrated greater reliability with the surface electrodes (Soderberg 2000). This finding supports Basmajian and De Luca (1985), who showed that intra-subject variability of the signal was greater when detected with intra-muscular, rather than surface electrodes (Basmajian and De Luca 1985). Poor repeatability (i.e. inter-session reliability) of fine wire EMG is often highlighted as a methodological weakness by manuscript reviewers (Chapman et al. 2010), a problem likely due to difficulties associated with re-inserting wire electrodes into precisely the same position on re-examination. Considering an sEMG signal is more likely to be contaminated by activity from nearby muscles (Solomonow et al. 1994) than fine wire needles, intuitively sEMG recordings may be expected to be less reliable, however this is not apparent.

The use of fine wire electrodes is perhaps most appropriate therefore, when there is a requirement to measure EMG signals from deeper muscle fibres. There is a belief for example that the deep fibres of multifidus have a different functional role to the more superficial multifidus fibres and the erector spinae, with the former more involved in localised stabilisation and the latter as extensors or rotators of the lumbar spine (Richardson and Jull 1995), and considering such different roles the investigation of both superficial and deep multifidus (Tsao et al. 2010) would assist in any exploration of multifidus function. There is evidence to suggest however; that superficial multifidus and the erector spinae can also play an important function is stabilisation (Macdonald et al. 2006), and it has been suggested that the activation of superficial multifidus may be more closely aligned to that of the lumbar longissimus than of the

deep multifidus (Stokes et al. 2003). It could be argued therefore that investigations into multifidus activity would benefit from the use of both sEMG and fine wire needle techniques (Macdonald et al. 2006).

Despite this, there are cases in the literature where multifidus activity is measured using surface electrodes (Hodges 1996; Kim et al. 2015), those that use fine wires to record specifically from the deeper fibres (D'hooge et al. 2013; Macdonald 2009), and those that do both (Tsao et al. 2010). Another consideration with fine wire needling is their association with induced pain (Jonsson et al. 1968; Walker et al. 2001), and as pain is known to influence muscle activity patterns (Geisser et al. 2004) and spinal movement (Thomas and France 2008), this should be avoided when possible during the collection of normative data.

The choice between surface electrodes and fine wire needles therefore, involves consideration of requirements to measure muscle activity from deep or superficial muscles, the availability of technology and the expertise to use them, whether experimental repetitions involve the removal and replacement of the electrodes, and if the avoidance of pain is an absolute necessity. Fine wire electrodes would also require more stringent ethical approval and the use of imaging technology to help accurately guide electrode placement, which comes with additional costs in terms of time, funding and expertise.

2.5.6.2 Electrode positioning

The following discussion refers to sEMG unless stated. To help improve the standardisation of sEMG measurements, guidelines for the positioning of electrodes based on palpation of bony landmarks have been developed (Hermens et al. 1999). Hermens and Vollenbroek-Hutten (2004) discuss the high sensitivity of RMS sEMG to electrode placements, and suggest that whilst normalisation (see section 2.5.6.7) reduces sensitivity, it does not decrease variability to such an extent that RMS sEMG could be judged reliable (Hermens and Vollenbroek-Hutten 2004). A review by Geisser et al (2005) however suggests that EMG amplitude parameters can be used to compare individuals, but their characteristically large intra and inter-subject variability in amplitude behaviour should be considered (Geisser et al. 2005), a characteristic that has traditionally made such comparisons difficult (De Nooij et al. 2009; Lehman and McGill 1999). Indeed, the positioning of sEMG electrodes can contribute significantly to variations in the recorded signal (De Nooij et al. 2009), and therefore electrode application accuracy is an important aspect of study design.

The position of electrodes relative to the muscles' innervation zones (IZ) is also a consideration. The further the electrode from the IZ, the longer the natural delay between onset of actual

myoelectrical activity and the onset of the recorded EMG (Hug 2011). According to Farina and Merletti (2004) the average conduction velocity of a muscle action potential is approximately 4m/s, therefore if the electrode were positioned 4cm away from the IZ there would be a 10ms delay in the detection of EMG onset (Farina and Merletti 2004). When using temporal muscle activation variables the consistency of electrode positioning is therefore particularly important, and to improve the reproducibility of electrode placement Sanchez-Zuriaga et al (2010) utilised an 'anatomic map' documenting the distance of various markings from the 7th cervical vertebrae with the subject stood upright (Sanchez-Zuriaga et al. 2010).

It has been suggested that a general rule of good practice should be to avoid placing electrodes over innervation zones (IZ's), and that inter electrode distances should be small with respect to IZ to tendon distance, however Shiraishi et al. (1995) concluded that it is not even possible to detect the location of the innervation zones for the ES muscles (Shiraishi et al. 1995). There appears therefore to be no current consensus regarding the optimal location of electrodes, and a recent review concluded that further research is still required (Swinnen et al. 2012).

The SENIAM project (Surface ElectroMyoGraphy for the Non-Invasive Assessment of Muscles) was an attempt to standardise EMG methodology within the European Union, and recommended specific electrode positions for numerous lumbar muscles (Hermens et al. 1999). These guidelines are currently the gold standard, and so the electrode positions used in this study, are based primarily on SENIAM recommendations.

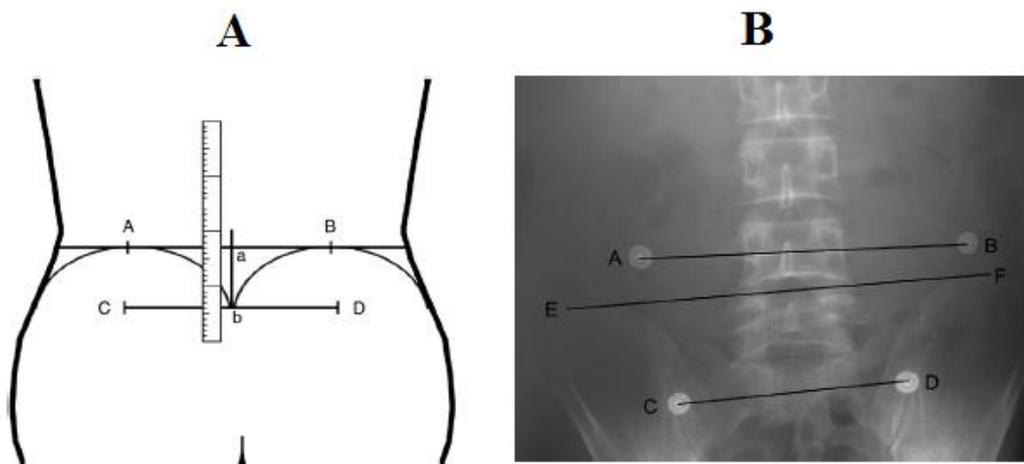
2.5.6.3 Electrode positioning for the paraspinal muscles

The SENIAM guidelines for electrode positioning to record from the superficial multifidus (LMU) state that "electrodes need to be placed on and aligned with a line from caudal tip posterior spina iliaca superior to the interspace between L1 and L2 interspace at the level of L5 spinous process (i.e. about 2 - 3 cm from the midline)" in alignment with the multifidus muscle fibres (Hermens et al. 1999) and in accordance with (Kim et al. 2015). In terms of longissimus, the literature suggests that due to intra-subject variation and minimal research in the field, there is no consensus on the alignment of the longissimus muscle fibres (Bogduk 1980; De Foa et al. 1989; Macintosh et al. 1993), and that fibre direction changes during forward bending (McGill 2000). Despite this, the SENIAM guidelines state that for longissimus "the electrodes need to be placed at 2 fingers width lateral from the spinous process of L1" and should be orientated vertically, parallel to the spine (Hermens et al. 1999).

2.5.6.4 Spinal Level Identification

When the muscles required for investigation have been selected, sEMG electrode application accuracy becomes reliant on subjective identification of bony anatomical landmarks. The use of the iliac crest level (ICL) and the posterior superior iliac spine (PSIS) are popular landmarks for identification of spinal levels, however they have been shown to be unreliable (Kim et al. 2007) Figure 21A and 21B.

Figure 21A and 21B: Manual marking versus radiographic measurement methods (taken from Kim et al. 2007)



Note: In figure 21A the Iliac crests are defined by points A and B, and the PSIS's by points C and D. In Figure 21B the Iliac crests are estimated by manual palpation to be at points A and B, however the x-ray shows the true position to be at points E and F.

Changes in the Iliac crest-lumbar relationship have been shown to change in a proportion of patients between standing and lying in a prone position (Chin et al. 2006), and locating spinal position has shown poor reliability (Billis et al. 2003; McKenzie and Taylor 1997). Chakraverty (2007) showed that different spinal levels were identified dependent on the technique used, for example ICL by palpation or imaging. Palpation most frequently identified the L3 or L3/4 interspinous space, whereas imaging most frequently identified L4 or L4/L5 interspinous space (Chakraverty et al. 2007). The poor reliability of using the Iliac crest to determine L4 has also been highlighted by (McGaugh et al. 2007). All electrode application methods are limited therefore by human subjectivity and variations in individual's anatomy, however it is suggested that accuracy can be improved significantly when palpation and imaging techniques are combined (Merz et al. 2013).

Note: In order to assess the effect of changes in electrode positions on paraspinal sEMG recordings, a preliminary investigation into electrode displacement was conducted (Chapter 3, section 3.3).

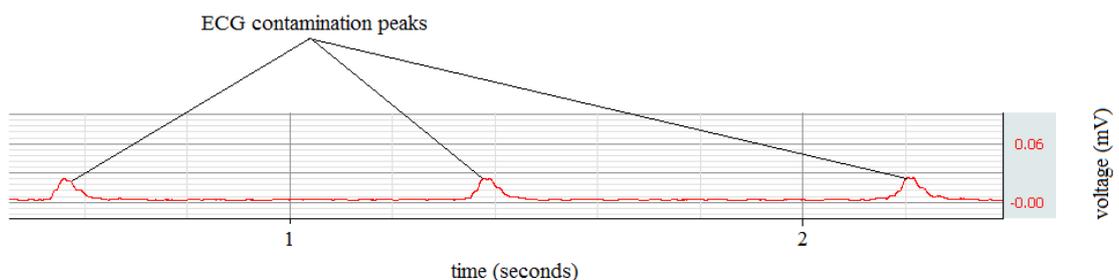
2.5.6.5 Filtering

It is also recommended by SENIAM and the International Society of Electrophysiology and Kinesiology (ISEK), that scientific research studies using sEMG maintain as much of the signal originating from the desired muscle as possible. Unfortunately the signal is readily contaminated by artefacts or noise from the skin-electrode interface, cross-talk from other muscles (including the heart), electronics within the amplifiers and other external sources. Contamination from such sources affects the lower end of the frequency spectra (De Luca et al. 2010), and it is possible to account for them with the use of signal filters. Filtering should be limited as much as possible in order to preserve the desired section of the signal, but a balance is required in order to avoid incorrect interpretations resulting from the contamination. A concern for investigators is the potential for interference from local electrical devices, as amplifiers can pick up ground noise from these sources resulting in an increase of 50Hz baseline noise. The effects of this problem can be minimised by 'earthing' any sources of electrical output, and through the use of modern sEMG equipment that can further reduce noise contamination. It is possible to use a 'notch filter' to remove the 50Hz from the signal (Nelson-Wong et al. 2012), however this technique is avoided whenever it is viable to do so, as it removes too much signal information. The EMG signal typically ranges between 0 and 400Hz, and a large proportion is from the lower frequencies (De Luca et al. 2010), making the use of filters particularly disadvantageous.

2.5.6.6 The problem of ECG

The EMG signal recorded from paraspinal muscles, can be contaminated by the cross-talk from a cardiac muscle contraction (Figure 22). This phenomenon can be particularly problematic if muscle activation onset or offset parameters are the focus of an investigation, as measurement of muscle activation timings may be contaminated due to the overlap of the ECG and EMG signals. Therefore methods for its removal can be applied in these kinds of studies.

Figure 22: An example of ECG contamination of a baseline recording from the erector spinae (in a state of relaxation). The sEMG electrodes were located 5cm lateral to the T9 spinous process



In an investigation into the effect of ECG contamination on the sEMG assessment of back muscles, Hu (2009) used an Independent Component Analysis (ICA) technique to remove the ECG from the signal, and concluded that RMS EMG was reduced after its removal (Hu 2009). They also suggested however that ECG contamination was more prominent in static postures than during dynamic tasks. In a recent study, Coxon (2011) further investigated the effect of removing ECG on RMS EMG, and in agreement with (Hu 2009) concluded that typically RMS EMG was reduced, but that the change was very consistent throughout the signal (Coxon 2011).

The simplest and most widely available method to address ECG contamination is through the use of EMG software filters. It has been suggested that an optimum for ECG removal may be a filter that removes any signal component below 30Hz (D'hooge et al. 2013), however this technique still removes large portions of potentially important lower frequency signal information. The ability to remove ECG peaks is therefore obviously of importance, especially when muscle activity onset or offset are being considered as variables, however their use warrants careful consideration.

2.5.6.7 Normalisation

Normalisation is a process where the raw EMG signal is converted to a scale relative to a known and repeatable value. Due to the inherent variability of the EMG signal, Lehman and McGill (1999) concluded that normalisation is required for interpretation and comparison between bilateral muscles, between the same muscle on different days and between different subjects (Lehman and McGill 1999). The most appropriate and reliable method of normalisation however is an area of disagreement (Norcross et al. 2010). Controlled reference voluntary contractions (RVC's) are useful for clinical populations that cannot achieve a maximal voluntary contraction (MVC) (Hu et al. 2010; Olson et al. 2004), but the two most widely used normalisation techniques utilise either a (MVC) or a sub-maximal voluntary contraction (sMVC).

If signal data is to be interpreted using ratios, there is contention in the literature about whether or not that data should still require normalisation, or if indeed the use of ratios is a normalisation process in itself (Lariviere and Arsenault 2008). The problem with using raw signal data, is that it does not account for variations in participant soft tissue thickness (STT), and so it may be inaccurate to consider that EMG ratios completely circumvent the requirement for EMG normalisation (Lariviere and Arsenault 2008). Lariviere and Arsenault recommend the use of a submaximal normalisation task that loads each of the paraspinal muscles equally as the most appropriate solution, and suggest a modified Sorensen test (In a prone position, the participants

lower body is fixed to a bench, whilst the upper body is unsupported in the horizontal plane (Demoulin et al. 2006)), as a possible method (Lariviere and Arsenault 2008).

2.5.6.8 Reference Contractions

An adapted Sorensen method was used by Claus et al. (2009) in order to recruit paraspinal muscles throughout the length of the lower thoracics to lower lumbar (Claus et al. 2009), and is evidence of how an extension contraction can be used to normalise activity recorded during a flexion based examination. An extension maximal contraction was also used by Peach et al (1998), for EMG normalisation during sagittal bending (Peach et al. 1998). The choice of reference contraction should also be considerate of population under study, as if at any point protocols are intended for use with LBP groups, then a sMVC is perhaps more appropriate than a MVC to minimise the risk of injury, or contamination through fear of the movement (Section 2.5.6.11).

The Sorensen test (Demoulin et al. 2006) requires only a submaximal contraction, perhaps making it more appropriate than maximal contractions for low back pain patients, with contractions found to be no greater than 40-52% of the (MVC) (Mannion and Dolan 1994; Muller et al. 2010). Indeed the use of sMVC's are perhaps more common in a clinical setting for this reason (Dankaerts et al. 2006; Dankaerts et al. 2004). There are varied interpretations of what constitutes an MVC or a sMVC in the literature. Dankaerts et al. (2004) consider the Sorensen test with resistance as a test of MVC, and used the 'prone lying double leg raise' as a sMVC (Dankaerts et al. 2004). This involved having the subject lying prone with knees bent to 90 degrees and both knees lifted 5cm off the ground for 3 seconds. They concluded that a sMVC is more appropriate for the normalisation of trunk muscle EMG when a between days repeated measures study design is employed, which is a consideration for methodologies designed for clinical outcome studies.

There is therefore still much debate over the most appropriate normalisation technique (Soderberg 2000). In terms of reliability, within day reliability was found to be good for both MVC's and sMVC's (Dankaerts et al. 2004). Knutson et al (1994) however, found that measurements were most reliable when normalised to a MVC as opposed to mean or peak dynamic EMG data (Knutson et al. 1994), and Soderberg (2000) suggests that although sMVC, peak and mean dynamic provide reasonable alternatives, they recommend MVC use until further clarification (Soderberg 2000). A review of recent research by Burden (2010) however suggests that the use of isometric sMVC methods provides outputs that have equally good reliability (Burden 2010).

A common sampling interval for reference contractions is 3 seconds with 1 second either side to allow time to achieve peak amplitude (Claus et al. 2009), and although no studies examine the appropriateness of this time period, the general consensus in the literature is to use this time frame (Soderberg 2000). One of the main reasons for a short time sample is to avoid fatigue, which may have its own effect on the signal (Mannion 1997). The number of repetitions for standardisation is also an area of contention, with no scientific basis for any particular number (Soderberg 2000), however Yang (1985) found that the reliability of MVCs and sMVCs increased proportionately with the number of trials (Yang 1985).

2.5.6.9 Normalising to the peak

An alternative method to the sMVC and MVC normalisation techniques discussed, is to use the maximum recorded sEMG obtained during the examination cycle (peak sEMG). Although normalising to a peak value during a dynamic exercise has been demonstrated to decrease the variability between individuals (Chapman et al. 2010), it does not account for the difference between individuals in terms of motor control strategies to produce the same movement. This may result in different activation patterns during the reference contraction of a given muscle between individuals, making comparisons of different individuals and muscles invalid (Naik 2012). This reduction in variability between individuals through normalising to a peak, may also be costly in terms of the loss of valuable biological variation, such as the strength difference between individuals (Knutson et al. 1994). It has also been shown that normalising to peak amplitude during an activity is less reliable between days in the same individual especially in comparison to MVC's (Knutson et al. 1994), and may therefore be of less value if a protocol is intended for use in longitudinal studies.

2.5.6.10 EMG amplitude measurement reliability

Paraspinal muscle amplitude measurements have been shown to have acceptable reliability (Ahern et al. 1986; Daneels et al. 2001), however the number of potential influencing factors associated with EMG recording and analysis warrant a more detailed exploration. In any study combining two technologies, the reliability of different measurements becomes complex and arguably more important due to the possible cumulative effect of poor reliability in multiple parameters. These could include the kinematic measurements as discussed, however EMG elements such as electrode placement, normalisation technique, the choice of electrode and how the electrode position was determined in terms of bony landmarks, can all affect the overall reliability and agreement of EMG measurements.

Note: The reliability and agreement of the mean normalised sEMG amplitude during weight-bearing sagittal flexion is investigated in Chapter 5, section 5.2.

2.5.6.11 Fear of movement

It is possible that sEMG measurements may be influenced by psychosocial factors, such as fear of movement and re-injury (Geisser et al. 2005; Vlaeyen 1999), and whilst less likely in a healthy population group, still required consideration in the study design. It has been suggested by Geisser et al. (2004) that pain related fear is responsible for reduced lumbar flexion, and increased sEMG activity in full flexion (Geisser et al. 2004). Indeed, Karayannis et al. (2013) also demonstrated that a fear of movement is associated with increased trunk stiffness (Karayannis et al. 2013), and Thomas and France (2008) showed that lumbar motion was inversely related to pain, suggesting that pain related fear limits or reduces lumbar spinal movements (Thomas and France 2008). A fear of pain has also been associated with changes in MVC's (Geisser et al. 2004; Lindstroem et al. 2012), and Flexion Relaxation Ratio (FRR) (Geisser et al. 2004; Geisser et al. 2005; Watson 1997), and therefore has the potential to influence sEMG data recorded during flexion based protocols. Even in studies that control the participant's movement range (i.e. not reaching full flexion), pain at any point of the examination needs to be a consideration. Whilst questionnaires such as the Tampa Scale for Kinesiophobia (TSK) have been developed to measure fear of movement (Roelofs 2007; Swinkels-Meewisse 2003; Swinkels-Meewisse 2006), a fear of movement is not anticipated in populations of healthy volunteers.

Section 6

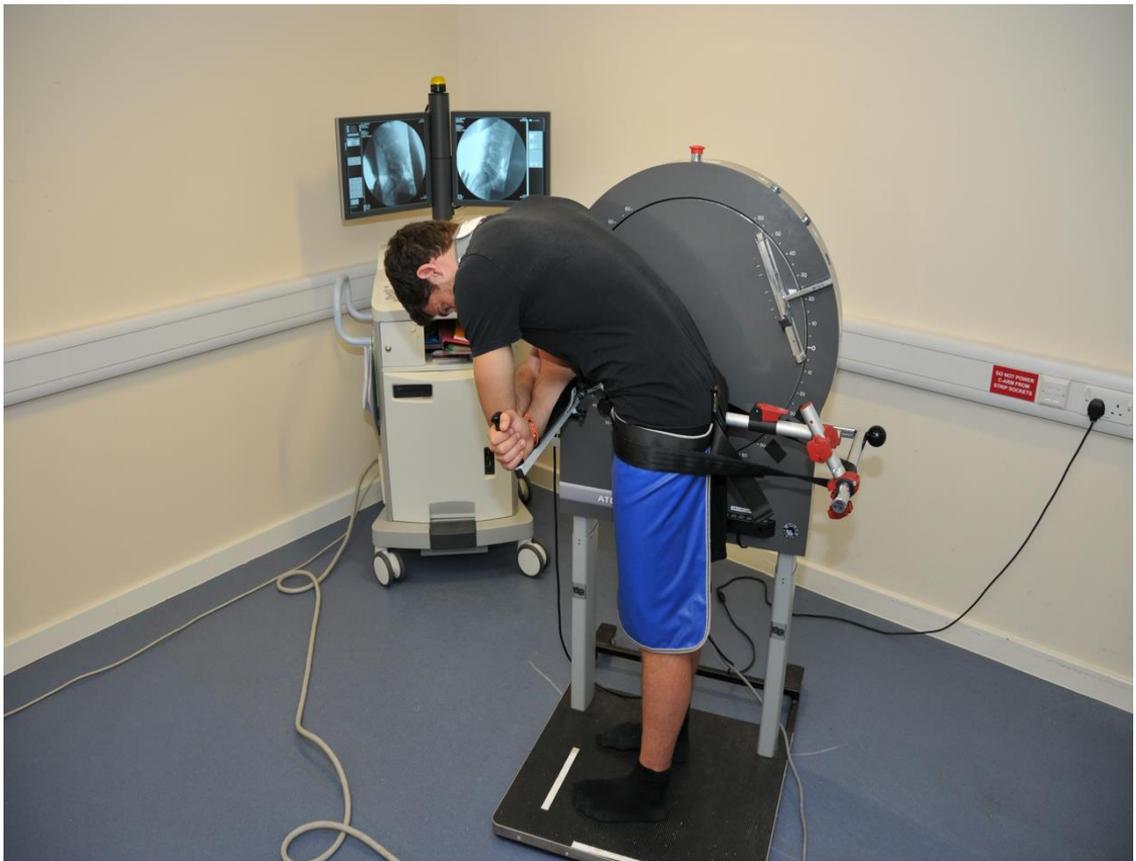
2.6 Dynamic task standardisation

2.6.1 A case for standardisation and the use of pelvic restraint

A key methodological difference between previous investigations that have measured spinal kinematics during a forward bending task, is whether the movement is one of free bending (Ahmadi et al. 2009; Wong 2004) or if the pelvis is somehow stabilised to restrict or prevent its motion (Ahern et al. 1988; Du Rose and Breen 2016a; Du Rose and Breen 2016b; Kingma et al. 2007; Oddsson and De Luca 2003; Peach et al. 1998). There are limitations associated with both techniques. During free bending, participants whilst performing an arguably more natural movement, do so at their own pace and over a comfortable range for the individual. Some studies do attempt to exert a moderate degree of control over these factors using pre-set bending instructions and metronomes for consistent timing (Ross et al. 2015), but adherence to such measures is always likely to be influenced by individual interpretation and motivation. Recently, continuous inter-vertebral measurements have most frequently relied on x-ray or fluoroscopic imaging which are currently limited by a requirement to keep the spinal area of

interest within the image field during movement (Mellor 2014). Such techniques usually use pelvic restraint to do so, and whilst perhaps creating an unnatural bending movement, do achieve isolation of motion specifically through the lumbar segments (Ross et al. 2015), and minimise contributions of pelvic, hip and lower limb musculature (Kingma et al. 2007). Indeed, at a forum that aimed to reach consensus over the most appropriate methods to record and analyse QF information, there was agreement that weight-bearing protocols should stabilise the pelvis (Breen et al. 2012) (Figure 23).

Figure 23: Pelvic stabilisation for a QF sagittal flexion protocol



During sagittal examinations this stabilisation not only restricts anterior pelvic rotation via a belt restraining the ASIS's but also applies pressure over the sacrum (at approximately S2 tubercle). In doing so, it is likely that pressure is applied to the TLF composite (the main connector of the thoracolumbar fascia to the sacrotuberous ligaments and to communication with the posterior thigh muscles) effectively reduces the restraint from muscles such as the biceps femoris and gluteus maximus during flexion, and subsequently allows greater inter-vertebral range. The use of QF also allows standardisation through controlling of rate and range (Breen et al. 2012), but such techniques are still limited by the inability to assess complete ranges of spinal movement

(e.g. full sagittal flexion), and a capacity to only measure narrow regions of the spine (i.e. lumbar region, but not pelvis or thoracics).

The simultaneous rhythm between the lumbar spine and the pelvis is well documented in the literature, with the lumbar spine being shown to contribute more to sagittal trunk rotation during the early stages of flexion, and the pelvis during the latter (Paquet et al. 1994; Tafazzol et al. 2014). The spine, pelvis and lower limbs therefore have fundamental links during movements such as sagittal flexion, relationships which many believe are integral when considering spinal stability (Leinonen et al. 2000; McGregor and Hukins 2009). McGregor and Hukins (2009) for example use the analogy of an inverted pencil balanced on a fingertip. When the pencil begins to fall, relocation of the finger can restore stability by repositioning its centre of gravity. They argue that if the spine is analogous to the pencil, then the pelvis and lower limbs may be considered the fingertip (base), capable of stabilising the spine in the sagittal plane via movement of the hip, knee and ankle joints (McGregor and Hukins 2009). Indeed during sagittal flexion, it has been shown that alongside lumbar extensor muscles there is concurrent activation of other muscle groups including the hamstrings and gluteals (Kim et al. 2013; Leinonen et al. 2000; Nelson-Wong et al. 2012; Sihvonen et al. 1991), and a holistic understanding of control throughout the kinematic chain is obviously desirable. A criticism however of studies that investigate spinal control during movement, is that currently it is simply not possible to analyse every muscle that may be contributing to the control mechanism. If a truly systems level perspective of spinal control is to be taken, then all muscles with the potential to provide control should be included for investigation. The system is arguably too complex to investigate in this way. By stabilising the pelvis, there will be an obvious alteration to normal movement patterns, but doing so will also focus movement above the pelvis and minimise the influence of the associated musculature.

In addition, if the spine and the pelvis are studied concurrently, then it is difficult to determine how each functions independently of the other. For example, natural spine function would be unlikely to be seen in an individual with compromised hip function and vice versa. Therefore if the goal of a study is to investigate the contribution to control of the lumbar spine elements specifically, then every effort should be made to remove as many confounders/influences over this control as possible.

McGregor et al. (2002) investigated spinal and pelvic mobility in groups of elite rowers with and without LBP using MRI (McGregor et al. 2002). They found that the healthy control group had more movement in their lower lumbar spines and relatively less pelvic rotation. In participants

with a current or prior history of LBP however, this trend was reversed with less movement of the spine and increased pelvic rotation. Their findings suggest that stiffening of the lumbar spine in LBP individuals is compensated for by increased movement of the pelvis or the thoracic spine. Due to the limitations of MRI acquisition speeds, the data was taken from static postures however, and it is unclear how dynamic movements may have influenced the findings. There was also no analysis of motion sharing within the lumbar spine, which may have revealed more localised compensation mechanisms.

There are studies that have attempted to quantify inter-segmental contributions to global lumbar ROM, however the patterns demonstrated have been somewhat inconsistent (Aiyangar et al. 2015). There are those that report a relatively increased share of motion apportioned to the cephalic segments (Li 2009; Wong 2006; Wong 2004), and others that describe an increased contribution from the caudal (Boden 1990), whilst other recent studies have suggested that there is actually no significant difference between levels (Aiyangar et al. 2014; Wu et al. 2014). This inconsistency may be due to population variations in relatively small samples, but again may also be attributable to differences in experimental set-ups.

Wu et al (2014) discuss other possible reasons for discrepancies such as active, passive, static or dynamic motions, participants who have or do not have a history of low back pain, and whether the testing was *in vivo* or *in vitro* (Wu et al. 2014). Their study concludes that more standardisation of the experimental set-up is required so that more meaningful comparisons between subject groups and studies can be made, an issue addressed at a forum of researchers using QF to measure spinal biomechanics, and has led to the development of standardised QF protocols (Breen A.C. et al. 2012). This collective agreement regarding QF standardisation was possible due to the relatively small number of QF research groups and a shared willingness to produce comparable data. Perhaps due to its much wider use, common consensus regarding the most appropriate EMG recording methodologies, signal processing and analysis techniques has not been reached.

2.6.2 Rate of movement and gross measurements

A review of the literature evaluating the effects of motion on the biomechanics of the trunk, considered studies that controlled the speed of the movement to be of higher quality than those relying on subjective measures of velocity (Davis and Marras 2000). As discussed previously, the moment of the trunk during a task such as forward bending needs to be offset, and this is achieved primarily through eccentric activation of the vertebral and hip extensor musculature (Sihvonen et al. 1991). The greater the moment the more muscle activation required to

compensate, with the subsequent effect of additional loading on the spine. If an individual chooses to bend forward at a high velocity and stop suddenly at the end of the movement, the reaction of the trunk muscles will be different to that of a slowly decelerating movement strategy (Marras and Mirka 1990). The strategy selection may also relate to pain and dysfunction in the low back itself, as it has been shown that flexion in LBP groups is actually generally performed at a decreased velocity when compared to healthy controls (Marras and Wongsam 1986; Mayer 1984; McGregor et al. 1997).

It seems logical then that velocity will have effect on motor control strategy, and Shirado et al. (1995) speculated that FRP would be affected by the rate of movement (Shirado et al. 1995). In a study investigating the effect of trunk velocity on FRP, Sarti et al. (2001) demonstrated that increasing trunk velocity delayed the onset of FRP to larger angles of trunk flexion (Sarti et al. 2001). The proposed mechanism was a difference in the elastic force produced between groups, which was subsequently reflected in the feedback provided by mechanoreceptors involved. This was in disagreement with Steventon and Ng (1995), who found no difference in FRP onset between groups, and so methodological discrepancies may again be limiting. Indeed, in the case of studies investigating FRP specifically, there has been little consistency regarding the control of movement velocity in the literature (Table 2), making comparisons between studies difficult. There is evidence however, that differences in velocity can possibly affect motor control strategies, limiting the value of comparisons between individuals. Controlling movements in a standardised way can reduce the variation in velocity, providing a way to minimise the confounding effect of this variation on these types of studies. It should be noted however that whilst temporal activation parameters have been shown to alter with velocity of movement, the level of activity (i.e. mean (RMS)) has been shown not to be sensitive (Sihvonen 1988). Such conclusions should be viewed with caution however, due to the limited research in the area, and the intuitive relationship between high velocity movements and the muscle activity required to control them.

Table 2: Methodological differences between well-known FRP studies

Author	Electrode location and inter-electrode distance	Inter-electrode distance	Range and rate of flexion	Gross lumbar angle measurement	Interpretation of FR
Kippers and Parker (1984)	5cm lateral to the spinous process of L3	Not described	Full flexion in approximately 5 seconds	The angle between a line between (PSIS to ASIS) and (PSIS and T1)	Myoelectrical silence (MS)
Gracovetsky et al. (1989)	L5 over multifidus (no other details given)	Not described	No details given	Lumbar flexion angle measured using steel markers and x-ray	Not measured
Sihvonen et al.(1991)	2cm lateral to the spinous processes of L4 and L5 (and intramuscular needles)	Not described	Full flexion in approximately 4 seconds	Finger-floor distance	Not measured
Steventon and Ng (1995)	3cm lateral to spinous process of L3	Not described	Full flexion in approximately 6 seconds	The angle between a line between (L1 and PSIS) and (PSIS and ASIS)	Critical Point (CP) (i.e. Onset of muscle deactivation)
Sarti et al. (2001)	3cm lateral to the spinous process of L3	2cm	Full flexion at 8 seconds (slow) and 3 seconds (fast)	Difference in position of the lumbar spine relative to the pelvis using the T12 spinous process and S1 as reference points	CP and MS
Solomonow et al. (2003)	4-6cm lateral to the L3-4 interspace	2.5cm	Full flexion in approximately 2-3 seconds	Trunk inclination (calculated using lines between Iliac crest and Rib cage) minus hip flexion angle	MS
Dickey et al. (2003)	3cm lateral to the spinous process of T9 and 3cm lateral to the L2-3 interspace	Not described	Full flexion in 4.5 seconds	Calculated using angle between T6 and Sacrum (resulting angle multiplied by 0.77 to estimate angle between L1 and Sacrum)	MS
Olson et al. (2004)	3.5 cm lateral to the L2-3 and L4-5 interspaces	4cm	Full flexion in 5 seconds	As per Solomonow et al. (2003)	MS
Olson et al. (2006)	3cm lateral to the L3-4 interspace	2.5cm	Full flexion in 5 seconds	As per Solomonow et al. (2003)	MS
Descarreaux et al. (2010)	2cm lateral to the L2-3 interspace	2.5cm	Full flexion in 5 seconds	The angle between a line between (L1-L2 interspace and PSIS) and (PSIS and Iliac crest)	MS

Differences in methods are a general problem when trying to interpret study findings, and variations make comparisons between studies difficult. Using examples such as gross lumbar angle measurements and electrode positioning, table 2 highlights the differences that can exist between studies. The lumbar angle measurements are so different in some cases that it would certainly not be possible to compare studies, and although electrode positions are typically placed to record generically from the erector spinae, even slight differences in positioning have been shown to dramatically influence EMG amplitudes (De Nooij et al. 2009), and so even subtle differences require careful interpretation.

Section 7

2.7 Systematic Review

2.7.1 Introduction

As previously described, spinal stability was interpreted by Panjabi (1992) to be dependent on the highly co-ordinated and optimised interactions between three sub-systems, the passive (ligaments, discs, fascia and bones), the active (muscles and tendons) and the neural control systems. If there is dysfunction within a specific system, compensation may be provided by adaptations in the other systems (Panjabi 1992). As an example, Panjabi suggested that abnormally increased muscle activation is a stabilisation mechanism compensating for a loss of spinal stability (Panjabi 1992), a theory repeatedly supported in the subsequent literature (Olson et al. 2004; Shin et al. 2010; Van Dieen et al. 2003). Such adaptations have also been proposed as possible precipitators of LBP, a theory that is difficult to investigate given the inherent heterogeneity of EMG signal data (Lariviere et al. 2000).

In order to improve understanding of the complex interactions between sub-systems, it is necessary to take an approach that incorporates the measurement of both lumbar kinematic and trunk muscle activation data (Dankaerts et al. 2009). A popular method has been to investigate adaptations within the spinal system in response to perturbation (Silfies et al. 2009), and how such responses are influenced by paraspinal muscle fatigue (Granata et al. 2001; Sanchez-Zuriaga et al. 2010; Abboud et al. 2016) and spinal creep deformation (Hendershot et al. 2011; Sanchez-Zuriaga et al. 2010; Abboud et al. 2016), however a recent systematic review suggests that although the literature provides some insight into possible spinal stability mechanisms, the high methodological heterogeneity between studies means that the current evidence is inconclusive (Abboud et al. 2017).

In terms of the investigation of dynamic movement, the study of the FRP (Floyd and Silver 1955) is a possible way in which insight into sub-system interaction can be gained. The deactivation of paraspinal muscle activity during the final stages of forward flexion has been interpreted as the transfer of moment between the active and passive sub-systems (McGill and Kippers 1994), and feasibly provides an insight into sub-system interaction. It has therefore been extensively studied (Luhning et al. 2015; O'Sullivan et al. 2006; Sarti et al. 2001; Descarreaux et al. 2008), however the majority of studies only incorporate the measurement of regional kinematics, and therefore do not provide any insight from the level of the motion segment (Kaigle et al. 1998). It could be argued that investigations at the spinal level are important, as inter-system feedback mechanisms are believed to act at this level (Solomonow et al. 1998).

It has also been common for studies to focus on individual systems in isolation, in an attempt to relate changes within each system to conditions such as LBP. Indeed, in terms of the active system, LBP has been associated with changes in paraspinal muscle cross sectional size (Fortin et al. 2013), activation timings (Williams et al. 2013; Nelson-Wong et al. 2012) and muscle activation amplitudes (Van Dieen et al. 2003; Reeves et al. 2006; Sanchez-Zuriaga et al. 2015; Ahern et al. 1988; Kuriyama et al. 2015). Focus on the passive system has shown potential links between LBP and lumbar ROM (Dankaerts et al. 2009; Taghipour-Darzi et al. 2012; McGregor et al. 1997; Abbott et al. 2006; Kulig et al. 2007; Teyhen et al. 2007; Mellor et al. 2014), and sagittal balance/postural (Mehta et al. 2012) parameters, however such investigations, by considering only one spinal control element, can only speculate as to how such changes may relate to adaptations in the other sub-systems.

In addition, many of these studies have produced conflicting results, and there is therefore an argument that attempts should first be made to improve understanding of normal, so as to better understand what is abnormal (Peach et al. 1998). In their study, Peach et al. whilst considering both kinematics and muscle activity to develop a database of normal movement and activation patterns, did not relate their findings in any detail to mechanisms of spinal stabilisation. Investigations of the kinematics of normal controls has shown how changes in regions of the spine may be associated with changes in another (Mitchell et al. 2008; Hemming et al. 2016), however again, such adaptations again cannot be explained in terms of sub-system adaptation, as only a single system was considered.

The complexity and inaccessibility of investigating spinal control mechanisms makes the interpretation of study findings difficult. A key problem is that sub-system interaction is dynamic, and therefore the study of two or more systems concurrently in living humans requires

instrumentation that can do so dynamically. Physical activities involving sagittal bending are commonplace activities of daily living (Colloca and Hinrichs 2005), and so an improved knowledge of sub-system interaction during lumbar flexion would be of clinical interest. To the author's knowledge, there is currently no review of studies that have investigated dynamic flexion movements using a combination of EMG and lumbar kinematic measurements. As such it is not clear how understanding of Panjabi's spinal stability concepts has advanced with regards to this functional movement of the spine.

This review addressed two fundamental questions. 1) Can the information acquired by combining lumbar kinematic and muscle activity measurements during functional movements (i.e. forward bending) assist in distinguishing between groups of healthy controls and those with low back pain? 2) How have such studies conducted since Panjabi's seminal 1992 paper improved understanding of lumbar spinal stability mechanisms (i.e. sub-system interactions)?

2.7.2 Literature search strategy

Pubmed and Cochrane databases were searched in March and April 2017. The systematic search was performed using combinations of the following keywords: (Electromyography OR EMG or Flexion Relaxation OR FRP AND Kinematics OR Range of Motion OR ROM AND Low Back Pain OR Lumbar Spine AND Flexion OR Bending AND Stability OR Stabilization). Article screening was conducted by the author, and was restricted to English publications.

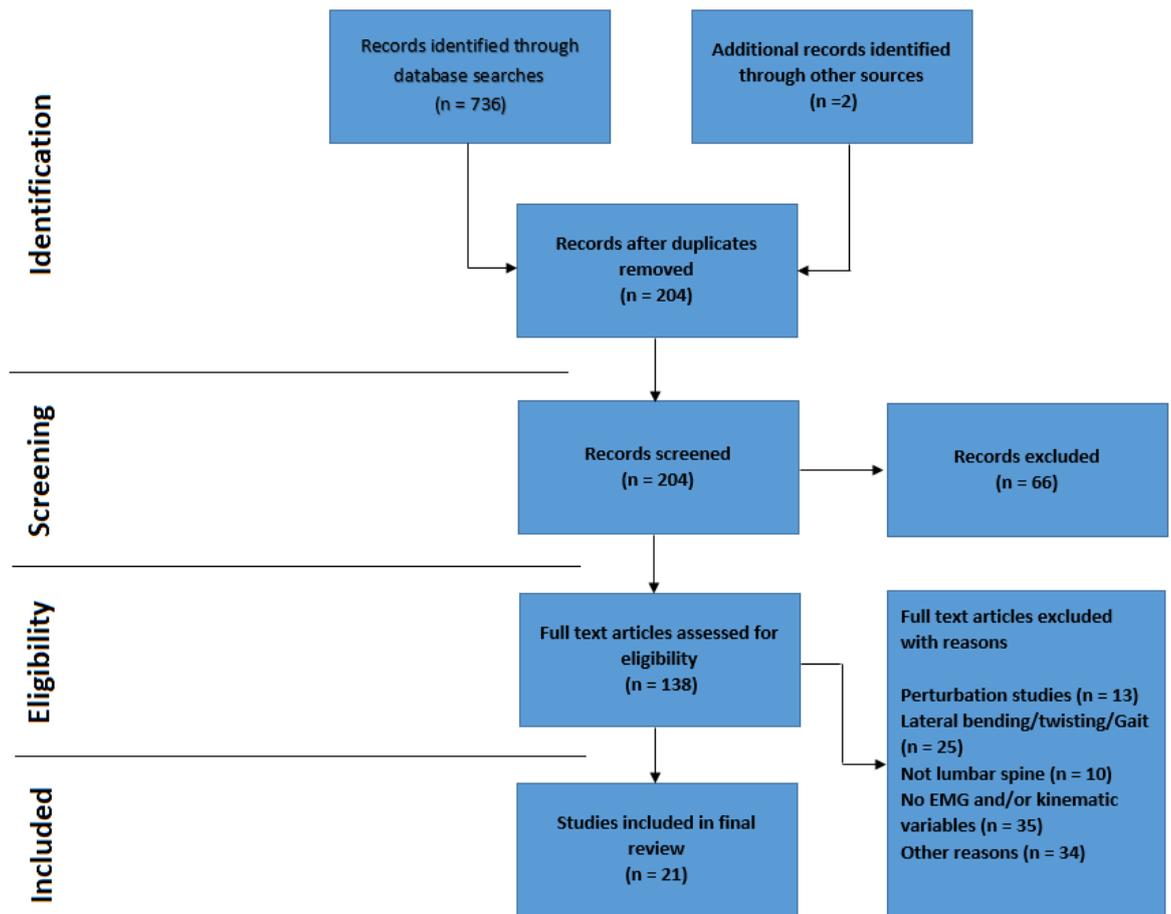
2.7.3 Inclusion and exclusion criteria

Articles were included for review if they met the following inclusion and exclusion criteria. Inclusion criteria consist of 1) studies must be *in vivo* using adult participants 2) dynamic weight-bearing movement in the sagittal plane including forward flexion 3) Include both EMG (including the lumbar paraspinal muscles) and lumbar kinematic measurements 4) Relate study findings to stability theories or spinal stabilisation. Exclusion criteria included 1) Perturbation studies (as the articles of interest were to include active movement 2) Studies measuring creep or fatigue (as single cycles of dynamic tasks will unlikely result in either 3) Studies not investigating the lumbar spine specifically (i.e cervical, thoracic or shoulder) 4) Studies investigating lateral flexion, axial rotation or gait (i.e. not including sagittal flexion) 5) Non-human studies (e.g. feline studies) 6) Repeatability trials. A flowchart outlining the citation selection process is shown in figure 24. Other reasons for study exclusion included manipulation by design (e.g. investigations into the effects of noxious stimuli, high heels, taping, exercise etc.).

2.7.4 PRISMA (Preferred Reporting Items for Systematic reviews and Meta-Analyses)

This systematic review broadly adheres to the PRISMA guidelines (Moher et al. 2009), recommendations that were designed to enhance the value and quality of systematic reviews, and improve the reader’s ability to assess it’s strengths and weaknesses (Appendix S).

Figure 24: Prisma flowchart



2.7.5 Study quality assessment

This review uses a quality assessment tool developed by (Abboud et al. 2017) that was adapted from the Quality Index of Downs and Black (Downs and Black 1998). Abboud et al. 2017 also created an assessment designed to specifically interpret the quality of studies incorporating EMG, which was based on SENIAM (Hermens et al. 1999) and ISEK (Merletti 1999) guidelines. This novel assessment was also incorporated.

2.7.5.1 Overall quality assessment (Abboud et al. 2017)

The original quality index developed by Downs and Black (1998) has been shown to have good test-retest ($r = 0.88$) and inter-rater observability ($r = 0.75$) (Abboud et al. 2017). The adapted tool consists of 10 items that were deemed appropriate for the purpose of this review. The items included the following questions 1) Is the hypothesis/aim/objective of the study clearly described? 2) Are the main outcomes to be measured clearly described in the Introduction or Methods section? 3) Are the characteristics of the patients included in the study clearly described? 4) Are the interventions of interest clearly described? 5) Are the main findings of the study clearly described? 6) Does the study provide estimates of the random variability in the data for the main outcomes? 7) Have actual probability values been reported (e.g. 0.035 rather than <0.05) for the main outcomes except where the probability value is less than 0.001? 8) Were those subjects who were prepared to participate representative of the entire population from which they were recruited? 9) If any of the results of the study were based on “data dredging”, was this made clear? 10) Were the statistical tests used to assess the main outcomes appropriate? All items were scored either 0 or 1. This produced a total quality score out of 10 for each study, with the exception of those articles that did not require population comparison, and so were scored out of 9 (Table 3). Final scores were converted into percentages and combined with the EMG quality scores, providing an overall impression of study quality (Table 5).

2.7.5.2 Specific EMG quality assessment

The checklist developed by Abboud et al. 2017 consists of 12 items divided into 4 sections (Table 4). The first section considers the use of sEMG electrodes and comprises a score for inter-electrode distance, electrode material and construction (i.e. bipolar). The second section considers participant skin preparation, the use of reference electrodes and electrode placement and fixation. The third section considers signal processing and includes items regarding the use of filters, rectification methodology, sampling and processing. The final section considers the appropriate use of normalisation. Each item was scored 0 or 1, and a score of 1 was attributed to a section if the item totals reached 2 or more. This produced an EMG quality score out of 4 for each study, with the exception of those articles where normalisation was not deemed necessary, and so were scored out of 3. These scores were also converted into percentages and combined with the study quality assessment scores above (Table 5).

2.7.6 Results

Out of a total of 736 articles identified through the literature search only 21 satisfied the inclusion/exclusion criteria. The screening process is outlined in the PRISMA flowchart (Figure 24).

2.7.6.1 Overall and EMG quality assessment

The overall quality assessment scores ranged from 44-100% with a mean total score of 80% (Table 3). All of the selected studies scored a 1 for their descriptions of methodology and study findings. The studies also performed well in terms of the quality of hypothesis and outcome descriptions (19/21 and 20/21 respectively), and their use of appropriate statistics and absence of data dredging (both 20/21). Areas in which the studies generally scored poorly included the description of participant characteristics (9/21) and the reporting of actual probability values (7/21). The EMG quality assessment showed scores ranging from 25-100% with a mean total score of 73% (Table 4). The assessment showed that the majority of EMG studies adequately reported the normalisation and signal processing elements, however it also highlighted a mixture of study quality when considering the detail of electrode use. The combined overall and EMG quality index scores ranged from 47-100% with a mean total score of 77% (Table 5).

Table 3: Quality index assessment scores (*Studies that did not compare healthy controls to a low back pain group were rated using a 9 point scale instead of 10)

Quality check											Score (/9* or /10)	Score (%)
Authors (year)	1	2	3	4	5	6	7	8	9	10		
Arjmand et al. 2010	0	1	0	1	1	0	0	N/A	1	0	4*	44
Burnett et al. 2004	1	1	0	1	1	1	1	1	1	1	9	90
Callaghan and Dunk 2002	1	0	0	1	1	0	1	N/A	1	1	6*	67
Cholewicki et al. 1997	1	1	0	1	1	1	0	N/A	1	1	7*	78
Dankaerts et al. 2009	1	1	1	1	1	1	1	1	1	1	10	100
Hashemirad et al. 2009	1	1	0	1	1	1	0	N/A	1	1	7*	78
Hay et al. 2016	0	1	0	1	1	0	0	1	1	1	6	60
Kaigle et al. 1998	1	1	1	1	1	1	0	0	1	1	8	80
kienbacher et al. 2016	1	1	1	1	1	1	1	1	1	1	10	100
Lariviere et al. 2000	1	1	1	1	1	1	1	0	1	1	9	90
Liu et al. 2011	1	1	0	1	1	1	0	0	1	1	7	70
Luhring et al. 2015	1	1	1	1	1	1	0	N/A	1	1	8*	89
Mayer et al. 2009	1	1	0	1	1	1	0	1	0	1	7	70
McGill and Kippers 1994	1	1	0	1	1	1	0	N/A	1	1	7*	78
Nairn et al. 2013	1	1	0	1	1	0	1	N/A	1	1	7*	78
Neblett et al. 2003	1	1	1	1	1	1	0	N/A	1	1	8*	89
Ning et al. 2012	1	1	0	1	1	0	0	N/A	1	1	6*	67
O'Sullivan et al. 2006	1	1	1	1	1	1	1	N/A	1	1	9*	100
Paquet et al. 1994	1	1	1	1	1	1	0	1	1	1	9	90
Peach et al. 1998	1	1	0	1	1	1	0	N/A	1	1	7*	78
Sanchez-Zuriaga et al. 2015	1	1	1	1	1	1	0	1	1	1	9	90

Table 4: EMG quality assessment scores (*Studies that did not require normalisation were rated using a 3 point scale instead of 4) as per Abboud et al. 2017

EMG quality check													score (/3* or /4)	score (%)
Authors (year)	1.1	1.2	1.3	2.1	2.2	2.3	2.4	3.1	3.2	3.3	3.4	4		
Arjmand et al. 2010	0	1	0	0	1	0	0	1	1	1	0	1	2	50
Burnett et al. 2004	1	1	0	1	1	0	0	1	1	1	0	1	4	100
Callaghan and Dunk 2002	0	1	0	0	1	0	0	1	1	1	0	1	2	50
Cholewicki et al. 1997	0	1	0	0	1	0	0	1	1	1	1	1	2	50
Dankaerts et al. 2009	1	1	1	0	1	1	1	1	1	1	1	1	4	100
Hashemera d et al. 2009	1	1	1	0	1	1	1	1	1	1	1	1	4	100
Hay et al. 2016	0	1	1	0	0	0	0	1	1	1	1	N/A	2*	67
Kaigle et al. 1998	0	0	0	0	1	1	0	1	1	1	1	N/A	2*	67
kienbacher et al. 2016	1	1	1	0	1	1	1	1	1	1	1	1	4	100
Lariviere et al. 2000	0	1	0	1	0	1	0	1	1	1	0	N/A	2*	67
Liu et al. 2011	0	0	N/A	1	0	1	0	1	1	1	1	1	3*	100
Luhring et al. 2015	1	1	1	0	1	0	0	1	1	1	1	1	3	75
Mayer et al. 2009	0	0	0	0	0	0	0	1	1	1	0	0	1	25
McGill and Kippers 1994	1	1	1	0	1	0	0	1	1	1	1	1	3	75
Nairn et al. 2013	1	0	1	1	1	0	1	1	1	1	1	1	4	100
Neblett et al. 2003	1	1	0	1	1	0	0	0	0	0	1	N/A	2*	67
Ning et al. 2012	0	0	1	0	1	0	0	1	1	1	1	1	2	50
O'Sullivan et al. 2006	1	1	1	1	1	1	1	1	1	1	1	1	4	100
Paquet et al. 1994	0	0	0	1	0	0	0	0	1	1	1	0	1	25
Peach et al. 1998	1	1	1	0	1	0	0	1	1	1	1	1	3	75
Sanchez-Zuriaga et al. 2015	1	1	1	0	1	1	1	1	1	1	1	1	4	100

Table 5: Combined quality index and EMG quality scores

Authors (year)	Quality index score (%)	EMG quality score (%)	Combined score (%)
Arjmand et al. 2010	44	50	47
Burnett et al. 2004	90	100	95
Callaghan and Dunk 2002	67	50	58.5
Cholewicki et al. 1997	78	50	64
Dankaerts et al. 2009	100	100	100
Hashemerad et al. 2009	78	100	89
Hay et al. 2016	60	67	63.5
Kaigle et al. 1998	80	67	73.5
kienbacher et al. 2016	100	100	100
Lariviere et al. 2000	90	67	78.5
Liu et al. 2011	70	100	85
Luhring et al. 2015	89	75	82
Mayer et al. 2009	70	25	47.5
McGill and Kippers 1994	78	75	76.5
Nairn et al. 2013	78	100	89
Neblett et al. 2003	89	67	78
Ning et al. 2012	67	50	58.5
O'Sullivan et al. 2006	100	100	100
Paquet et al. 1994	90	25	57.5
Peach et al. 1998	78	75	76.5
Sanchez-Zuriaga et al. 2015	90	100	95

2.7.6.2 Study characteristics

Table 6: Study characteristics (N = 21)

Authors	Study aim	EMG variable and lumbar paraspinal muscles recorded (sEMG unless otherwise stated *)	Lumbar kinematic measurements	Study findings	Participants	Analysis
Arjmand et al. 2010	To compare a single joint model to kinematic driven model during trunk flexion.	Normalised EMG activity. Muscles Longissimus (3cm lateral to L1) Iliocostalis (3cm lateral to L3) Multifidus (2cm lateral to L5).	Optotrak 4 camera system (regional) Lumbar region LED's placed on pelvis and T12.	In both models, global extensor activity peaked around 30° of flexion, due to the increase in contribution of passive structures at this point. Extensors became silent between 50-70°.	N = 1 A male participant with no recent history of LBP.	Quantitative comparison was not performed.
Burnett et al. 2004	To determine whether differences exist in spinal kinematics and trunk muscle activity in cyclists with and without NSCLBP.	EMG activity was quantified by obtaining the mean activation, during a 5 crank revolution period. Muscles TES (5cm lateral to T9) LMU (2-3cm lateral to L4-L5).	3-Space Fastrak (regional) Lower lumbar L3 relative to S2 Upper lumbar T12 relative to L3.	The LBP group demonstrated greater lower lumbar flexion than controls associated with a loss of multifidus co-contraction.	N = 18 mean age 37.6 years 9 non low back pain 9 NSCLBP.	Independent sample t-tests.
Callaghan and Dunk 2002	To determine if FRP occurs in seated and slumped postures.	Ensemble average normalised EMG activity. Muscles TES (5cm lateral to T9) LES (3cm lateral to L3).	3-Space ISOTRAK (regional) Lumbar region Sacrum relative to L1.	FRP was shown in the TES, but not the LES during Slumped sitting. TES silence during sitting also happened at earlier angle of lumbar flexion than during standing.	N = 22 low back pain free participants 11 males mean age 21.3 years 11 females mean age 21.9 years.	Three way ANOVA, and Tukey's post hoc multiple comparisons.

Cholewicki et al. 1997	To test the hypothesis that the flexors and extensors of the trunk are co-activated around a neutral spine posture.	Normalised EMG activity. Muscles TES (5cm lateral to T9) LES (3cm lateral to L3) LMU (2 cm lateral to L5-L5).	The use of 2 pieces of string attached to a chest harness and two potentiometers (regional).	Co-activation of trunk flexors and extensors was shown in healthy participants around a neutral posture.	N = 10 low back pain free participants 8 males and 2 females mean age 27 years.	A two factor repeated measures ANOVA.
Dankaerts et al. 2009	To test the ability of a model to distinguish between FP and AEP subgroups and healthy controls using lumbar kinematics and trunk muscle activity.	Normalised EMG activity. Superficial LMU (at the level of L5 orientated by a line between the PSIS and the L1-L2 interspace). Iliocostalis lumborum pars thoracis (lateral to L1).	3-Space Fastrak (regional) Upper lumbar T12 relative to L3 Lower lumbar L3 relative to S2.	Differences in muscle activity and spinal kinematics during flexion suggest that 2 distinct motor control patterns can exist in CNSLBP patients.	N = 67 participants 34 low back pain free controls, mean age 32 20 Flexion pattern NSLBP patients, mean age 36 13 Extension pattern NSLBP patients, mean age 40.	ANOVA and <i>post hoc</i> Bonferroni.
Hashemirad et al. 2009	To investigate the relationship between lumbar spine flexibility and LES activity during sagittal flexion and return.	Normalised EMG amplitude and signal onset/offset. Muscle LES (4cm lateral to L3-L4).	Estimated using a camera and markers placed at the spinous processes of T12, L3 and S2 (regional).	During bending the ES of participants with high toe touch score deactivated at greater trunk and hip angles. Those with high modified Schober scores deactivated later and reactivated sooner in accordance with lumbar angle.	N = 30 low back pain free participants.	Pearson correlations and multiple linear regression analysis.
Hay et al. 2016	To show that wavelet coherence and phase plots can be used to provide insight into how muscle activation relates to kinematics.	EMG amplitude (linear envelope). Muscle Lumbar erector spinae (no details of positioning).	Oqus 400 motion capture system (regional) Reflective markers placed over T12 and S1.	The study showed good agreement between lumbar kinematics and linear enveloped sEMG. Validating the use of the wavelet coherence technique.	N = 14 low back pain free male participants.	The coefficient of determination (R^2).

Kaigle et al. 1998	To concurrently quantify muscle activation of LES with the kinematics of lumbar motion segments, in low back patients and controls.	Root mean square (RMS) sEMG amplitude. Muscle LES (3cm lateral to L3-L4).	A linkage transducer system secured by interosseous pins to L2-L3, L3-4 and L4-L5 motion segments (inter-vertebral).	ROM was less in low back pain patients and FRP occurred in participants when IV-ROM was complete before full trunk flexion	N = 13 6 low back pain free participants, mean age 40. 7 low back pain patients with suspected lumbar instability, mean age 51.	Wilcoxon rank-sum test and Wilcoxon matched-pairs signed rank test.
Kienbacher et al. 2015	To determine whether lumbar extensor activity and flexion relaxation ratios could differentiate low back pain patients (of various age groups) during flexion-extension task.	Normalised RMS sEMG amplitudes. Muscle LMU (lateral to L5) a line joining the iliac crests, and 2–3 cm bilateral and distal from their middle).	3-D accelerometers placed at the levels of T4 and L5. Used to calculate hip, lumbothoracic and gross trunk regions. (regional).	The sEMG activation was highest in over 60's and female groups during standing. This possibly relates to why this group showed minimal changes during flexion. This group also demonstrated the highest hip, and lowest lumbothoracic angle changes.	N = 216 low back pain patients. 62 (60-90 year olds) 84 (40-59 year olds) 70 (18-39 year olds).	ANOVA and bootstrap confidence intervals.
Lariviere et al. 2000	To evaluate the sensitivity of trunk muscle EMG waveforms to trunk ROM and low back pain status during flexion-extension tasks.	Mean normalised EMG activity. Muscles LES and TES (exact locations not specified).	Video cameras and reflective markers. Trunk angles relative to the vertical plane were used to determine trunk flexion (A line between the hips and the centre of C7-T1) (regional).	Principal component analysis (PCA) distance measures were sensitive to trunk ROM but not low back status. The usefulness of PCA as an effective clinical tool was not established.	N = 33 15 low back pain patients, mean age 40 18 low back pain free participants, mean age 39.	ANOVA and ICC's.
Liu et al. 2011	To develop a new test based on lumbar sEMG activity (the sEMG coordination network analysis approach) during flexion-extension, to distinguish	Normalised RMS sEMG activity. Muscles An sEMG electrode array placed over the lumbar region (16 electrodes, target muscles not specified).	30° of trunk flexion, measured by a protractor (no further details) (regional).	Group network analysis shows a loss of global symmetric patterns in the low back pain group.	N = 21 11 low back pain patients, mean age 40. 10 low back pain free participants, mean age 28.	Did not specify. (However, groups comparison statistics and symmetry scores were used).

between healthy control and low back pain groups.

Luhring et al. 2015	To determine a kinematic measurement that best determines the onset and offset of the FRP.	Normalised sEMG onset and cessation. Muscle LES (4cm lateral to L3).	Vicon MX motion capture camera system. Reflective markers placed at various locations throughout the spine including T12, L5 and pelvis (regional).	Lumbar kinematic measurements are preferential when the FRP is considered clinically.	N = 20 low back pain free participants, mean age 24.	Coefficients of Variation (CV) and ICC's.
Mayer et al. 2009	To determine when FRP occurs in patients and to correlate the findings with lumbar ROM.	Mean RMS sEMG with pre-determined cut-off values. Muscles Not identified within paper.	Gross lumbar, hip/pelvic ROM using an inclinometer (no further details provided) (regional).	After a functional restoration program, both normal FRP and normal lumbar ROM were restored in the majority of patients.	N = 134 30 low back pain free participants, mean age 38. 104 low back pain patients (mean age not provided).	Descriptive statistics including mean and SD. Sensitivity and specificity. P-values and Odds ratios (not specified).
McGill and Kippers 1994	To examine the tissue loading during the period of transition between active and passive tissues during flexion.	Normalised sEMG activity. Muscles TES (5cm lateral to T9) LES (3cm lateral to L3).	3-Space Isotrak (regional) with sensors placed over the sacrum and T10.	The deactivation of lumbar extensor muscles during FRP occurs only in an electrical sense as they still provide force elastically.	N = 8 low back pain free participants, mean age 26.	Dynamic modelling.
Nairn et al. 2013	To quantify slumped sitting both in terms of spinal kinematics and sEMG.	Mean normalised sEMG activity. Muscles Lower TES (5cm lateral to T9) LES (4cm lateral to L3) LMU (Adjacent to L5 orientated along a line between the PSIS and the L1-L2 interspinous space).	Vicon motion capture camera system. Reflective markers placed at various locations throughout the spine including T12, L1 and bilateral PSIS's (regional).	During slumped sitting lower sEMG activity was found in the thoracic and lumbar erector spinae compared to upright sitting. Patterns varied depending on the degree of bending at each area of the spine. Thoracic kinematic and EMG information is therefore useful in	N = 12 low back pain free participants, mean age 23.	ANOVA and Bonferroni correction.

these type of studies

Neblett et al. 2003	To assess EMG activity in terms of the FRP during dynamic flexion and to determine whether abnormal FRP patterns in NSLBP patients can be normalised.	RMS sEMG cut-off values. Muscles LES (2cm lateral to L3).	Inclinometers at T12 and the sacrum (regional).	In asymptomatic participants, the flexion relaxation (FR) angle was always less than the maximal voluntary flexion (MVF) angle. Of the patients that completed a functional restoration program, 94% achieved FR compared to 30% pre-treatment.	N = 66 12 low back pain free participants, mean age 34. 54 chronically disabled work-related spinal disorder (CDWRSD) patients	Descriptive statistics for ROM and FRP
Ning et al. 2012	To determine a boundary at which the passive tissues begin to take a significant role in trunk extensor moment (and therefore at what point EMG assisted modelling is no longer valid).	Normalised EMG activity. Muscles LES at two levels (3cm lateral to L3 and 4cm lateral to L4).	A magnetic-field based motion tracking system with sensors placed at T12 and S1. Lumbar flexion calculated as the pitch of T12 relative to S1 (regional).	EMG-assisted models should consider the action of the passive tissues at lower flexion angles than previously thought.	N = 11 low back pain free participants, mean age 26.	ANOVA and Tukey–Kramer post-hoc testing
O’Sullivan et al. 2006	To investigate the FRP of spinal muscles in healthy participants during slumped sitting from an upright position.	Normalised EMG activity offset. Muscles TES (5cm lateral to T9) LMU (Adjacent to L5 orientated along a line between the PSIS and the L1-L2 interspinous space).	3- Space Fastrak with sensors placed over T6, T12 and S2. (regional).	LMU is active during neutral sitting and demonstrates FRP when moving from upright to slumped sitting. FRP of these muscles is also different to when standing. More variation was found in EMG patterns of the TES.	N = 24 low back pain free participants, mean age 32.	ANOVA and ICC’s

Paquet et al. 1994	To compare healthy controls and low back pain patients in terms of hip-spine movement interaction and EMG, and to verify the relationships between kinematics and EMG in these groups.	Raw EMG envelope. Area under the curve and ratio of activity at different parts of the flexion-extension cycle (not-specified). Muscles LES (at the level of L3, distance not-specified).	Electrogoniometers measured angular displacements at the hip and lumbar spine using landmarks of T8 and S1 (regional).	LES activation patterns were found to be significantly different between groups when flexion was performed at the same rate and range. Abnormal hip-spine movement related to an absence of the FRP at full flexion.	N = 20 10 low back pain free participants, mean age 34. 10 low back pain patients, mean age 38.	Mann-Whitney U test and Kruskal-Wallis test
Peach et al. 1998	To document the lumbar kinematics and trunk EMG activation patterns of healthy controls during tasks including sagittal flexion	Mean normalised EMG. Muscles TES (5cm lateral to T9) LES (3cm lateral to L3) LMU (1-2 cm lateral to L5).	3-Space Isotrak with sensors placed over T12 and Sacrum. (regional).	A database of normal lumbar spinal kinematics and EMG patterns was created for future reference against LBP groups.	N = 24 low back pain free participants, mean age 22.	Descriptive statistics, ANOVA and Tukey's honestly significant difference (HSD) post-hoc testing
Sanchez-Zuriaga et al. 2015	To compare healthy controls and LBP patients in terms of lumbopelvic kinematics and erector spinae activity	Mean normalised EMG activity, and start and end of FRP. Muscle LES (3cm lateral to L3).	A 3-dimensional videophotogram metric system, with markers placed at T12, L3, L5 and the sacrum (regional).	During pain free periods, recurrent LBP patients showed significantly greater LES activity during flexion and extension. Lumbar ROM and FRP were not found to be useful to distinguish between groups.	N = 30 15 low back pain free participants, mean age 41. 15 patients with recurring low back pain (currently in a pain free stage), mean age 45.	Mann-Whitney U test

2.7.6.3 General characteristics of the reviewed studies

All of the studies reviewed could be placed into one of 4 categories, the majority being studies relating in some way to the flexion relaxation phenomenon (FRP): Callaghan and Dunk 2002, Hashemirad et al. 2009, Kaigle et al. 1998, Kienbacher et al. 2015, Luhning et al. 2015, Meyer et al. 2009, McGill and Kippers 1994, Neblett et al. 2003, Ning et al. 2012, O'Sullivan et al 2006, Paquet et al. 1998 and Sanchez-Zuriaga et al. 2015, or comparisons between LBP and healthy control participant groups: Burnett et al. 2004, Dankaerts et al. 2009, Kaigle et al. 1998, Lariviere

et al 2000, Lie et al. 2011, Mayer et al. 2009, Neblett et al. 2003, O'Sullivan et al. 2006, Paquet et al. 1994, Sanchez-Zuriaga et al. 2015. There was a degree of crossover however as some comparison studies also incorporated the FRP. Other study areas included EMG activation studies (other than FRP): Burnett et al. 2004, Cholewicki et al. 1997, Dankaerts et al. 2009, Hay et al. 2016, Lariviere et al. 2000, Liu et al. 2011, Nairn et al. 2013, Paquet et al. 1994, Peach et al. 1998, Sanchez-Zuriaga et al. 2015, and spinal modelling: Arjmand et al. 2010, Lariviere et al. 2000, McGill and Kippers 1994, Ning et al. 2012.

Table 6 shows that typically regional kinematics were measured, with the exception of the inter-vertebral methodology used by Kaigle et al. (1998). Indeed the methods used to measure regional ROM varied a lot between studies. This trend was also apparent in terms of electrode positioning, with many different sites being used to record activity from the same designated muscle. The table also highlights the generally small sample sizes used in this type of study, with the majority using fewer than 30 participants. The only exceptions were the studies of Mayer et al. (2009), Kienbacher et al. (2015), Lariviere et al. (2000) and Neblett et al. (2003) with participant numbers of 134, 216, 33 and 66 respectively.

2.7.6.4 Comparing healthy control and low back pain groups

Of the studies above comparing LBP and healthy control groups, the majority found objective differences between the groups. Burnett et al. 2004: showed that the LBP group had greater lower lumbar flexion and reduced multifidus activity compared to controls, whilst controls showed greater upper lumbar flexion. In Dankaerts et al's study 2009, differences were found in terms of multifidus activity and spinal kinematics between both flexion pattern (FP) and active extension pattern (AEP) provocation sub-groups and healthy controls. In summary, multifidus activity was increased in the AEP group relative to the FP at the end of flexion, and the FP group demonstrated increased activity compared to the healthy controls. These patterns were attributed to the maintenance of the lumbar lordosis during flexion in the AEP group, and the similar spinal curvature between FP and healthy control groups. The Kaigle study provided the only inter-vertebral insight into active and passive system interactions, using intra-osseous pins connected to a sliding linkage transducer system to measure inter-vertebral angular rotation (Kaigle et al. 1998). The study showed that inter-vertebral angular range was significantly smaller in the LBP group, and that the majority of patients showed no reduction in paraspinal muscle activity at the end ranges of flexion. Indeed, the FRP was only present in participants who demonstrated near complete inter-vertebral rotation before maximum global trunk flexion was attained.

Two of the studies were linked and provided similar conclusions. Neblett et al. 2003 showed that in terms of the FRP and patients, all LBP patients that underwent a rehabilitation program achieved normal ROM, and subsequently demonstrated the FRP, whilst Mayer et al. 2009 likewise concluded that normal lumbar ROM appears to correlate with the FRP, and was therefore absent in many LBP participants. However, both FRP and ROM measurements responded well after a generic rehabilitation program.

Using a network modelling and analysis approach Liu et al. 2011 claimed to be able to clearly distinguish LBP and healthy control participants using symmetric patterns and network features, and Paquet et al. 1994 showed that when flexion was performed over the same rate and range, LES activity was significantly greater in the LBP group. Participants in the study with an absent FRP also demonstrated increased ROM of the hip around full flexion.

Not all studies demonstrated an ability to differentiate between LBP and control groups however. Lariviere et al. (2000) for example used a novel principal component analysis (PCA) technique to investigate whether EMG and kinematics could distinguish between the two. Their PCA analysis consisted of two steps. Firstly using EMG activity envelopes from control subjects, a reference model was developed (i.e. a criteria for normal). Secondly 'distance measures' were calculated relative to the reference model. The EMG waveform of a participant was labelled as abnormal if the 'distance value' was outside a 95% confidence interval calculated from the control subjects. Whilst being sensitive to trunk ROM, the distance measures were not sensitive to low back pain status. The authors argued that this was likely due to the relatively small sample size, and therefore inadequate considering the large heterogeneity control populations. In conclusion it was considered that the tool developed was not useful in terms of distinguishing between LBP patients and controls. Sanchez-Zuriaga et al. 2015 also demonstrated contrasting results, as the authors found no significant difference between LBP and healthy groups, in either FRP or lumbar ROM. The study did however show significantly greater LES activity in LBP participants during the flexion-extension task, and the LBP patients were participating during a pain free period.

2.7.6.5 Flexion relaxation studies

The results of some of the FRP studies have already been mentioned (Kaigle et al. 1998; Paquet et al. 1994; Mayer et al. 2009; Neblett et al. 2003; McGill and Kippers 1994). Callaghan and Dunk (2002) showed that during slumped sitting the TES exhibited the FRP, but the LES did not. The authors also demonstrated that this deactivation occurred earlier (i.e. at a smaller lumbar flexion angle) than LES deactivation during flexion from standing (Callaghan and Dunk 2002). In contrast

to these findings, O'Sullivan et al. showed that although LMU activity decreased (i.e. FRP was present) when going from a neutral to a slumped seated position, there were varying patterns in TES activity, as approximately half the participants showed an increase in activity and half a decrease (O'Sullivan et al. 2006). Hashemirad et al. showed that trunk flexibility can influence FRP, with greater flexibility relating to FRP onset at larger flexion angles (Hashemirad et al. 2009), and Luhning et al. (2015) chose to address the problem of using different methodologies to measure regional kinematics in FRP studies (by acknowledging a wide range of normalised and un-normalised FRP onset angles), investigated whether lumbar (i.e. T12-L5) or trunk (i.e. shoulders and hips) angles were more consistent in terms of EMG cessation and onset. The study found that lumbar kinematic measurements were more consistent.

Finally, the study conducted by Ning et al. (2012) suggested that passive tissues can produce significant loads at earlier trunk flexion angle than previously believed i.e. those suggested by kaigle et al. (1998) where erector spinae deactivation was shown to begin at between 71° and 77° of grouped inter-vertebral level flexion, or Peach et al. (1998) where FRP was shown to occur between 60° and 70°.

2.7.6.6 Models

Arjmand et al. (2010) compared EMG-driven (EMGAO) and multi-joint Kinematics-driven (KD) models in terms of muscle force and spinal load estimation. During a flexion task the KD model predicted greater paraspinal muscle activity compared to the EMGAO model and therefore shear and compression forces were also higher. Predictions made using the EMGAO model were also found to be level specific (i.e. L5-S1), and could not be an accurate representation of other lumbar levels (Arjmand et al. 2010). Ning et al. 2012 as discussed above, determined at what trunk flexion angle the passive tissues were able to generate a significant extensor moment during forward bending (Ning et al. 2012), and McGill and Kippers 1994 showed that although paraspinal muscles are electrically silent at the end range of forward flexion, these muscle continue to provide elastic resistance via passive stretching.

2.7.7 Discussion

2.7.7.1 Quality assessment

The mean of the combined quality check and EMG scores was 77%, suggesting that the overall quality of the studies reviewed was generally good. Of particular note were the studies of Dankaerts et al. 2009, Kienbacher et al. 2016 and O'Sullivan et al. 2006, which all scored 100%. The majority of studies used muscle activity amplitude as their key EMG parameter, and it was

apparent that the majority also reported the relevant normalisation technique. The high percentage of good scores in this area, therefore makes it easier to compare amplitude results between studies. Other areas of apparent good quality reporting included the descriptions of the hypothesis, aims, and objectives of the studies, the main outcomes to be measured, the interventions of interest and the main findings. In terms of EMG quality, relevant signal processing information was also usually well reported.

This high standard of reporting was not evident throughout the review however, and trends in areas that were weaker emerged. In terms of the Quality Index assessment scores, the reporting of participant characteristics (including inclusion and exclusion criteria) and actual probability values was poor, with over half of all studies included scoring zero for these categories. Regarding the EMG quality assessment scores there was notably poor reporting of skin preparation techniques, the placement and fixation of electrodes and details regarding the use of reference electrodes, information that would be important if these studies were to be replicated. Sample sizes were also generally small, with 17/21 studies using samples of <30 participants. This potentially weakens the statistical power of these studies, and increases the chance of Type II errors.

2.7.7.2 Spinal stability and sub-system interaction

None of the studies included in this review had the specific objective to investigate sub-system interaction (Table 6). The findings therefore can only loosely related to spinal control mechanisms, with only studies providing inter-vertebral information discussing possible mechanisms at the motion segment level. Indeed the objectives were so varied that making comparisons between studies was difficult. That said, the majority of studies do consider stabilisation, at least in a broad sense, and the following insights were provided.

McGill and Kippers (1994) suggested that an insight into interaction between sub-systems can be found by examining the transfer of moment from active to passive tissues at the limits of forward bending. Their investigation concluded that although electrically silent during full flexion, paraspinal muscles continue to provide elastic resistance via passive stretching. They suggest that this silence is an indication of the cessation of input from the central nervous system, likely as a result of some sort of active or passive tissue feedback. As the study was based on regional spinal measurements, nothing more than generalised theories could be extrapolated. In agreement with McGill and Kippers and again highlighting a requirement for inter-vertebral data, Arjmand et al. (2010) showed that in both models increased abdominal

coactivity was predicted at the end of forward flexion. This mechanism is proposed by both studies to counterbalance moments in addition to the contributions of paraspinal muscles (passive) and spinal ligaments.

In agreement with these studies, Paquet et al. (1994) suggested that increased paraspinal activity permits the transmission of forces via these muscles, and is a mechanism to protect damaged passive structures. It was proposed that the alteration in hip-spine movement pattern in those with an absent FRP, may be a strategy to protect the lumbar spine near its maximum range (i.e. near its peak bending moment). This raises the importance of being able to measure kinematics in different regions of a chain (i.e. not just the lumbar region). Callaghan and Dunk 2002 found that FRP was not present in the TES muscle during bending. As the study did not measure thoracic angular ROM however, and it is logical that considering the normal cascade of spinal flexion, some thoracic movement will have been expected to occur before the onset of movement in the lumbar region, it is difficult to comment on deactivation mechanisms. However, the results do support the common conclusion in FRP studies that as passives tissues are stretched, they eventually reach a point at which they can counter the moment produced by bending the lower back. In this case, as flexion moment may be expected to be less during slumped sitting than standing flexion, the passive tissues are able to support the moment produced at a smaller lumbar angle. This is as much detail as the authors provided, and so it was not possible to relate their findings to interactions between systems or feedback mechanisms. The study of Hashemirad et al. (2009) was based on the idea that flexibility is linked to characteristics of the active and passive tissues. The authors suggested that in agreement with Panjabi's hypotheses, when the CNS contends with increased flexibility in the passive tissues, it responds by increasing the contribution of the active system. This mechanism is represented in the study by the increased paraspinal activity associated with increased participant flexibility. The authors go on to suggest that such a mechanism is likely a spinal stabilisation strategy, however without inter-vertebral information this claim is difficult to support, and the reliability of the flexibility tests used could also be questioned.

Generally speaking therefore, increased muscle activity is proposed as a mechanism that increases spinal stability, the review did however provide some contrasting opinions. Peach et al. (1998) investigating healthy controls, found a lack of co-contraction of abdominal and paraspinal muscles during flexion. This therefore raises interesting questions concerning the purpose of co-contraction in LBP patients, and optimally efficient strategies employed by healthy spines. In this case no speculation was provided regarding subsystem interactions. This

is in contrast to the findings of Cholewicki et al. (1997) who showed that trunk flexor and extensor co-activation was present during dynamic sagittal movement. The study however only considered approximately 20° of flexion (i.e. around the neutral position) and cannot be compared directly with studies such as Peach et al. (1998) where full flexion was performed. The authors again conclude that the co-activation is a neuromuscular activation strategy to increase stability of the lumbar spine. As a regional kinematic study, it was not possible to extrapolate insights into system interactions, however the results do support Panjabi's theory that any loss of spinal stiffness as a result of passive tissue damage, can be compensated by an overall increase in trunk muscle activation. As such muscle activity may be useful as a clinical indicator. Further work was suggested which would benefit from investigations at the inter-vertebral level.

The findings of Sanchez-Zuriaga et al. (2015) question commonly held beliefs regarding spinal feedback mechanisms. Their results suggested that paraspinal activity was increased irrespective of the lumbar range of flexion achieved, and may therefore indicate that deactivation mechanisms are not purely related to mechanoreceptor thresholds as suggested elsewhere. Burnett et al. (2004) suggested that the LBP group in their study may have an underlying motor control dysfunction, either as a response to, or predisposing factor to a lumbar strain associated with the increased lower lumbar flexion and decreased local stabiliser activity. This is of course in direct contrast to the results of FRP studies considered in this review, which suggest that LBP is reflexively related to the increased activity of the paraspinals (i.e. the absence of the FRP). The authors also suggest that examining regions of the lumbar spine is more revealing than global, given the contrast in kinematic behaviours found between groups in terms of lumbar regions. In agreement Dankaerts et al. 2009 concluded that their results (found in both FP and AEP groups) likely represent maladaptive motor control strategies that potentially act as catalysts for ongoing strain and pain production, increase spinal load and result in impeded recovery. Yet again, no detail about the proposed mechanisms are provided, however the value of further dividing kinematic regions (i.e. upper and lower lumbar spine) was shown.

The study by Kaigle et al. (1998) was unique in that it was the only study reviewed with the capacity to comment on subsystem interactions at a motion segment level. In agreement with the theory that ligaments stretched in full flexion provide afferent impulses that then inhibit paraspinal muscles (Floyd and Silver 1955), the authors conclude that as the patient group showed comparatively reduced inter-vertebral movement, the ligamentous mechanoreceptors were not sufficiently stimulated to provoke muscular inhibition. Unfortunately, due to a small

sample size, the inter-vertebral data was pooled between levels, and so even this study did not provide a truly inter-vertebral insight, something that is arguably required to advance understanding in this area. Indeed, whilst the study of Arjmand was only small (n= 1), one of the author's key conclusions was that multi-joint kinematics combined with paraspinal EMG recordings would improve modelling accuracy.

Taking a slightly different slant on stabilisation mechanisms, O'Sullivan et al. (2006) discussed their findings in relation to global and local paraspinal activity (Bergmark 1989). The study showed that TES activity was extremely variable in participants during bending, a finding the authors suggested may be as a result of its role as a global muscle. As a globally acting muscle, it was argued to have more potential for variation in motor pattern, as it is was not directly responsible for local stabilisation as is the case for LMU. It may also be that the increase in TES activity is a strategy to maintain stability when LMU activity decreases, a mechanism perhaps employed to avoid excessive loading as a result of contraction (Granata and Marras 2000), or as additional resistance to the moment of flexion provided by the passive structures. In addition, Lariviere et al. (2000) showed that TES muscles likely compensate for LES muscles when less active (such as during FRP). The authors suggest therefore it is likely that TES muscles have an important role to play in LBP patient motor control strategies, and so consideration of thoracic muscle activity should perhaps be given, even when investigations are focussed on dynamic movement within the lumbar spine.

2.7.7.3 Can the information acquired by combining lumbar kinematic and muscle activity measurements during functional movements assist in distinguishing between groups of healthy controls and those with low back pain?

The review would suggest that there are many studies that have found distinguishing features in LBP populations, however, generally the study populations were small, and the large variations in methodology (particularly EMG placement and kinematic recordings) makes further analysis (including meta-analysis) difficult. There were also studies however that showed contrasting findings, or that were not able to distinguish between the two groups. The wide range of methodological approaches makes it difficult to generalise such findings beyond the specific populations involved, which is a major limitation of research in this field. Table 6, shows that in no two studies were the EMG electrode locations the same, and likewise all kinematic measurements differed in some way. This lack of standardisation makes the interpretation of results and contrasting study results very difficult, and so relationships between kinematic and EMG parameters and LBP are difficult to substantiate beyond the individual studies. The review does however highlight the potential of some variables for this

purpose. As an example, Kienbacher et al. (2015) using root mean square EMG amplitude, and regional measurements, showed that neuromuscular activation and kinematics can distinguish between CNSLBP patients with impaired or unimpaired muscle activation strategies. They suggest that the aging process is a stronger facilitator of this neuromuscular activity (i.e. increased paraspinal activity) than the pain associated with the condition. This the authors attribute to a likely increased excitability of the motor neurone pool associated with increased age. The overall increase in activity is again associated with a stabilisation strategy for all low back pain age groups.

This raises an important point, as it is unclear how pain can influence EMG and kinematic measurements, should studies focus on healthy participants, or perhaps LBP groups that are currently pain free, in order to account for the influence of pain? In the O'Sullivan et al. (2006) and Callaghan and Dunk (2002) studies, both investigated low back pain free populations, and therefore the disagreement in their results is most likely explained by methodological differences. The authors also suggest however that as TES activity is highly variable between individuals, this could possibly represent inherently different motor control strategies. In addition to O'Sullivan's findings (where no thoracic kinematic data was available), Nairn et al. (2013) measured thoracic movement, and showed that the deactivation of the TES during slumped sitting was related to increased angles of the thoracic segment movement. This supports the view that the decrease in activity is somehow related to stretch feedback of the ligaments, and the authors concluded that regional information was therefore important. In agreement, Lühring et al. (2015) argued that the global approach (i.e. global trunk angle) was less preferable to the local approach (i.e. lumbar angle) as the mechanism of FRP is proposed to be dependent on local lumbar structures. This is a logical conclusion to make, and in continuation it is likely preferable still to obtain inter-vertebral information that relates directly to the lumbar structures involved.

2.7.8 Conclusions

Many studies found differences in kinematic or EMG variables capable of distinguishing between LBP and healthy control groups, however the differences in methodology between studies mean that no broad generalisations can be made.

No one study set out with the explicit objective to explore sub-system interaction, however many did attempt to relate their findings to such mechanisms. A common weakness in study design was that studies used regional kinematic measurements, which can only ever at best,

provide a broad interpretation of sub-system interaction. It was therefore unsurprising that conclusions relating to system interaction were limited. The studies that did were arguably those that took a closer look using regions divided or inter-vertebral kinematic measurements (Dankaerts et al. 2009; Kaigle et al. 1998), and even these did not use truly inter-vertebral data, as the data was pooled from several inter-vertebral levels.

There is an apparent unmet need to better understand spinal stability and Panjabi's 1992 assertion that the passive, active and motor control systems need to act in concert for function to be optimal. If there are changes in one sub-system it is assumed that there will be changes in the other sub-systems to compensate. It would appear however that since Panjabi's seminal spinal control papers, not much has been learnt. This is perhaps partially due to the fact that studies either focus on sub-systems individually or that it has not been possible to study their interactions during dynamic tasks.

It has been shown that although it is possible to measure numerous variables relating to spinal function, until one can measure *in vivo* inter-vertebral dynamic kinematics and relate it to one of the other sub-systems in detail, it will not be possible to make significant progress in this area. This lack of progression was reflected in this review, and highlights the requirement for new approaches to research that incorporate these elements. Future studies should consider technologies that enable inter-vertebral measurements, not just in the lumbar spine but ideally throughout the thoracic, pelvic, hip and cervical regions too. It has been shown that stabilisation during forward bending can be influenced by the paraspinal muscle activity of both flexors and extensors, and abdominals, and that the TES may play an important role in lumbar stabilisation (Reeves et al. 2006; Van Dieen et al. 2003). These muscles should therefore be included in studies whenever possible. Standardisation of investigation methodologies is also recommended, as the current heterogeneity in approaches, makes any comparison between studies difficult.

Section 8

2.8.1 Summary and conclusions

It is believed that passive, active and neural control systems combine and interact in order to stabilise the lumbar spine during dynamic movements, and that the moments and reaction forces produced by the active and passive tissues provide equilibrium (Willard 2012). Previous studies' attempts to identify biomechanical factors associated with CNSLBP however, frequently demonstrate contradictions in their findings. This may be partially explained by the large array of potentially influential biomechanical factors, by the use of different methodologies, and the unknown biomechanical influence of pain. Therefore, in order to better determine biomechanical links with CNSLBP, it is first necessary to improve understanding of normal spinal biomechanics⁵.

Investigations into normal spinal biomechanics at an inter-vertebral level will provide a deeper insight into how interactions at this level contribute to the control mechanisms of the spine. In order to investigate the inter-play between the active, passive and neural control systems (Panjabi 1992a) during sagittal flexion, a method that combines continuous kinematic and EMG information is required, however the selection of an appropriate technique requires the consideration of many factors.

As a capacity to measure continuous inter-vertebral kinematic information is the principal methodological requirement, the options available are limited, and all have their own advantages and disadvantages. The use of skin surface markers is non-invasive, allows normal free-bending, and is not restricted in terms of range; however its ability to accurately and reliably measure inter-vertebral movement is inferior to other techniques. The use of imaging modalities such as x-ray, QF and MRI are therefore preferable in this regard.

There are exceptions (Olsson 1976), but typically the use of x-ray and MRI are limited by an inability to measure spinal movements continuously throughout their range, and cannot be used to acquire higher order variables such as initial attainment rate, or IV-RoMmax. These requirements can be met with the use of fluoroscopy however, and QF is a fluoroscopic measurement technique that has also undergone much standardisation (Breen et al. 2012).

The standardisation of QF (i.e. regulating movement rate and range using a motion frame and pelvic stabilisation) minimises the effects of behavioural variations in terms of the rate and range

⁵ For the purpose of this thesis, all references to normal biomechanics refers to populations who were free from low back pain (stating an absence of any historical low back pain) at the time of investigation.

of movement. Also, by restraining the pelvis, the technique effectively isolates movement to the lumbar spine, which is arguably essential if outcome measures are to be compared between individuals; however it is also associated with methodological disadvantages. The QF technique is currently confined to the measurement of a single spinal region (i.e. lumbar spine only), and can therefore not accommodate influences from the thoracic spine, pelvis or lower limbs. The restriction of range also means that if QF were combined with EMG, muscle activity information between the range limit and full sagittal flexion would be excluded, and the use of a pelvic restraint will reduce the influence of pelvic and lower limb musculature (although it could be argued that such activity is actually confounding).

Whilst desirable, there have been very few previous attempts to obtain concurrent inter-vertebral motion and EMG measurements (Kaigle et al. 1998), which is reflective of the historic technological limitations associated with doing so, and the ethical considerations linked with more invasive techniques. The development of QF means that if combined with EMG, the technology now exists that can provide a solution to these problems, and their concurrent use would be the first time these technologies have been combined in order to examine the biomechanics of the lumbar spine *in vivo*.

In this study therefore, a protocol was developed to investigate the relationships between inter-vertebral kinematics and muscle activity, whilst addressing many of the limitations associated with previous study designs. Whilst the study combines two pre-established technologies, this has not been done previously, and therefore merging two existing technologies in order to create new knowledge is novel. The reliability and agreement of the kinematic and EMG amplitude measures used in this study will also be determined.

IV-RoMmax was selected as the primary variable, and in consideration of the literature review and some preliminary research (Chapter 3), the following hypotheses, aims, research questions and objectives were developed.

2.8.2 Study hypotheses, aims, research question and objectives

2.8.2.1 Hypothesis

- Relationships will be found between lordosis, kinematic (i.e. IV-RoMmax and initial attainment rate) and muscle activity variables and the IV-RoMmax of inter-vertebral levels between L2 and S1 during standardised weight-bearing sagittal flexion

2.8.2.2 Sub hypotheses

- There will be an inverse relationship between muscle activity and the IV-RoMmax

- There will be an inverse relationship between the LES/TES ratio and IV-RoMmax
- There will be a direct relationship between the size of muscle deactivation and IV-RoMmax
- There will be a direct relationship between the size of lordosis and IV-RoMmax in the upper lumbar segments, and vice versa
- There will be a direct relationship between initial attainment rate and IV-RoMmax at the same level
- There will be an inverse relationship between IV-RoMmax in the upper and IV-RoMmax in the lower lumbar segments

2.8.2.3 Aim

- The aim of this thesis is to investigate the relationships that exist between lumbar inter-vertebral motion and lumbar spinal muscle electrical activity in healthy adults during standardised weight-bearing forward bending

2.8.2.4 Secondary aims

- To investigate the relationships that exist between lumbar inter-vertebral motion and other lumbar kinematic variables (including lordosis) in healthy adults during standardised weight-bearing forward bending

2.8.2.5 Research question

Do any of the morphological (i.e. lordosis), kinematic or muscle activity parameters investigated demonstrate significant relationships with IV-RoMmax?

2.8.2.6 Objectives

- Develop a protocol that combines QF and sEMG technologies in order to address the study's hypotheses
- Determine the reliability and agreement of QF IV-RoMmax and initial attainment rate, and (sEMG) RMS amplitude measurements recorded during a standardised weight-bearing sagittal flexion protocol
- Determine the relationships between lordosis and the IV-RoMmax achieved during a standardised weight-bearing sagittal flexion protocol
- Determine the relationships between initial attainment rate and the IV-RoMmax achieved during a standardised weight-bearing sagittal flexion protocol
- Determine the relationships between IV-RoMmax at other inter-vertebral levels and the IV-RoMmax at a specific lumbar level achieved during a standardised weight-bearing sagittal flexion protocol

- Determine the relationships between mean sEMG muscle activity ratios and the IV-RoMmax achieved during a standardised weight-bearing sagittal flexion protocol
- Determine the relationships between sEMG inter-level muscle activity ratios and the IV-RoMmax achieved during a standardised weight-bearing sagittal flexion protocol
- Determine the relationships between sEMG amplitude changes and the IV-RoMmax achieved during a standardised weight-bearing sagittal flexion protocol

In order to address these objectives, an appropriate study protocol was developed, which forms the basis of chapter 4. Prior to this however, and in addition to the literature review, the following chapter outlines the preliminary works that were conducted to inform various aspects of the study's design. These include investigations into, 1. The most appropriate plane of motion for study, 2. The relationships between lordosis and lumbar IV-RoMmax, and 3. The effect of electrode displacement on sEMG signal amplitude.

Chapter 3: Preliminary studies

The following preliminary studies were conducted to inform the design of the main study. The initial proposal for the project funding of this study outlined the investigation of the biomechanics of the lumbar spine in lateral bending. As supported by the literature review however, the sagittal plane is most frequently investigated, and so an exploration of the benefits and drawbacks of investigations into both sagittal and coronal planes was required. The first preliminary study therefore considered which of the two planes would be most suitable to investigate, by examining recumbent and weight-bearing, coronal and sagittal flexion motion data. The second study investigated the relationship between lordosis and IV-RoMmax. This was an area where gaps in the literature were apparent, and the study was therefore required to support the development of a study hypothesis relating to these parameters (section 2.7.2.2). Finally the third study investigated the effect of electrode displacement on sEMG signal amplitude, providing an indication of how important electrode placement accuracy would be in the main study.

3.1 Choosing an appropriate plane of motion

3.1.1. Introduction

The purpose of this preliminary study was to compare lumbar spinal kinematics (measured using QF) between coronal and sagittal planes in both recumbent and weight-bearing protocols, in order to gain an insight into which plane may be best suited for further investigation. As the muscle activity during recumbent examinations may be considered negligible (Mellor 2009), it was hypothesised that any difference in the kinematic behaviour found during weight-bearing may be partially attributable to the associated muscle activity. An investigation incorporating both recumbent and weight bearing data was therefore carried out to determine the feasibility of the coronal and sagittal planes in the context of this study.

3.1.2 Method

Coronal images from five participants and sagittal images from ten participants were selected from an on-going normative database study (Section 4.2.1) to be marked-up and analysed. Coronal image sequences of participants' left and right lateral flexion whilst recumbent and weight bearing at baseline and follow up at six weeks (40 motion graphs in total) were analysed. Sagittal image sequences of participants' flexion and return whilst recumbent and weight-bearing, were analysed at baseline only (20 motion graphs in total). Details of the marking-up process and protocol can be found elsewhere (Section 4.2.6). IV-RoMmax, motion share

(calculated as each motion segments proportional contribution to overall L2-S1 angular range) and initial attainment rate values were read off the Graphical User Interface (GUI) (Figure 40) for ten participants in both the sagittal recumbent and weight bearing examinations.

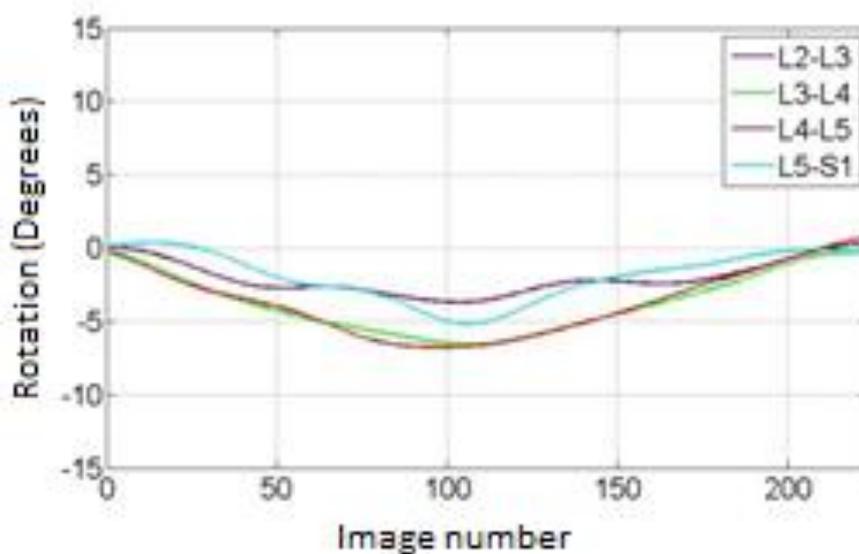
3.1.3 Data Analysis

Graphs from both planes (e.g. Figures 25-28) were analysed visually for patterns and coronal and sagittal plane graphs were compared. Box and whisker plots were produced from the sagittal plane data. The normality of each data set was tested using the Shapiro-Wilk test. IV-RoMmax, motion share and initial attainment rate during sagittal recumbent and weight-bearing sagittal examinations were compared using the Wilcoxon signed rank test for non-parametric data, and the paired t-test for parametric data. All statistical tests were performed using IBM SPSS (Version 21). Tables of raw data can be viewed (Appendix K).

3.1.4 Results

Figures (25-28) are examples of the inter-vertebral range of motion (IV-ROM) graphs produced for each plane in one participant. The x-axis represents image number⁶, and the y-axis motion segment angular rotation in degrees. The coloured key shows the individual inter-vertebral levels.

Figure 25: A sagittal plane recumbent (flexion and return) motion graph



⁶ Image number can also be considered a surrogate for time (e.g. the entire sagittal plane flexion and return sequence (Figure 25)) represents approximately 20 seconds.

Figure 26: A sagittal plane weight-bearing (flexion and return) motion graph

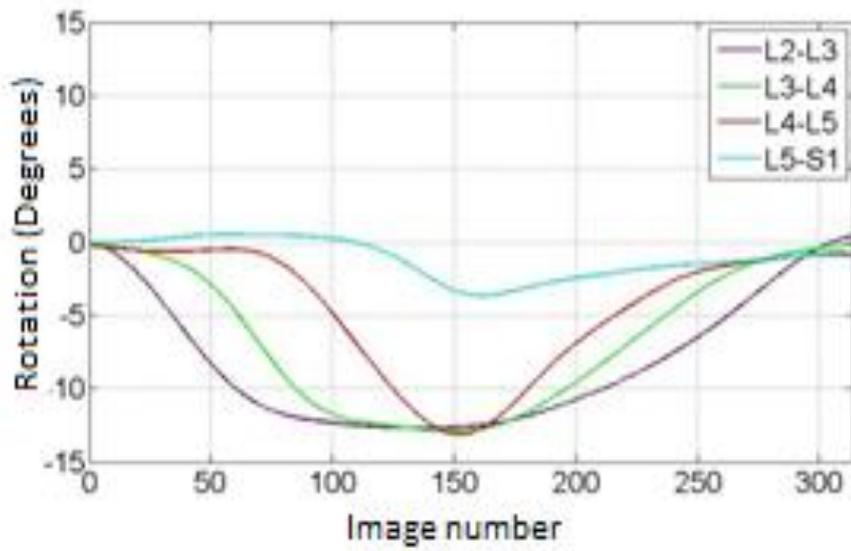


Figure 27: A coronal plane recumbent (bending to the left) motion graph

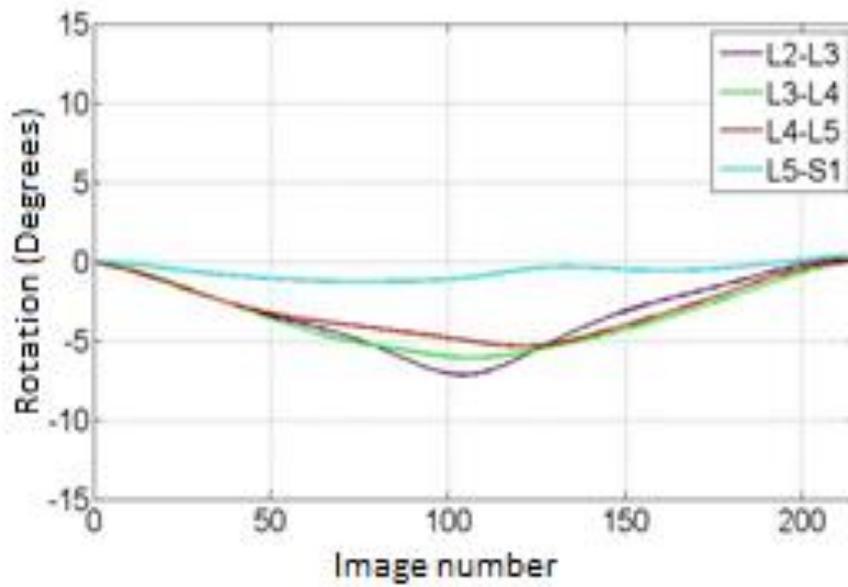
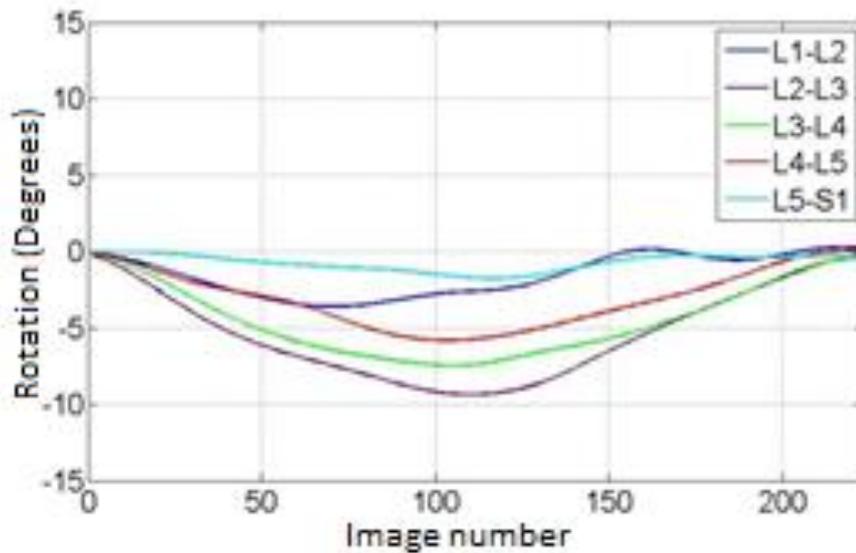


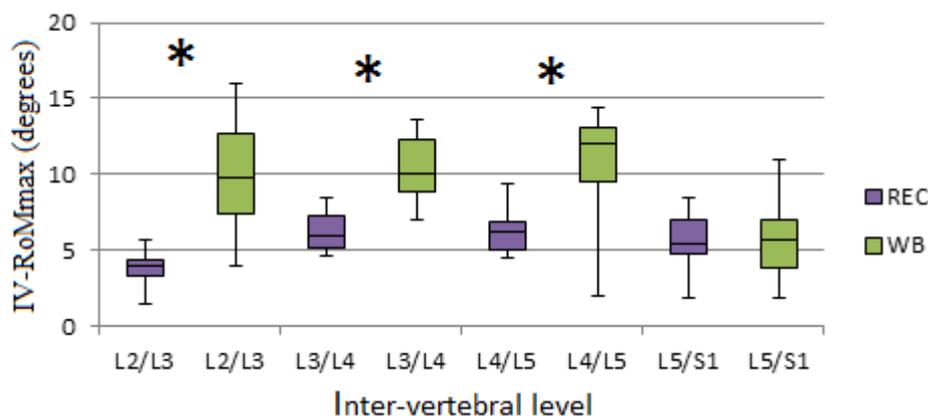
Figure 28: A coronal plane weight-bearing (bending to the left) motion graph



Note: The coronal plane weight-bearing motion graph includes an additional inter-vertebral level (L1-L2) represented by the dark blue line. This highlights the fact that if the L1 vertebra is visible in the x-ray image throughout the cycle, then it can be included. This is not possible in the majority of cases when L5-S1 is also required.

In the comparison of sagittal plane recumbent and weight-bearing data, a statistically significant difference between groups was taken as a two tailed p-value of < 0.05. Significant differences were found in IV-ROMmax at inter-vertebral levels L2-L3, L3-L4 and L4-L5 (Figure 29), motion share at L2-L3 and L5-S1 (Figure 30), and initial attainment rate at L2-L3 (Figure 31).

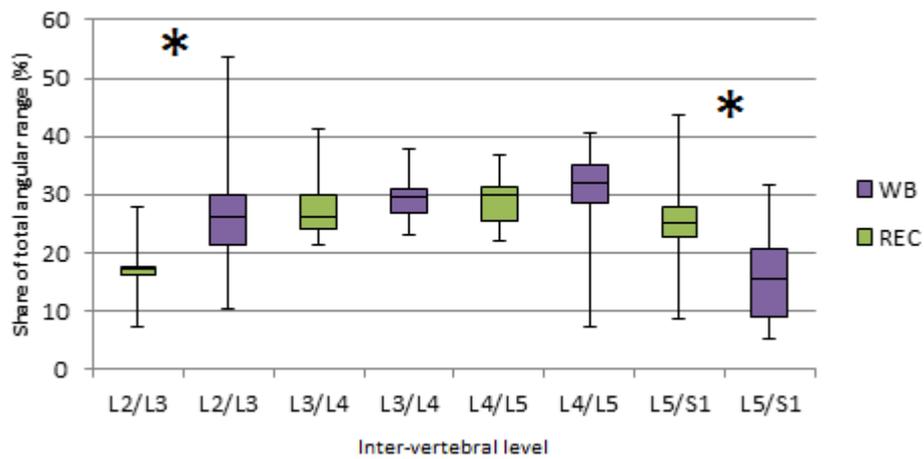
Figure 29: Box and whisker plot comparing IV-RoMmax between recumbent and weight-bearing groups during sagittal flexion and return



Note: The black horizontal line within each box represents the mean of the data. The top and bottom whiskers represent the maximum and minimum data values. The top and bottom sides of each box represent the upper and lower quartiles of each data set. Statistically significant differences (< 0.05) between recumbent and weight-bearing measurements at each inter-vertebral level are represented by *.

The most notable differences between recumbent and weight-bearing groups are seen in the IV-ROMmax at all levels excluding L5-S1 (Figure 29). This was to be expected however as the sagittal QF weight-bearing protocol rotates the participants an extra 20° (60° in total compared to 40° in recumbent protocol, see section 4.2.8) and so any comparison in terms of IV-ROMmax should be interpreted with care. It should be noted however that in the majority of cases, IV-RoMmax is reached before 40° of forward flexion during this protocol and therefore the results may actually be reflective of a true difference between groups. IV-RoMmax was therefore included for completeness.

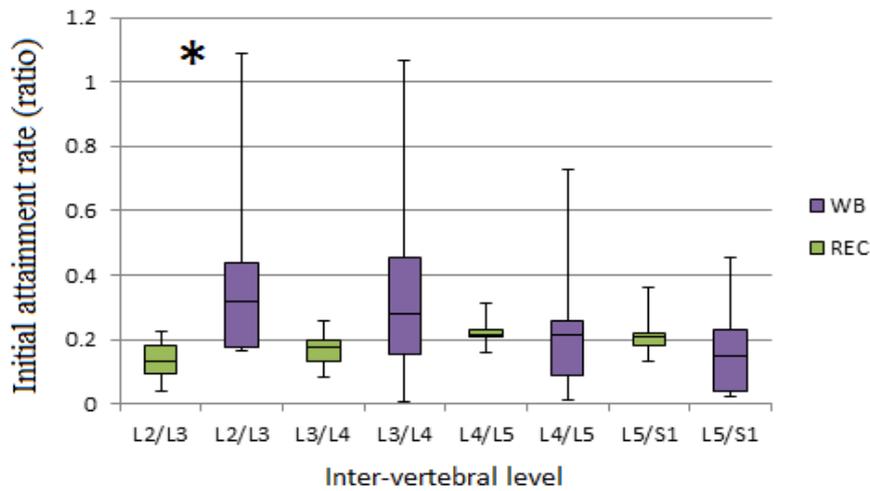
Figure 30: Box and whisker plot comparing the percentage of motion share between recumbent and weight-bearing groups during sagittal flexion and return



Note: Statistically significant differences (< 0.05) between recumbent and weight-bearing measurements at each inter-vertebral level are represented by *.

In terms of motion share, the contribution of the mid-lumbar levels (L3-4, L4-5) appears to be similar between examination types. At the upper and lower ends of the lumbar spine (L2-3, L5-S1) however, significant differences are seen (Figure 30). It seems that during weight bearing the inter-vertebral rotation at L5-S1 is lower than in the lying examination, whilst L2-L3 moves comparatively more.

Figure 31: Box and whisker plot comparing initial attainment rate between recumbent and weight-bearing groups during sagittal flexion and return



Note: Statistically significant differences (< 0.05) between recumbent and weight-bearing measurements at each inter-vertebral level are represented by *.

A significant difference in initial attainment rate was found only at the level of L2-L3 (Figure 31). It appears from the results however that the range of initial attainment rate amongst the 10 participants is notably more variable at all levels whilst weight-bearing (Figure 31). Generally the upper most segments of the lumbar spine have greater initial attainment rate whilst weight-bearing.

3.1.5 Discussion

3.1.5.1 Planes of investigation

There are three planes of motion to choose from when assessing the kinematics of the spine, sagittal (flexion and extension), coronal (side-bending) and transverse (axial rotation) (Appendix B). It is not feasible to assess the transverse plane with uniplanar QF technology, as the fluoroscope would need to be placed above and below the spine, which is not possible. Using a biplanar radiographic technique however, Pearcy (1985) managed to demonstrate that there was only minimal movement of the lumbar spine during axial rotation (2-3 degrees of axial rotation during upright posture) (Pearcy 1985), which further made rotation less appealing in terms of this study. Also, as the rotator muscles (which are chiefly responsible for rotation) are very small and deep, any investigation into the associated muscle activity would most likely require needle EMG to measure accurately.

3.1.5.2 Visual analysis of the motion graphs

The decision to focus on either sagittal or coronal movements was more complicated. It has been shown that muscles remain relatively silent during the recumbent QF examinations (Mellor 2009), so it is logical to assume that muscular activity (along with loading) may have a role in any changes in kinematic patterns found during weight-bearing. A visual comparison of the IV-ROM graphs for coronal and sagittal planes (Figures 25-28) highlights some general differences between the planes and between the weight-bearing and recumbent groups. Firstly in the sagittal plane during weight-bearing (Figure 26), there is a marked difference in the shape of the motion pattern compared to all other groups (i.e. coronal weight-bearing and recumbent, and sagittal recumbent). The example (Figure 26) shows delays between segmental movement initiation, steeper outward curves and a larger range of angular rotation. This type of pattern being in contrast to the recumbent sagittal motion (Figure 25) where shallower, more evenly distributed curves are seen. The coronal plane kinematics did not display such distinctive differences between recumbent and weight-bearing groups, with a notably more similar pattern (Figures 27 and 28). It could be argued that the more similar the kinematic patterns in weight-bearing and recumbent examinations, the more difficult it will be to find relationships between kinematic behaviour and muscle activity, when the muscles are perhaps less influential during weight-bearing.

Due to the apparently greater impact of weight-bearing on kinematic behaviour visualised in the sagittal plane, focus for statistical comparisons was placed on this plane. Some of the possible effects that weight-bearing can have on kinematic patterns (partially attributable to increased muscle activity) are demonstrated in (Figures 29-31). All three kinematic variables under investigation, demonstrated significant differences between recumbent and weight-bearing examinations at one or more inter-vertebral levels. For all of these variables, it is known that the timing, rate, and range of inter-vertebral rotation are regulated by moments exerted on the vertebrae. If the focus is on forward bending, these moments are produced by the weight of the participant's head and body during flexion or by agonist muscle activity (Bogduk 1995). During flexion these moments are resisted by paraspinal muscles, the longitudinally orientated ligaments, the facet joint capsules and the annulus fibrosus of the disc. Bogduk (1995) goes on to suggest that during an activity such as forward bending, an increased IV-RoM for example, is unlikely to be the result of increased agonist muscle activity, rather a reduction in restraint due to weakened muscles, impaired ligaments or reduced disc tension (Bogduk 1995). Muscle activation then, is likely a key component of the control of weight-bearing lumbar spinal flexion, and is highly likely to relate to the kinematic variables described.

3.1.5.3 The pros and cons of investigating the coronal plane

A major benefit of investigating the coronal plane is that in terms of muscular activity there is a very clear relationship between the agonist paraspinal activity and the movement of lateral bending (i.e. in order for an individual to bend to the right, the right erector spinae will activate, and typically the left erector spinae will relax), yet according to Lariviere et al. (2000) the most commonly studied plane is the sagittal (Lariviere et al. 2000). There are many possible reasons for this apparent preference in the research, especially when considering the use of EMG. For example, there are potential issues when using the coronal plane in terms of additional sources of EMG signal contamination. There is an increased risk of creasing of the skin during side-bending manoeuvres which can contaminate the signal, and due to the close proximity of electrodes required to record from multiple levels of the lumbar musculature simultaneously, side-bending can result in electrodes touching, a problem less likely to be seen during flexion where a small separation is seen.

Kasman (1997) states that “EMG findings on the sagittal plane are more discriminate between subjects who are healthy and those with chronic dysfunction and that side-bending manoeuvres are subsequently of subsidiary interest” (Kasman et al. 1997). It has also been shown that differences in sEMG measurements are much more pronounced in the sagittal plane compared to the coronal (Van Dieen et al. 2003), a beneficial quality when determining patterns from the recorded data. The relatively smaller changes observed during side-bending could be problematic, as activity patterns are less distinctive, and with only a small difference between active recordings and those at baseline (during rest), the interpretation of true muscle activity becomes more difficult.

In terms of the kinematic measurements, Pearcy (1985) also demonstrated that while there is only minimal axial rotation and lateral bending accompanying flexion and extension movements, lateral bending was associated with a substantial amount of concurrent axial rotation (Pearcy 1985). Although a previous QF based study concluded that up to 10° of out of plane motion does not significantly affect inter-vertebral angle measurement accuracy (Breen et al. 2006), this lack of coupled movement, combined with the relatively larger segmental motion in the sagittal plane, perhaps make it preferable to the coronal in terms of kinematic assessment (Edwards et al. 1987; Keessen 1984).

3.1.5.4 Muscle activity onset and offset parameters

One of the sEMG variables considered for use in the main study was signal onset and offset. If relationships were to be found between muscle activity onset or offset and patterns of spinal movement, they were most likely if temporal patterns also existed within the spinal segmental

movement. The results show that the only configuration that regularly demonstrated such phase lag, was sagittal weight-bearing (Figure 26), and was therefore arguably the most suitable option if signal onset or offset parameters were to be used.

Given the evidence of a 'phase lag' in terms of inter-vertebral movement onset observed in the weight-bearing sagittal plane motion graphs, it was hypothesised that such delays may be related to the surrounding musculature preventing the initialisation of the movement, and then subsequently allowing the motion to occur (via deactivation) later in the flexion phase. The potential link between phase lag and muscle activity onset and offset was therefore considered to be of interest.

Note: During the sEMG reliability and agreement studies (Section 5.2) it became clear that determining muscle activity onset was not possible using the current weight-bearing protocol. In simple terms, it was difficult to obtain relaxation at all three muscle levels at the same time, at the forward bending starting position (i.e. neutral upright standing position). It has been suggested that during neutral standing, participants will have a tendency to fall forwards or backwards depending on their centre of gravity (Floyd and Silver 1955), and that males in particular tend to stand in a posture of slight flexion (Norton 2004). As the sagittal alignment of the majority of participants was apparently anterior to the L5-S1 disc, there was generally an intermittent or constant activity in one or more of the paraspinal levels whilst standing, hence the naming of the erector spinae muscle group, as they maintain the spine in the erect position, or return it to this position after movement (Kippers and Parker 1984). In terms of signal offset, the sEMG repeatability studies also highlighted the fact that the range of forward flexion (60°) performed by participants during the weight-bearing protocol was not sufficient to initiate the FRP phenomenon. Therefore whilst deactivation of muscle activity may have begun, complete electrical silence will typically not occur.

3.1.5.5 The influence of kinematic behaviours during sagittal weight-bearing on sEMG recording site positioning decisions

The IV-RoMmax, motion share, initial attainment rate and phase lag variables were all demonstrated to be viable options for use in the main study, and the significant differences in these variables observed between weight-bearing and recumbent sagittal groups influenced decisions regarding sEMG electrode positioning decisions. The upper and lower sections of the lumbar spine were of particular interest, as the L2-3 level for example demonstrated significant differences for all variables measured, as did L5-S1 in terms of motion share. Significant differences were less frequently observed in the mid-lumbar spine (although phase lag was regularly observed at these levels). The lumbar lordosis was also a consideration, as it was believed that muscle activity would vary dependent on spinal curvature, and so it was decided on balance that the upper and lower ends of the lumbar spine would be the most appropriate sites for sEMG recordings. In order to examine potential relationships between kinematic behaviour and the more globally acting paraspinal musculature, sEMG recording from an additional thoracic level was also incorporated into the study design (Section 4.2.15).

3.1.6 Conclusion

It has been shown that lumbar spinal kinematics in the sagittal plane demonstrate more variation in inter-vertebral phase lag, greater initial motion curve steepness, and greater variation in angular range than those exhibited during lateral bending. When examining the sagittal plane further, significant differences in all kinematic variables were found between weight-bearing and recumbent groups. It is proposed that these differences will partially be due to the muscular activity associated with the weight-bearing examination, which lends weight to the argument that relationships could most readily be found between lumbar kinematics and muscle activity during a weight-bearing sagittal plane protocol. The ease of collection, heterogeneity (between participants) and the relative size of variables achievable in the sagittal plane, also better lend themselves to the detection of patterns within them. In light of these conclusions, and in addition to the findings of the literature review, the sagittal plane was selected as the plane of investigation for the main study.

Note: It cannot be assumed that the activity of muscles is entirely responsible for the differences in kinematic behaviour observed between recumbent and weight-bearing participants, as passive influences such as the discs, ligaments, bony anatomy and sagittal alignment may also change when weight-bearing. For example, it has been suggested that lordosis during neutral standing is comparable to that of a patient in supine position, with legs straight (Been and Kalichman 2014), the recumbent protocol used in the normative study (section 4.2.1) however requires side-lying and bent legs, which results in a different resting lordosis to the weight-bearing cycle, and may therefore have affected the movement patterns. The possible relationships between lordosis and lumbar spinal kinematics (i.e. IV-RoMmax) are explored in the next section.

3.2 An exploration into the relationships between the degree of lordosis and lumbar IV-RoMmax during weight-bearing sagittal flexion, and a visual analysis of the order and magnitude of inter-vertebral movements.

3.2.1 Introduction

The literature review was unable to provide a clear idea as to how IV-RoMmax may change in relation to changes in lordosis. It was therefore necessary to conduct the following study to help inform the development of a study sub-hypothesis (Section 2.7.2.2). This preliminary study therefore had two main aims. As the degree of lumbar lordosis is thought to influence the kinematics of the lumbar spine (Keorochana 2011), the first aim was to explore how the degree of lordosis relates to lumbar IV-RoMmax. The second aim was to determine the order of segmental movement initiation and the magnitude of inter-vertebral movements. Information regarding the order of movement initiation and the angular range achieved at different inter-vertebral levels may provide an insight into the likely associated spinal muscle activity.

3.2.2 Methods

The data retrieved from the 10 weight-bearing sagittal plane images taken from the 10 participants used in Preliminary Study 1 (Section 3.1.2), were re-used in this study. The mean angular ranges of each inter-vertebral level during weight-bearing sagittal plane flexion and return were compared when the starting lordosis angle was divided into 3 groups within its total range, group A = 30-45°, B = 46-60° and C=61-75°. The angle of lordosis was calculated as the sum of all absolute inter-vertebral angles (L2-L3 – L5-S1) taken from the first QF image. For interpretation purposes, group A was taken as the lower range, group B as the mid-range, and Group C as the higher range of normal lumbar lordosis. Motion graphs were also visually ranked according to angular range (Table 8) and in the order of segmental onset (Table 9). The data were tabulated and colour coded in order for patterns to be easily distinguished. (The splitting of the lordotic angle into 3 groups was done arbitrarily, and may not be representative of normal ranges in other populations).

3.2.3 Data analysis

The results were analysed using visual interpretation of the graph (Figure 32) and tables (Tables 7,8 and 9).

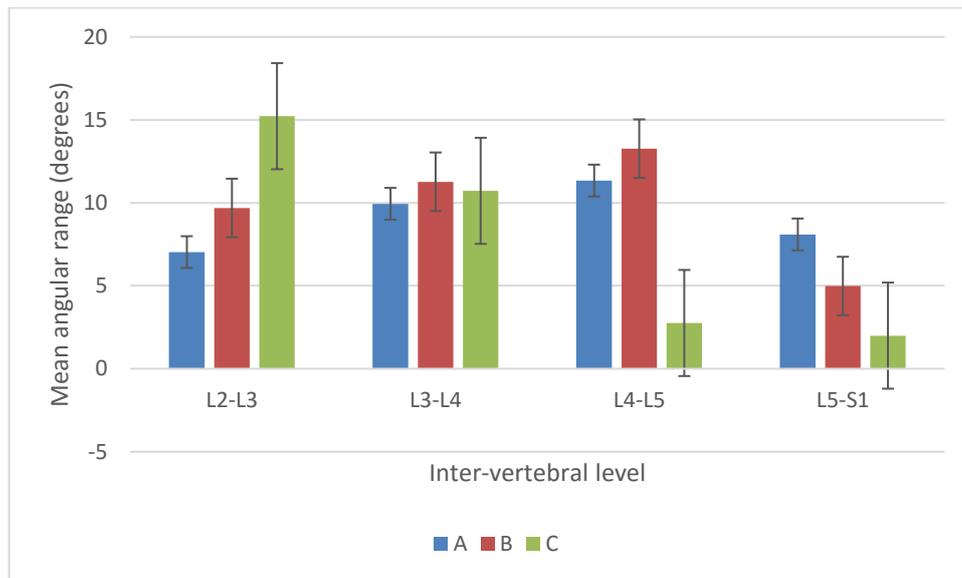
3.2.4 Results

The results show how large angular ranges reached by the uppermost lumbar level (L2-L3) appear to relate to a starting lordosis angle in the upper range of normal (Table 7) and (Figure 32). In participants with a starting lordosis in the lowest range, the opposite of this pattern occurs, with comparatively higher angular ranges in this group at L5-S1. The lordosis would appear to initially flatten from the top in those participants with greater lordotic curvature.

Table 7: IV-RoMmax for each inter-vertebral level and lordosis groups

		L2-L3	L3-L4	L4-L5	L5-S1
Group A 30-45 degrees	LS1	9.8	9.7	11.4	6.2
	LS2	7.3	9.9	9.3	7.1
	LS3	4.0	10.2	13.3	11.0
	Mean	7.0	9.9	11.3	8.1
	SD	2.9	0.2	2.0	2.6
Group B 46-60 degrees	LS4	7.6	10.5	12.6	5.1
	LS5	7.0	8.0	13.0	6.9
	LS6	12.7	12.9	13.1	3.7
	LS7	11.5	13.6	14.4	4.3
	Mean	9.7	11.3	13.3	5.0
	SD	2.8	2.5	0.8	1.4
Group C 61-75 degrees	LS8	16.0	13.0	3.6	1.8
	LS9	14.5	8.5	2.0	2.2
	LS10	15.0	10.5	2.6	2.0
	Mean	15.1	10.7	2.7	2.0
	SD	0.7	2.2	0.8	0.2

Figure 32: Mean angular range of inter-vertebral levels during sagittal weight-bearing flexion and return when the starting lordosis angle is divided into 3 groups



Note: A = a lordosis between 30-45° (n = 3), B = a lordosis between 46-60° (n = 4) and C= a lordosis between 61-75° (n = 3). Standard error bars included.

Visual analysis of the motion graphs revealed that the inter-vertebral movement sequence is varied in this convenience sample (Table 9). This is in agreement with Gatton (1999) who categorised the four most frequent movement sequences ‘top down’, ‘bottom up’, ‘middle last’, and ‘all together’ (Gatton 1999). The data were analysed for movement patterns in the same way, and also for which level reached the largest range (Table 8).

Table 8: Ranking of inter-vertebral levels in order of largest IV-RoMmax

Participant ID	Rank 1	Rank 2	Rank 3	Rank 4
RS066	L4/L5	L3/L4	L2/L3	L5/S1
RS027	L3/L4	L4/L5	L2/L3	L5/S1
NS023	L2/L3	L3/L4	L4/L5	L5/S1
RS055	L2/L3	L3/L4	L4/L5	L5/S1
RS014	L4/L5	L5/S1	L3/L4	L2/L3
NS006	L4/L5	L5/S1	L2/L3	
NS001	L4/L5	L3/L4	L2/L3	L5/S1
RS050	L4/L5	L3/L4	L2/L3	L5/S1
RS013	L4/L5	L3/L4	L2/L3	L5/S1
NS003	L4/L5	L3/L4	L2/L3	L5/S1

Note: Column '1' indicates the level with greatest range and '4' the smallest. Yellow boxes represent L2-L3, red boxes represent L3-L4, blue boxes represent L4-L5 and green boxes represent L5-S1

Weight-bearing sequences show a tendency for phase lag, and a cascade in inter-vertebral movement procession. The most common pattern in this sample was 'top down' e.g. L2-3 then L3-4 then L4-L5 then L5-S1 (Table 9). In terms of the inter-vertebral level that reaches the largest angular range, the most common pattern was to see L4-L5 move the furthest, and L5-S1 to move the least (Table 8). There were notable exceptions in participants NS023 and RS055. It was hypothesised that the main study would show demonstrably different muscle recruitment patterns between such contrasting kinematic patterns.

Table 9: Ranking of inter-vertebral levels in order of movement initiation sequence

Participant ID	Rank 1	Rank 2	Rank 3	Rank 4
RS066	L2/L3	L3/L4	L4/L5	L5/S1
RS027	L3/L4	L2/L3	L4/L5	L5/S1
NS023	L2/L3	L4/L5	L3/L4	L5/S1
RS055	L2/L3	L3/L4	L4/L5	L5/S1
RS014	L3/L4	L2/L3	L4/L5	L5/S1
NS006	L2/L3	L3/L4	L4/L5	L5/S1
NS001	L4/L5	L3/L4	L5/S1	L2/L3
RS050	L2/L3	L3/L4	L4/L5	L5/S1
RS013	L2/L3	L3/L4	L4/L5	L5/S1
NS003	L2/L3	L3/L4	L4/L5	L5/S1

Note: Column '1' showing the first segmental pair to move and column '4' the last. Yellow boxes represent L2-L3, red boxes represent L3-L4, blue boxes represent L4-L5 and green boxes represent L5-S1

3.2.5 Discussion

Lumbar lordosis increases when the spine is in the standing position (Fernand 1985). A study of 300 asymptomatic participants by Vialle et al. (2005) measured standing lordosis as the angle between the superior endplate of L1 and the inferior endplate of L5 (Vialle et al. 2005). They found that the mean lordosis in this population was 43° (SD 11.2) (Range 13.6-69°). No direct comparison can be made due to differences in measurement method and sample size, but our small study mean was 52° (SD 11.1) (Range 36.8-69.5°).

In a review of methods that evaluate sagittal plane curvature in 2D images, Vrtovec et al (2009) commented on the limitations of the widely used 'modified Cobb method', suggesting that although the technique reflects endplate tilt, it is not revealing regarding regional changes in the curve (Vrtovec et al. 2009). The method used in the current study accounted for variation between the caudal and cephalic measurement point, by taking the sum of all lumbar intervertebral angles. It would appear that the lumbar lordosis flattens predominantly from the top in those with a greater curvature. These results agree with Keorochana (2011) who concluded that differences in lumbar lordosis may be associated with such differences in the lumbar spine kinematics (Keorochana 2011).

The spinal kinematic patterns produced during sagittal flexion (Tables 3 and 4) indicate a degree of variation between individuals. The literature is conflicting, and is undecided as to whether lumbar segments begin their movement simultaneously (Ahmadi et al. 2009; Lee 2002; Wong 2006; Wong 2004), sequentially (Kanayama 1996) or a mixture of the two (Okawa 1998; Takayanagi 2001). This led Ahmadi et al (2009) to comment that a 'normal' movement pattern of the lumbar spine during flexion is yet to be determined (Ahmadi et al. 2009). This study has shown a mixed range of movement patterns, with evidence of a lag between initiations of movement between levels frequently apparent in the weight-bearing group. This is in agreement with the findings of (Gatton 1999), however this study using QF has the advantage of analysing true segmental motion, which must be a consideration when compared to such investigations (e.g. Gatton (1999)) that used skin mounted sensors.

The order and magnitude of segmental movement are variables that can provide an indication as to the possible concurrent muscle activation patterns. It may be hypothesised for example that greater paraspinal muscle activity (larger sEMG amplitudes) will be recorded at levels demonstrating the smaller angular ranges relative to those levels with more rotation, and that decreased activity may also be found in the upper lumbar region in participants with a relatively greater lordosis. It may also be expected that sEMG signal offset will relate to the order of

segmental motion, for example in the most frequently observed 'top down' cascade we may see a delay in activity deactivation between L2 and L5, a pattern that may be reversed in segmental sequences that initiate from lower segments.

The results suggest that lordosis may have an influence over both kinematic and muscle activity behaviours during weight-bearing sagittal flexion.

3.2.6 Conclusion

The angle of lordosis taken from the first QF image appears to affect the subsequent kinematic patterns. This supports the inclusion of Lordosis for analysis in the main study. It is also logical to suggest that participants demonstrating different kinematic movement patterns such as movement initiation from the upper lumbar compared to those with initiation from the lower lumbar, may also have corresponding differences in the controlling muscle activity. This supports the recording of sEMG from both upper and lower sections of the lumbar spine (e.g. LES and LMU).

3.3 Electrode Displacement Study

3.3.1 Introduction

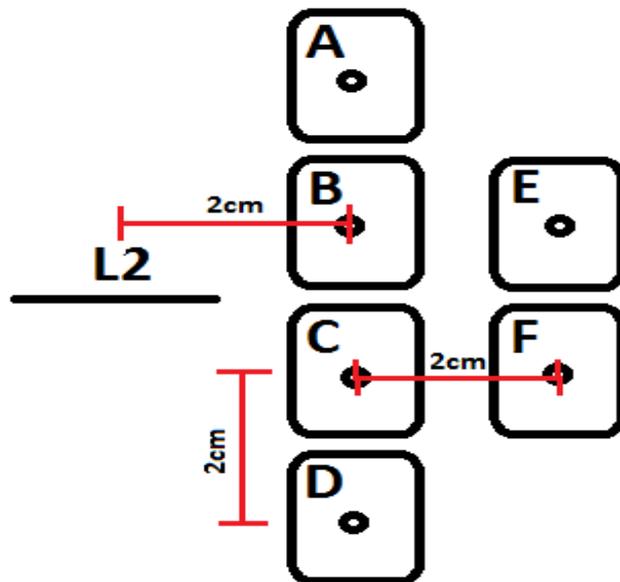
The positioning of sEMG electrodes can contribute significantly to variations in the recorded signal (De Nooij et al. 2009). In order to try and standardise sEMG recordings, guidelines have been developed recommending specific electrode application sites for each muscle under investigation (Hermens et al. 1999). These sites are localised on the basis of bony landmark palpation, a process that is dependent on the interpretation of the person applying the electrodes, and therefore subject to subjective error (Chakraverty et al. 2007; Kim et al. 2007; McGaugh et al. 2007). The purpose of this mini study was to assess the effect of electrode displacements (well-defined changes in electrode positions) on sEMG amplitude recordings, during the sagittal lumbar flexion and return QF protocol. The discussion section explores the possible implications of inaccurate electrode application.

3.3.2 Method

In order to assess the impact of electrode displacement on the sEMG amplitude recordings, a single participant was selected to perform the weight-bearing forward flexion and return examination protocol, without irradiation (Section 4.2.10). All results were taken from the mean RMS sEMG amplitude of 4 examination cycles.

Using the Iliac crest as an anatomical reference point to locate the L3 spinous process, electrodes were placed 2cm lateral to the L2 spinous process (see (BC) Figure 33). This electrode position is believed to record myoelectric activity from the lumbar longissimus muscle, and was used as the reference site from which to compare the amplitudes recorded from electrodes displaced 2cm vertically (AB higher), (CD lower), and 2cm horizontally (EF lateral) (Figure 33). In an adaptation of the investigation conducted by (De Nooij et al. 2009), the ratio between the mean RMS sEMG amplitude over the entire flexion and return cycle recorded from the displaced electrode sites and that from the reference at (BC) was calculated. Normalisation of the recorded data was not required due to the use of ratios, and there was no between subject comparison.

Figure 33: Electrode positions for the electrode displacement study



Note: Electrodes were applied to the participant's right side only. L2 indicates the position of the L2 spinous process.

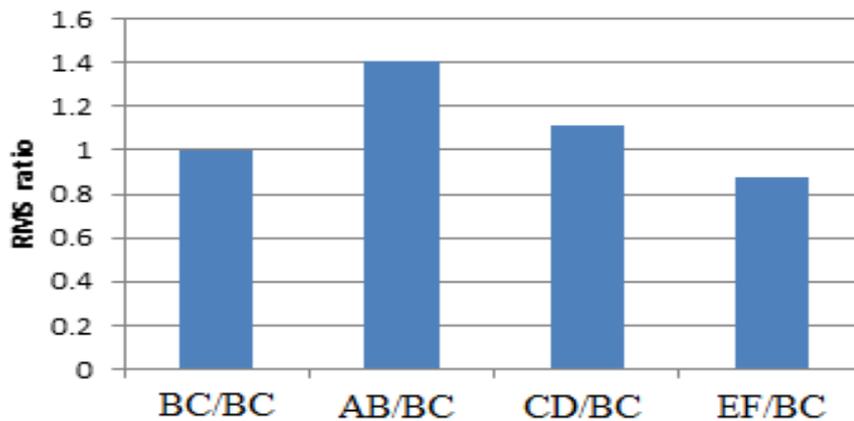
3.3.3 Data analysis

To assess the effect of electrode dislocation the ratios of the average RMS sEMG amplitudes for the displaced electrodes and the reference electrodes were calculated as follows BC/BC , AB/BC , CD/BC and EF/BC . Ratios above 1 indicate a relative increase in amplitude compared to the reference site and below 1 a relative decrease.

3.3.4 Results

Electrode dislocation has the most dramatic effect when moved in the vertical plane superiorly (AB), with a 40% increase in amplitude (Figure 34). All dislocation positions appear to affect the signal however, with an 11% increase at (BC), and a 12% decrease at (EF).

Figure 34: The effect of electrode displacement on signal amplitude during weight-bearing sagittal flexion and return



3.3.5 Discussion

As previously described, this mini study design was adapted from De Nooij et al. (2009) who used a reference at the level of L1, and found that lateral displacement resulted in a significant decrease in sEMG amplitude (De Nooij et al. 2009). In contrast to this study, they did not find longitudinal displacements to have a significant effect. The effect of electrode dislocation, especially in the superior direction has been shown in this study however, to affect the amplitude of the recorded signal, a factor that possibly contributes to the intra-subject variations observed in sEMG studies.

A small participant number means the results are in no way generalisable to a larger population, but the results do serve to highlight the potential problem of inaccurate electrode placement. Electrode placement has the potential to significantly affect signal amplitude, and therefore the interpretation of muscle activity patterns. This small study highlights the importance of electrode placement accuracy in the main study, particularly if muscle activity ratios are to be used as variables.

3.3.6 Conclusion

Accurate electrode positioning was an important aspect of the main study design, as electrode displacement has been shown to substantially alter the recorded signal. The main study design therefore incorporated a methodology that ensured electrode application was as accurate as possible. The novel technique used to improve this accuracy is outlined (Section 4.2.15).

3.4 Summary

These preliminary investigations helped to inform the design of the main study. They assisted in the decision to focus on the sagittal plane, and highlighted some of the benefits and drawbacks of the different kinematic variables available. It was concluded that IV-RoMmax, initial attainment rate, and lordosis had potential for use as parameters in the main study. Motion share was also considered to be of potential value, but in order to limit the complexity of analysis, it was decided that this variable would not be investigated further at this time. The importance of accurate sEMG electrode application was also highlighted.

The following chapter describes the QF and sEMG variables selected for investigation, and outlines the main study protocol.

Chapter 4: Methodology

4.1 Introduction

In Chapter 2 it was determined that QF and sEMG were the most appropriate techniques to provide concurrent information regarding inter-vertebral movement and myoelectric activity during sagittal forward bending of the lumbar spine. This chapter outlines the combined QF and sEMG methodology designed to address the study's aims and objectives, and describes the variables selected. The chapter consists of two sections. Section 1: describes the kinematic, morphological and sEMG variables selected for inclusion in the study, and section 2: outlines the main study methodology.

4.1.1 Section 1: Variables selected for investigation

The following section outlines the parameters that were selected for investigation in the main study. Their suitability for inclusion was based on the previous section and the literature review.

4.1.1.1 Main outcome variable: IV-RoMmax

As the primary aim of the study was to investigate biomechanical relationships, it was important to select an outcome variable that would likely demonstrate associations with other variables. IV-RoMmax can be considered as an indication of the resistance to inter-vertebral rotation during bending, and will therefore relate to other parameters representative of active and passive tissue function. These include the longitudinally orientated paraspinal muscles (characterised by sEMG amplitude measurements), as they are ideally positioned to resist sagittal flexion (Bogduk 2012), and the discs and ligaments (characterised by the initial attainment rate (Mellor F.E. et al. 2014)). IV-RoM is also a variable that is easily understood, that can be measured with precision (Breen et al. 2006), and demonstrates a high degree of heterogeneity (Deitz 2011), a characteristic that increased the probability of finding relationships between IV-RoMmax and other mechanical parameters.

4.1.1.2 Other QF variables

- Lumbar lordosis
- Initial attainment rate

4.1.1.3 sEMG variables

- The mean RMS sEMG amplitude (normalised to a sMVC) of TES, LES and LMU over the flexion phase of the cycle
- RMS sEMG amplitude ratios normalised to a sMVC (e.g. LMU/TES, LMU/LES and LES/TES)

- Changes in RMS sEMG amplitude normalised to a sMVC between 5 consecutive epochs over the flexion phase of the cycle⁷

4.2 Section 2: Main study methodology

4.2.1 Study design

This was an exploratory cross-sectional pilot study of healthy volunteers. All participants received a lumbar flexion QF examination in the sagittal plane with the concurrent sEMG recording of their lumbar paraspinal muscle activity.

This study was incorporated into an ongoing normative database study entitled: Characteristics of lumbar spine inter-vertebral kinematics in healthy adults and their reproducibility over time: A standardised reference and reliability study for future explanatory trials of mechanical interventions for non-specific back pain. The purpose of that ongoing work is to establish a database of the normal mechanics of the lumbar spine in people without low back pain, to which the kinematic data collected during this study, will contribute. By recruiting subjects that were eligible to participate in both studies, it was possible to avoid the unnecessary irradiation of additional participants.

4.2.2 Sample size

This study was the first of its kind, and there is no prior information from which to base a sample size on. It may therefore be considered an exploratory pilot trial, for which there is a minimum suggested sample size requirement of 12 participants (Julious 2005). The justifications for this sample size were based on feasibility, precision about the mean and variance, and regulatory considerations (Julious 2005). Previous studies using the QF technology acknowledge that a 20% loss due to technical issues, template tracking failures or drop outs should be anticipated (Branney and Breen 2014).

Due to the concurrent use of QF and sEMG technology, there was an increased risk that some component of the sEMG recording may also fail, resulting in unusable data for that participant. It was therefore decided that a minimum of 20 participants would be required to allow for a potential 40% combined data loss. The sample size was also limited due to time restraints, labour and equipment costs, and as the study was a sub-study of the above normative database

⁷ Note: To the author's knowledge, the change in RMS sEMG amplitude at different stages of the flexion cycle is a parameter that has not been reported elsewhere in the literature. As such the use of this parameter represents an innovation in the analysis of sEMG signal.

study, there was a limited number of males in the required age group for which ethical approval had been received. The sample size recruited for the main study was therefore 20 participants.

4.2.3 Eligibility and recruitment

Table 10 provides an overview of the inclusion and exclusion criteria for the main study. The rationalisation for these criteria is outlined below the table.

4.2.3.1 Inclusion and exclusion criteria

Table 10: Inclusion and exclusion criteria for study participants

Inclusion criteria	Exclusion criteria
Male Age 20-40 years	Female
Able to understand written information	Poor understanding of English
Willing to participate and able to freely give consent	Having treatment for osteoporosis
Consent to GP being informed	Recent abdominal or pelvic surgery
BMI<30	Previous lumbar spine surgery
No history of low back pain that prevented normal activity for at least one day in the previous year	BMI >30
	Any medical radiation exposure in the past 2 years with a dose greater than 8mSv (defined as CT scan of Chest, Abdomen or Pelvis or Interventional procedures under radiological control i.e. angiography)
	Current involvement in any other current QF study

4.2.3.2 Rationalisation of the inclusion and exclusion criteria

- The reasons behind the decision to only recruit males aged between 20 and 40 were firstly because males typically have less variation in soft tissue thickness (STT) than females. Generally the greater the thickness of subcutaneous tissue between the electrode and the contracting muscle, the lower the recorded electromyographic activity (Hemingway et al. 1995; Kuiken et al. 2003; Nordander et al. 2003). A proportion of the variance in EMG measures can therefore be explained by variation in subcutaneous tissue thickness, and therefore the use of a male sample reduces this effect. Secondly, people in this age bracket were considered less likely to have any form

of spinal degeneration, which has been shown to influence kinematic variables such as IV-RoM (Deitz 2011). It has also been shown that range of motion can be affected by age and gender, with larger ranges typically observed for sagittal flexion in young adult males (McGregor et al. 1995).

- An ability to understand the written information is a pre-requisite for informed consent, as each participant should fully understand the procedures and risks involved.
- Permission to inform the participant's GP is a recommendation of the National Research Ethic Service (NRES), and would only be done in the event of an adverse reaction to any of the study procedures or as a result of an incidental finding needing onward referral.
- The requirement for participants to have a BMI of less than 30 was put in place for several reasons. Firstly, in terms of the fluoroscopic image quality, generally the higher the BMI beyond the normal range, the greater the chance of degradation of the digital image. It is also likely that a higher radiation dose would be required to produce the image in the first instance (Mellor et al. 2014). Secondly, in terms of minimising the impact of confounding variables such as STT, it was thought that a BMI of less than 30 would increase the likelihood of recruiting participants with comparable subcutaneous tissue thickness, as for the selection of male participants only.
- The final inclusion requirement relates to the fact that the study is investigating so called 'normal' spinal biomechanics. A study investigating the prevalence of LBP in adults in the UK suggests that sufferers should be included if they had LBP over the previous 12 months (Mason 1994). It was taken from this that individuals that had no activity limiting LBP over the previous year could therefore be considered as healthy participants (i.e. 'normal').
- The exclusion criteria were designed to protect participants and to prevent the collection of potentially poor quality data. Although the radiation dose from a QF protocol is relatively small (Table 11, section 5.1.4), those that have taken part in other studies or been exposed to medical radiation over the last 2 years were excluded to avoid a cumulative effect. Images taken of participants with osteoporotic spines are more likely to be of sufficiently poor image quality that template tracking is not possible (Section 4.2.6), and those that have undergone previous lumbar/pelvis surgery were considered likely to have what could be considered as abnormal spinal biomechanics.

4.2.4 Recruitment

Participants were recruited from the male student population at the Anglo-European College of Chiropractic (AECC), and from visitors to the college. In order to advertise for volunteers, the author talked to students during lectures, sent e-mails to student cohorts groups and discussed the study directly with individuals working in the AECC teaching clinic.

4.2.5 Data collection

4.2.5.1 Study environment

All examinations were conducted at the same time of the day⁸ (in the mornings between 9.00am and 11.00am), at the AECC clinic x-ray room. The room temperature was set at 19°C, and all electrical equipment in the room was earthed.

4.2.6 The quantitative fluoroscopy technology

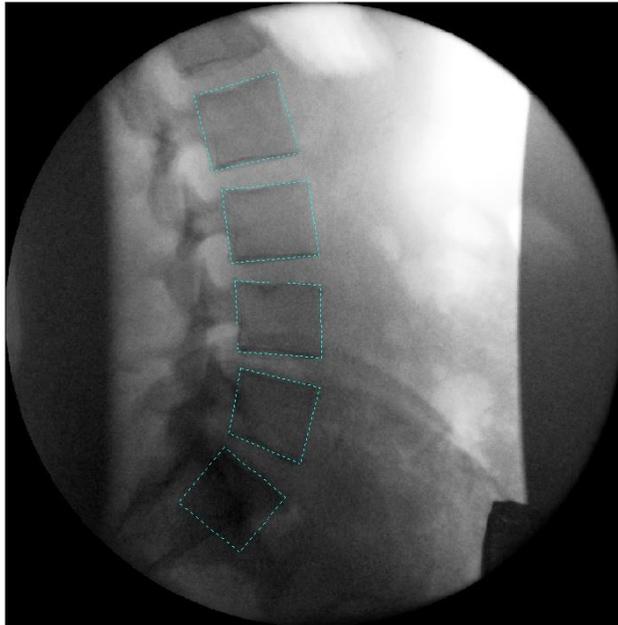
QF is a system that uses commercially available fluoroscopy imaging devices in order to measure continuous inter-vertebral movements in both the lumbar and cervical spines. The recumbent protocol uses a lying motion table (Figure 38), whilst the weight-bearing protocol uses a standing motion frame (Figure 39). These motion devices assist (guide in the case of the standing motion frame) study participants in the performance of standardised (pre-determined rate and range) spinal movements. The fluoroscope collects image data that are subsequently evaluated using image processing software that is able to identify each vertebra, and track it during the spinal movements. This QF technique has been previously validated (Breen et al. 2006).

In order to process the digital images from the fluoroscope, the software program obtains geometric and positional data for each vertebral body as it appears in image sequences of the lumbar or cervical spines. Each fluoroscopic sequence can include hundreds of individual digital images, effectively providing a continuous x-ray image. Participants are imaged whilst they perform controlled, standardised spinal movements (such as forward flexion and the return to upright). To enable the software to track the vertebrae during movement, only low resolution images are required. This allows for a relatively low dose imaging protocol compared to standard x-ray sequences. All images produced are anonymised and transferred to a secure computer for further processing.

⁸ Diurnal variations can influence the stress on the lumbar spine. It has been shown that creep loading of the disc throughout the day may gradually decrease the spines resistance to bending (Adams et al. 1987). Their results showed that the range of lumbar spinal motion (using electronic inclinometers) increased on average by 5 degrees over the course of the day. The time of day when the protocol takes place therefore needs to remain consistent.

Operators who have received adequate training in the procedure, place templates around each of the vertebrae from L2-S1 on the first image of the x-ray video sequence, a process referred to as marking up (Figure 35). Throughout each subsequent frame of the movement cycle the computer software registers the x and y co-ordinates of each vertebral body, therefore continually tracking each vertebra's position during the movement.

Figure 35: Templates placed around the lumbar vertebrae (L2-S1) on the first frame of the x-ray video sequence

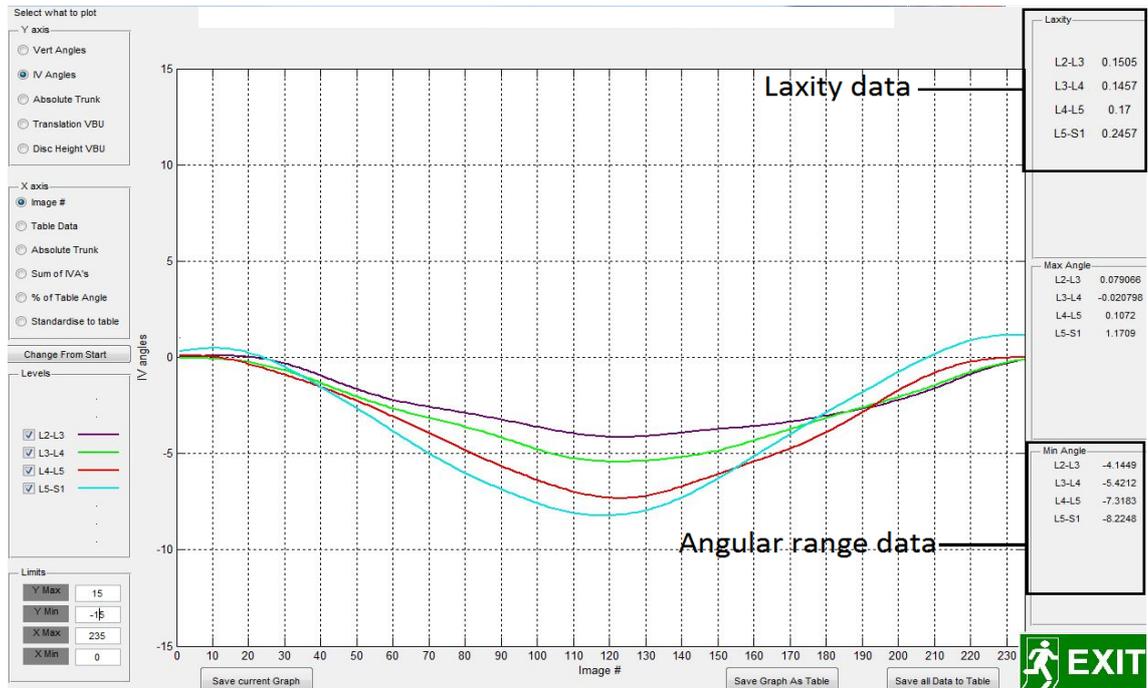


In order to reduce template positioning (marking-up) error, and to increase reliability, the vertebrae from the first digital image have their templates marked-up 5 times. When the initial template mark-ups are completed, they are also checked visually to ensure that they track the true position of each vertebra throughout the movement cycle. If a template does not track, then it is discarded. If the situation arises that all 5 mark-ups fail to track, then the process is started from scratch. If it is not possible to get the vertebra to track, then the data for that participant is discarded. This is usually an image quality issue, and beyond the control of the operator. The causes of such issues can vary, but include, poor bone quality, bony superimposition and excessive bowel gas.

The output from this bony movement analysis is displayed as a plot of inter-vertebral motion (Figure 36). The figure shows an example of the inter-vertebral rotation plot produced when a participant performs a flexion and return to standing weight-bearing sagittal protocol. From these data it is possible to retrieve the maximum inter-vertebral range of motion (IV-RoMmax), and initial attainment rate for each level, data that cannot be measured from static, end of range

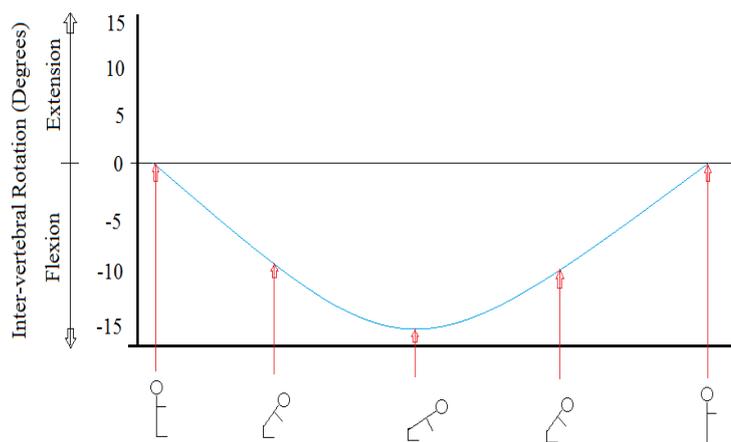
radiographs. Figure 37 is a simplification of an inter-vertebral rotation plot, and highlights relative phases of bending in relation to the plot, throughout the flexion and return sequence.

Figure 36: Weight-bearing sagittal plane motion graph produced by the bespoke software package from flexion and return movement data



Note: Laxity data = initial attainment rate data

Figure 37: A simplification of the graph shown in figure 36 (representing a single inter-vertebral level) highlighting the stages of forward bending that correlate to different sections of the plot



Participants in the normative database study (Section 4.2.1) are allocated to either a coronal or sagittal investigation group. They then participate in both a recumbent and weight-bearing QF

protocol conducted in that plane of motion. The participant group for this study consists of 20 of those who were allocated to the sagittal plane of motion, and who also consented to participate in the additional sEMG investigations.

4.2.7 QF equipment

The QF data were collected using the Siemens Arcadis Avantic VC10A (CE0123) (Figures 38 and 39). The motion tables are manufactured by Atlas Clinical Ltd (declared conformity under MDD93/42/EEC). The digital image analysis software is a bespoke program written to work with 'Matlab' (the Mathworks, Cambridge). This software has been developed and refined by the IMRCI in collaboration with a company called Orthokinematics Inc., who are commercial partners with the AECC.

4.2.8 The QF image acquisition protocol

Prior to any data collection, informed consent was collected by the author. As previously described, the participants in this study were a sub-group of those who were also participating in the normative database study (Section 4.2.1). As such, all participants performed a recumbent sagittal protocol prior to the application of any sEMG equipment and the subsequent weight-bearing sequence. This allowed the opportunity to improve sEMG electrode positioning accuracy, with the application of a bony landmark reference electrode that could be seen in the recumbent images (Figure 42). The reference electrode was placed when the participant first assumed the recumbent imaging position as described below.

For the recumbent sequence, participants were asked to lie on their right hand side, with their knees slightly bent, in order to flatten the lumbar lordosis (helping to avoid vertebral endplate overlap in the images). The bony landmark reference electrode was then applied to the spinous process of the third lumbar vertebra, and lead shielding was placed over the participant. Prior to any irradiation, participants were taken through the full range of motion (40° for the recumbent sequence) in 10° increments. This process assured that the participant could tolerate the overall range, and familiarised them to the required movements (It has been demonstrated that even pre-surgical low back pain patients can usually tolerate this degree of motion (Breen 2006)). When participants confirmed that they were able to tolerate the movement range and rate, the fluoroscopic imaging commenced with exposure beginning simultaneously with the onset of the table movement.

Figure 38: QF acquisition during a recumbent sagittal plane flexion and return protocol



Between the recumbent and weight-bearing sequences, the sEMG equipment application procedure was performed. In order to avoid any ambiguity, this procedure is outlined in a separate section to follow (Section 4.2.14). When the recumbent series and the electrode application process were completed, participants were asked to stand with their right sides next to the upright motion frame (Figure 39), in preparation for the weight-bearing sequence. As for the recumbent protocol the appropriate protection was applied in the form of a lead apron and a thyroid shield, and a belt was placed around the waist and an appendage of the motion frame, in order to stabilise the pelvis. The participants were then guided through the full range of motion (60° for the weight-bearing sequence) but this time in 20° increments. The participants also received additional instructions to follow the arm rest as a guide (not to rest on it), and to keep their body and head as straight as possible throughout the imaging cycle. When the participants were ready, they followed the motion table through 60° of forward flexion and return to the standing position during continuous fluoroscopic imaging.

Figure 39: QF acquisition during a weight-bearing sagittal plane flexion and return protocol

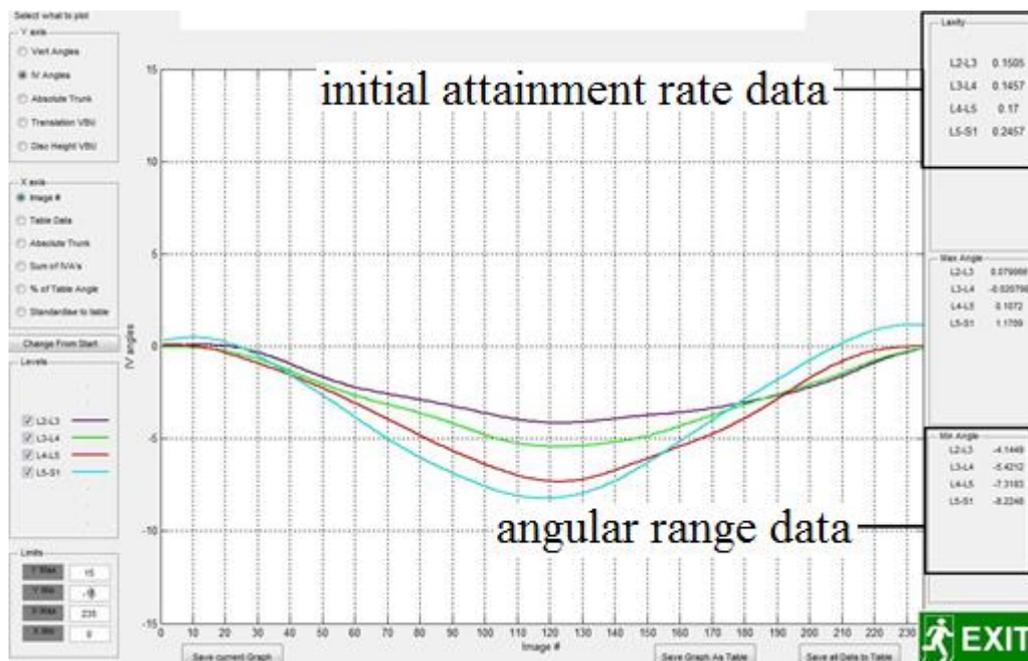


The motion of both the recumbent and weight-bearing motion frames was also concurrently recorded by electronic feedback from their motor drives. This provides global movement information which can be plotted against the inter-vertebral motion using the bespoke Matlab software.

4.2.9 Analysis of QF data

A screenshot example of the graphical user interface (GUI) from which the data values are read can be seen in (Figure 40). It should be noted that angular range values for inter-vertebral flexion in the sagittal plane are not taken as being negative. The computer program has to distinguish between angular ranges during both flexion and extension, with flexion values typically appearing as negative and extension values as positive. If an intervertebral level demonstrates paradoxical motion however, this situation can be reversed.

Figure 40: Graphical user interface (GUI) from which angular range and initial attainment rate values are taken



4.2.10 Radiation exposure

The most recent available data (Mellor et al. 2014) reports the mean exposure dose for a sagittal QF procedure as 0.24 mSv (SD 0.529). That is equivalent to approximately 11 weeks' background radiation and compares very favourably to a standard lumbar radiograph investigation (2 views) of 2.21mSV. The risk of inducing cancer from 1mSV is estimated to be 1:20,000 which when considered in terms of a lifetime risk of developing cancer as high as 40% in the UK (Sasieni et al. 2011), puts the risk of a QF examination into perspective.

4.2.11 Incidental findings

The digital images produced were screened for incidental findings by the author, a qualified chiropractor who is trained to interpret x-ray images. Due to the relatively poor image quality and as a backup measure, any findings that were a possible cause for concern were further screened by a Diplomate of the American Chiropractic Board of Radiology (DACBR) and if necessary referred onwards for medical opinion.

4.2.12 Surface electromyography

Typical sEMG systems require wire connections between the electrodes and base unit, which can be obtrusive, potentially affecting a participants movement or at the very least making them conscious of it. This study used wireless sEMG technology that enabled free participant movement, improved on the common mode rejection ratio achievable with typical wired systems, and reduced the requirement for the use of notch filters (50Hz mains interference), as the system's wireless technology was battery operated.

4.2.13 The sEMG equipment

The sEMG signal data were recorded using three pairs of BioNomadix Dual-Channel Wireless EMG Transmitters and Receivers (Biopac Systems, Inc., California, The United States of America). The sampling rate is 2000Hz. The unit has a CMRR of 110dB and an input impedance of 1GOhm.

The 6 signals were band pass filtered 10-500Hz and full wave rectified. The root mean square (RMS) was calculated for individual participant cycles and normalised to a sub-maximal voluntary contraction (sMVC) to be expressed as a % of sMVC. All signal processing and analysis was conducted using Acqknowledge software (version 4.2).

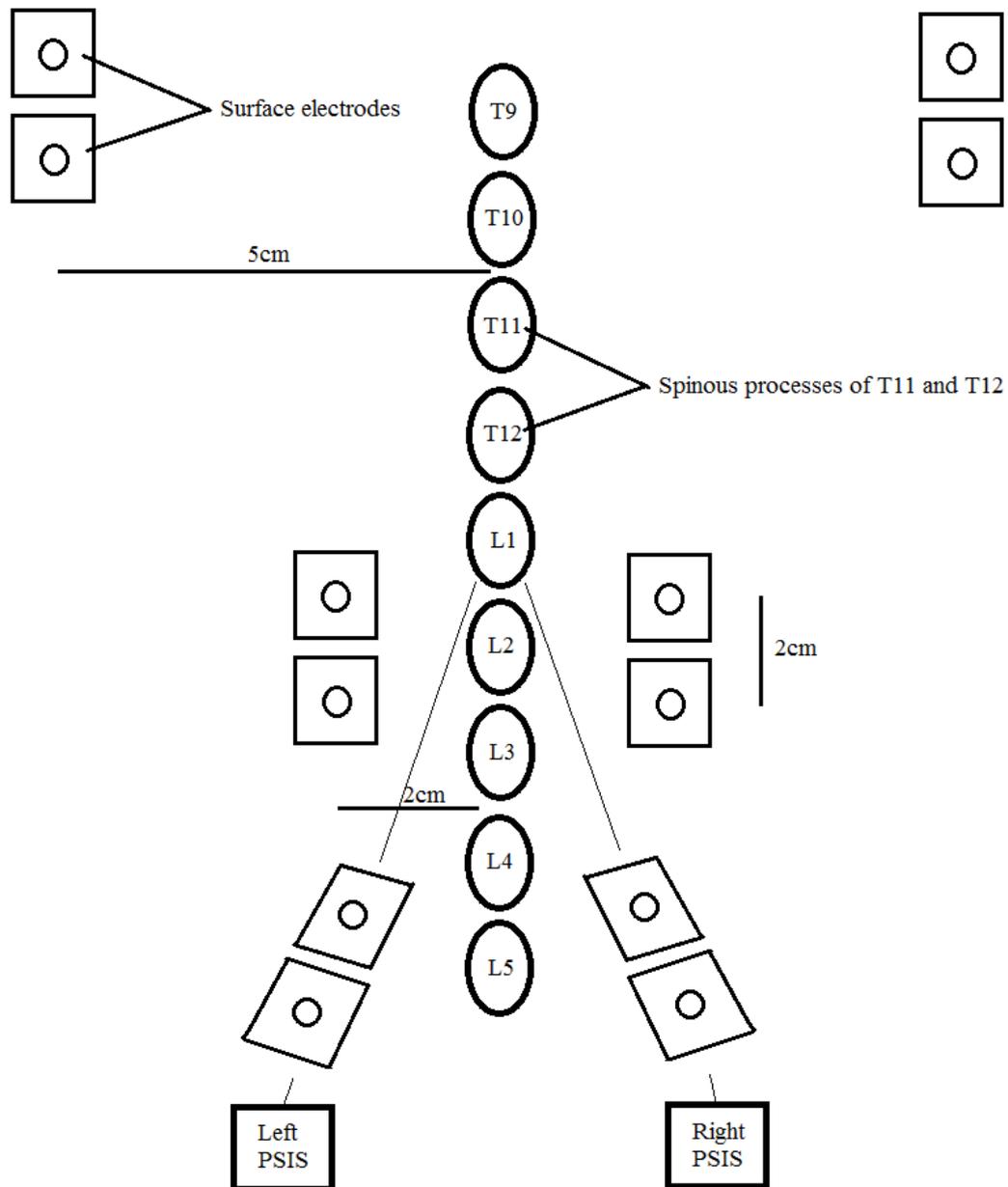
4.2.14 The sEMG recording protocol

In order to investigate inter-vertebral kinematics and paraspinal muscle activity concurrently, the QF protocol (Section 4.2.8) incorporated the following sEMG recording protocol.

When the recumbent QF sequences were completed, the participants were asked to lie prone on a bench, with a pillow under their waist to invoke slight flexion. They then had the skin over their lower backs prepared for sEMG electrode application by light abrasion, cleaning with an alcohol swab and when necessary shaving of the area. The next step was to mark 15 electrode sites on their backs marked with a skin pencil (6 electrodes bilaterally and 3 reference electrodes). Disposable pre-gelled self-adhesive electrodes Ag-AgCl (Silver / Silver Chloride) were then applied over 3 bilateral muscle groups with a 2cm centre to centre inter-electrode distance: TES (vertically 5cm lateral to the T9 spinous process), LES (vertically 2cm lateral to the

L2 spinous process), and LMU (aligned between the posterior superior iliac spine and the L1-L2 interspace, 2 cm lateral to the spinous process of L5) (Figure 41).

Figure 41: The bilateral sEMG electrode placements for TES, LES and LMU (Posterior view of thoracic and lumbar spinous processes)



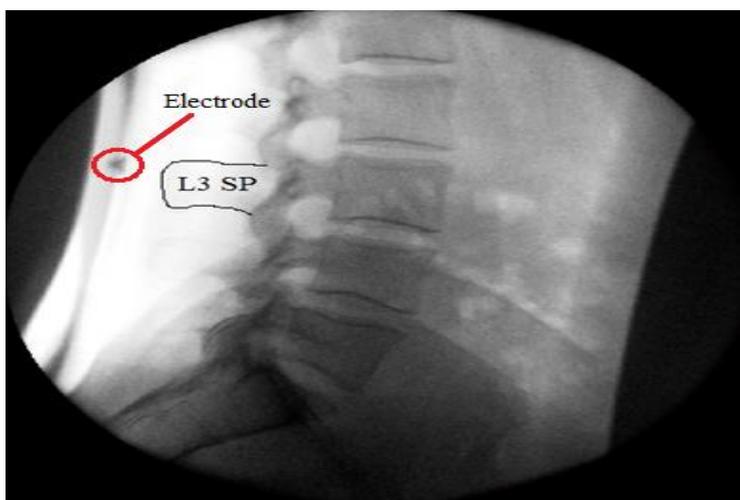
Although cross talk from multiple muscles will inevitably contribute to the signal recorded at each electrode site, cross-sections of the spine at each electrode site show that the muscle that will predominate at T9 and L2 is longissimus thoracis, and at L5 superficial multifidus (Appendix H). Three Biopac wireless transmitters (Bionomadix Dual Channel Wireless EMG) were then also placed on the lower back attached by self-adhesive Velcro pads. In order to test that all the electrode sites were recording and that signal quality was adequate, the participants were then

required to perform a single bend into forward flexion within their own comfort level. If all 6 channels were recording sufficiently clean signals, then the weight-bearing stage of the protocol could begin. The participants stand with their right side against the upright motion frame, with their forearms placed on the arm rest appendage (Figure 39). They then proceeded as normal with the weight-bearing sagittal plane image acquisition and sEMG signal data was synchronised with the beginning of the motion table movement using a simple electrical switch (Section 4.2.20.2). Participants have access to an emergency stop button, which halts the motion frame at any time if they so wish.

4.2.15 Electrode application accuracy

Electrode application accuracy is dependent on the subjective identification of bony anatomical landmarks. It has been shown that application techniques, are therefore limited by human error and variations in individual's anatomy (Kim et al. 2007, Chin et al. 2006, Billis et al. 2003, Chakraverty et al. 2007). It has been suggested however that accuracy can be improved significantly when techniques are combined (Merz et al. 2013). As this investigation was combined with a normative database study (Section 4.2.1), recumbent QF imaging was conducted before the weight-bearing imaging commenced. Therefore in order to improve electrode positioning accuracy, an electrode was placed over the spinous process of L3 during the recumbent protocol (Figure 42). This allowed the comparison between the true position of L3 spinous process and the position of L3 spinous process based on the use of the iliac crests as a bony landmark reference point. If there was disagreement, then the perceived location of the L3 spinous process was adjusted accordingly to improve electrode application accuracy.

Figure 42: An electrode placed over the spinous process of L3

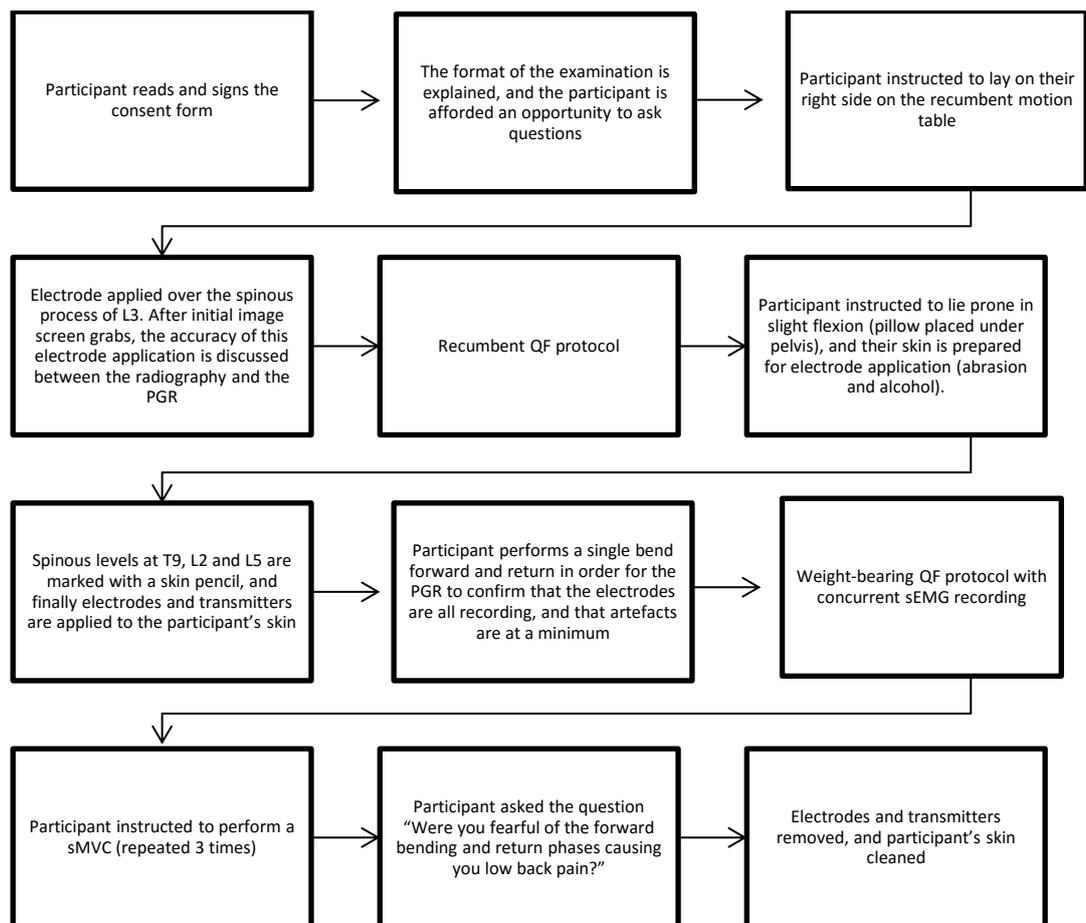


4.2.16 Reference contraction for the purpose of normalisation

In order to provide a reference contraction (a sMVC), when all image acquisitions were completed, participants were asked to lie prone on a padded bench with their hands behind their head. They were then instructed to raise their upper body off the couch and hold this position for 5 seconds whilst their legs and pelvis were supported (in a modification of the Sorensen test (Demoulin et al. 2006)). The participants were asked to repeat this procedure 3 times. Lastly each participant was asked a single question related to any fear of pain they may have had, prior or during either the weight-bearing forward bending and return sequences or the reference contraction. The electrodes were then removed, and the participant's skin was cleaned.

The combined recumbent, weight-bearing and sEMG aspects of the study protocol are outlined in (Figure 43).

Figure 43: Flow chart outlining the key stages of the study protocol



Due to the complex nature of the protocol and the multiple considerations involved, a safety check list was devised to be completed at every examination. The check list can be found in the appendices (Appendix G).

4.2.17 Checking for signal contamination

As described above as part of the sEMG protocol, after each participant had performed a voluntary forward bend, the signals from each of the 6 muscle sites were checked for contaminants. This quality assurance procedure involved a visual check of each signal, verification that baseline voltage readings were at an acceptable level (i.e. 3-5 μ V), and that a signal frequency analysis revealed no 50Hz mains interference.

4.2.18 Analysis of sEMG data

All data were recorded and analysed using Acqknowledge software (version 4.2). The software was used to process the signal into root mean square (RMS) sEMG, which is a representation of the mean power of the signal and is a common and preferable method of smoothing (Basmajian and De Luca 1985; Soderberg 2000). RMS EMG is based on the root mean square calculation, and is basically a process of squaring each value within the signal, generating an average, and then calculating the square root (Soderberg 2000). The RMS sEMG was then normalised to a sMVC (Section 4.2.16), and expressed as a % of the sMVC. The sEMG ratios (Reeves et al. 2006; Van Dieen 2003) were calculated from the mean left-right normalised RMS sEMG amplitudes during the flexion phase only as follows, LMU/LES, LES/TES and LMU/TES. In order to calculate sEMG changes at different stages of the flexion cycle, the forward bending phase was divided into 5 epochs for each participant (D'hooge et al. 2013). The change in mean RMS sEMG between epochs was then calculated e.g. the change during the early stage of flexion was calculated as (epoch 1 – epoch 2) for each TES, LES and LMU. This was repeated to determine changes between all epochs at all levels.

The standardised rate and range of the weight-bearing protocol motion frame guided each participant to 60° of flexion in approximately 10 seconds. This rate can change fractionally between participants, and so epoch lengths were calculated on an individual basis but were typically of 2 seconds duration.

4.2.19 Risks associated with sEMG

There are no significant risks associated with sEMG. Participants may however experience minor discomfort as a result of skin preparation prior to electrode attachment, or as a result of electrode removal, either of which could possibly result in transient minor red marks on the skin surface. There was also a very slight risk of allergy or irritation caused by the adhesive on the electrodes, although non-allergenic gels were used to minimise this risk.

4.2.20 Synchronisation of technologies

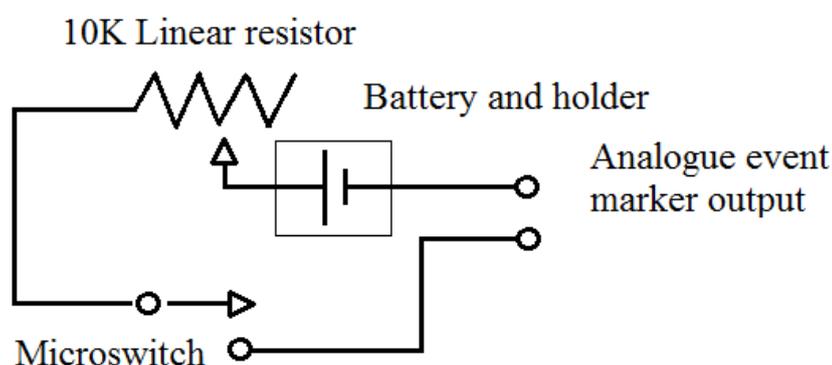
4.2.20.1 Synchronisation of QF and the onset of motion frame movement

When the QF imaging began, a light on top of the fluoroscope turned on. When the motion table operator saw this light, they initiated the movement of the motion table. This meant that there was a fractional difference between the onset of the motion table and the QF recording; however it did ensure that the recording had begun before any motion table movement. The bespoke software allowed the template marker to observe the first frame of trunk movement and correlate this image with the onset of table motion.

4.2.20.2 Synchronisation of sEMG and motion frame movement

In the sEMG amplitude agreement and reliability studies (Chapter 5), synchronisation between the table motion and sEMG recordings was achieved by simultaneously pushing the motion table start button, and a button that produced a marker within the sEMG data. This technique was not sufficiently accurate for use in the main study, and so a system incorporating a microswitch was developed (Figure 44). The switch triggered at the exact moment the motion table began its movement, producing an on/off trace in the sEMG data using a separate channel to the recorded signals.

Figure 44: Electronic diagram of the microswitch circuitry built for the synchronisation of the sEMG software and the motion table



4.2.21 Ethics

Ethical approval for the preliminary sEMG studies (with no requirement for radiation) was granted by the AECC Research Ethics Sub-Committee (Appendix E). Participants in the main study however are exposed to potentially harmful ionising radiation, and as such, ethical approval was required from the National Research Ethics Service (NRES). In order to avoid irradiating additional participants in a stand-alone study, it was decided that the investigation

could be incorporated into a pre-existing study through a 'substantial amendment' to its own ethical approval. The substantial amendment was approved (Appendix E).

4.2.22 Public and patient involvement (PPI)

The participation of members of the public in health research is central to the research policy of the UK (Boote et al. 2010). It is defined by its national advisory group INVOLVE as "doing research with or by the public, rather than to, about or for the public". The last 20 years has seen publication patterns demonstrate an increased use of PPI by researchers (Boote et al. 2015), and a systematic review of PPI in research recommends that studies should include in sufficient detail the process of involvement, and how it affected the study (Brett et al. 2010). A PPI group was therefore created from former/current patients of the AECC teaching clinic. The group met on two occasions with the author and 1st supervisor⁹. The first meeting outlined the purpose and role of a PPI group and established their willingness to participate. The second meeting was a live demonstration of the study protocol (excluding x-ray exposures), followed by an opportunity for the group to comment on any aspect of the protocol, participant information sheet or concerns that had arisen. The feedback was very useful, and amendments to the study protocol were made as a result. A summary of the main outcomes of the meetings can be found in (Appendix F).

4.2.23 Summary

The methodology that has been outlined represents the first attempt to combine QF and sEMG technologies in order to investigate the biomechanics of the human lumbar spine *in vivo*. Data that were collected in the main studies (Chapters 6 and 7) used the described methodology. Any deviations from this methodology (such as in the reliability and agreement studies described in Chapter 5) are documented accordingly.

⁹ Professor Alan Breen

Chapter 5: Accuracy, agreement and reliability

5.1 Part one: Intra- and inter-marker agreement and reliability of IV-RoMmax and initial attainment rate measurements in the sagittal plane QF protocol.

5.1.1 Introduction

When spinal kinematic variables such as IV-RoM are measured repeatedly under standardised conditions, differences between the measurements will typically be found, due to natural biological variation in individual participants, errors in the measurement process, or both (Mieritz et al. 2012). When possible to do so therefore, the accuracy, agreement and reliability of study variables should be determined before their inclusion for use. In terms of spinal motion measurements, there are numerous techniques available, each with their own advantages and limitations (Section 2.5). The spine is a relatively inaccessible area of the body, which means that manual assessment techniques can be crude and are typically influenced by their subjective nature. A review of the reliability of manual evaluations of inter-vertebral motion (posteroanterior pressure over spinal segments) concluded that they were unreliable, and that a participant's pain response was a more useful indicator of inter-vertebral mobility (Lee 1995).

A solution to this problem has been the development of instrumentation to objectively quantify segmental movements. Non-invasive skin surface devices have been created, and have been shown to be generally reliable for measuring spinal movements (Essendrop et al. 2002; Mannion 1999a); however there will always be inaccuracies in these methods due to landmark identification problems, skin movements over vertebrae, and the attachment of such devices to the skin (Mannion et al. 2004). These methods also typically assess regional ranges of motion and not localised inter-segmental movements (Mieritz et al. 2012). In a study comparing QF measurement to digitisation of X-rays at maximum voluntary bending angles (MVBA), and measurement of X-rays at MVBA by ruler and protractor, Breen et al. (2012) reported 'substantially' larger errors in the latter 2 methods (Breen A.C. et al. 2012). Despite the radiation considerations therefore, QF techniques have been shown to be the most precise when assessing spinal ranges of motion, and recent QF studies have demonstrated acceptable reliability (Ahmadi et al. 2009; Branney and Breen 2014; Mellor F.E. et al. 2014; Teyhen 2005; Yeager et al. 2014). The agreement and reliability of sagittal plane weight-bearing measurements using the IMRCI QF protocol however, is yet to be investigated. Template

marking is a learned skill, and as such the author's capacity to register templates on weight-bearing images to an acceptable standard also requires assessment.

The following section reviews the accuracy of IV-RoM measurements using QF when compared to other reference standards, and investigates the agreement and reliability of both IV-RoMmax and initial attainment rate variables when measured using the QF weight-bearing protocol (Section 4.2.8). Acceptable agreement and reliability of these measurements are important prerequisites to their inclusion in the main study. As the template marking and editing processes are the most likely sources of error, the studies are referred to as intra- and inter-marker (as opposed to intra-observer/examiner).

5.1.2 Accuracy

The accuracy of QF measurements of IV-RoMmax have been investigated previously, and have been shown to be accurate to less than one degree in both the lumbar and cervical spines (Branney and Breen 2014; Breen et al. 2006). In both studies, the accuracy of IV-RoM measurements was determined using calibration models consisting of a pair of dry vertebrae, fitted with protractors and joined by an inter-vertebral universal joint that allowed for rotation in the sagittal plane. In the lumbar spine, Breen et al. (2006) calculated the root mean squares of difference between the reference and computed inter-vertebral angles through 7 settings from -10° to $+20^{\circ}$ during sagittal flexion-extension, in both optimal (x-ray beam centred on the universal joint) and degraded (model axially rotated 10° out of plane, and beam inclined inferiorly 10°) conditions (Breen et al. 2006). The RMS error was 0.52° and 1.03° for optimal and degraded conditions respectively (Breen et al. 2006). Using a similar protocol in the cervical spine, Branney (2014) showed accuracy of 0.21° (optimal) and 0.50° (degraded) during flexion (Branney and Breen 2014), indicating comparatively reduced error in cervical segment measurements, and reflecting a similar doubling in error under degraded conditions. This underlines the importance of patient positioning protocols, and the skill of the radiographer.

In a more recent QF accuracy study of lumbar IV-RoM measurements, Breen et al. refer to a 2011 FDA study (Orthokinematics 2011) which reports an error of less than 0.70° (Breen et al. 2012). Validation of a different QF system, the Dual Fluoroscopy Imaging System (DFIS) evaluated accuracy by comparing vertebral orientations using DFIS to the RSA method (a beads position matching technique that the author refers to as the gold standard) (Wu et al. 2014). This study found sagittal rotational accuracy to be within 0.63° from 5 tested positions along a flexion-extension path, mirroring the findings of the FDA study. There was no testing of out of plane accuracy however, and so it is likely that accuracy would reduce *in vivo*.

These modern QF based techniques, including QF, appear to represent an improvement over traditional methods. Pearcy and Whittle undertook a validation study of biplane radiography, and reported accuracy of $< 1.5^\circ$ (Pearcy 1982). It has therefore been shown that the QF sagittal lumbar protocol demonstrates acceptable accuracy for use in the main study.

5.1.3 Agreement and reliability

As discussed, semi-automated techniques such as those employed in this study, have been shown to improve the accuracy of measurements. As another example, Pearson et al compared the agreement and reliability of a manual digitized with a semi-automated technique and demonstrated greater precision in the latter (Pearson et al. 2011). The QF protocol is however, still subject to tracking failures and human error, particularly with regard to the manual vertebral template marking process. Barlett and Frost (2008) state that the objective of an agreement and reliability study is to quantify the reliability and agreement of the measurement, thus determining the appropriateness of its use (Barlett and Frost 2008). If agreement and reliability are poor, then any conclusions based on such measurements would be open to questioning.

De Vet (2006) reviewed the parameters for quantifying agreement and reliability (De Vet 2006). They concluded that the standard error of measurement (SEM) is the most frequently used parameter of agreement, whilst intraclass correlation coefficients (ICC's) are most appropriate for repeated measurements on a continuous scale (De Vet 2006). The criteria used to report the reliability and agreement findings in this study, are based on the guidelines produced by Kottner et al. 2011 (Kottner J. et al. 2011). These guidelines were produced with the aim of improving the quality of reliability and agreement reports, and so outline the information that should be included in order for the results to be correctly interpreted.

The reliability and agreement of several QF inter-vertebral motion measurement techniques have already been investigated (Ahmadi et al. 2009; Mellor F.E. et al. 2014; Teyhen 2005; Yeager et al. 2014). However, gaps remain in the literature in terms of the inter-vertebral levels included in each study, and the focus on recumbent or weight-bearing images. The methodology designed for the main study data collection, requires acceptable agreement and reliability of measurements from L2-S1 during weight-bearing flexion in the sagittal plane. Mellor (2014) investigated the agreement and reliability of a recumbent QF protocol measuring IV-RoMmax from L2-L5, but excluded L5-S1 due to anticipated difficulties with image registration, as a consequence of superimposition of the iliac crests at this level (Mellor F.E. et al. 2014). In terms of weight-bearing, Teyhen et al. focussed on the levels of L3-S1, excluding inter-vertebral levels

of the upper lumbar spine (Teyhen 2005), and their methodology did not incorporate continuous data (using only upright and full flexion images). Most recently Yeager et al. assessed the whole lumbar spine (L1-S1), but reported reliability of lumbar measurements that combined vertebral levels, and did not analyse them individually (Yeager et al. 2014). All of these studies have demonstrated acceptable measurements of IV-RoMmax using QF technologies (i.e. ICC's > 0.9). However, knowledge of inter-segmental measurement agreement and reliability at all required levels, during sagittal weight-bearing, whilst incorporating continuous data, remains incomplete.

5.1.4 Why is it necessary to investigate agreement and reliability of measurements using both the recumbent and weight-bearing QF protocols?

It is intuitive that the better the quality of the images, the more accurate and consistent measurements made using them will be, and in a study assessing the effect of roentgenogram quality on the accuracy and consistency of sagittal plane measurements, Shaffer et al. (1990) concluded that higher quality images are more accurately evaluated than those of lower quality (Shaffer 1990). In a more recent study involving digital radiograph measurements, Aubin et al. (2011) showed that improved image quality also positively influences inter-observer reliability (Aubin et al. 2011).

In terms of weight-bearing versus recumbent images, generally recumbent images will be of better quality. There are changes in soft tissue locations between weight-bearing (standing) and recumbent (lying on the participant's side) positions, which alter the behaviour of x-ray beams as they pass through the abdomen. When a participant is weight-bearing, the soft tissues around the lumbar spine are brought together due to gravity. When the lower abdomen bulges under gravity, there is a crowding together of organs, which consequently causes an increase in the density of the soft tissues. This means that the attenuation of x-radiation is greater in weight-bearing images of the lumbar spine. This increase in 'Compton scatter' therefore results in degraded image quality. When participants are recumbent, the organs disperse and reduce the thickness of the tissues the x-rays are interacting with, resulting in comparatively superior image quality.

The radiation factors from the main study show this to be the case for kV, but surprisingly show that comparatively less mA were required in weight-bearing (Table 11). It was anticipated that more x-rays (higher mA) would be required during weight-bearing, because there is more of the abdomen to travel through. This was not the case for the study sample, which may be a limitation of its size. There was evidence to suggest however that the x-rays needed to travel faster (increased kV) in order to pass through the abdomen during weight-bearing. The

likelihood of ‘degraded’ weight-bearing images relative to those produced using a recumbent protocol, may make image template marking more difficult, and harder for the computer software to track. This is an important consideration, and highlights the need for an agreement and reliability study of measurements during the QF weight-bearing protocol, including the inter-vertebral level L5-S1.

Table 11: Means of radiation factors (kV and mA) in recumbent and weight-bearing QF flexion and extension sequences combined

Recumbent kV	Weight-bearing kV	Recumbent mA	Weight-bearing mA
78.6	81.4	56.2	54.3

(Data taken from the main study n=18)

An additional reason for the investigation of weight-bearing measurement agreement and reliability is the relative increase in variation seen in inter-vertebral motion patterns in comparison to recumbent. During weight-bearing, there is an increased occurrence of phenomena such as double peaks, and paradoxical motion seen in the motion graphs (Appendix M). It is anticipated therefore, that a higher incidence of such trends, may increase the likelihood of template tracking errors, subsequently affecting agreement and reliability.

Objective:

1. To determine the inter- and intra-marker agreement and reliability of IV-RoMmax and initial attainment rate measurements during weight-bearing sagittal flexion and return, using the QF protocol.

5.1.5 Methods

For both the intra- and inter-marker studies, motion sequences recorded from an ongoing normative database study (Section 4.2.1) were selected retrospectively for the vertebral template marking-up procedure. The intra-marker study used the images from 10 healthy male participants (mean age 22-29 SD 2.3) recorded using sagittal plane QF protocols. Images were processed by a single marker (the author) and repeated 6 weeks later. The inter-marker study used images from 10 separate healthy participants from the normative study (mean age 25-66 SD 14.6), but were processed by two independent markers. The first marker was a medical physicist, and the second was the author. The markers were blinded to the others’ results, and had 3 and 1 year(s) of template marking experience respectively.

A typical sagittal plane motion sequence produces over three hundred images (acquisition rate of 15Hz over approximately 20 seconds). The lumbar vertebrae from levels L2 to S1 were marked-up manually on the first image from this sequence. This creates a template around each vertebra, and the process was repeated 5 times. A computerised tracking algorithm then identifies the position of each template for all images, continuously tracking each segment's movement throughout the flexion and return phases (Section 4.2.8). The IV-RoMmax and initial attainment rate data were then extracted from the output and analysed. Data from all 20 sequences went forward for analysis, as they all satisfied the study's image quality control standards.

5.1.6 Data analysis

5.1.6.1 Rationale statistical method

There are many different statistical methods that can be used when analysing intra- and inter-rater reliability and agreement. Commonly, reliability analysis will include Kappa statistics or intraclass correlation coefficients. Agreement measures may include proportions of agreement, standard errors of measurement, coefficients of variation or limits of agreement (Bland-Altman method). The decision to use one statistic over another is based on assumptions regarding the treatment of random and systematic error, the sampling, and the type of data i.e. nominal, ordinal or continuous (Knottner J. et al. 2011). In the literature, the type of statistic used is often incorrect and sometimes unreported. In a study investigating the reliability of lumbar range of motion measurements using an inclinometer, Mayer et al. (2004) reported excellent intra- and inter-rater reliability using the Pearson's r correlation (Mayer 2004). Although their results showed r values >0.95 , the Pearson's r does not account for systematic intra- or inter-rater bias, and so an intraclass correlation coefficient may have been more suitable. The ICC takes into account such sources of error and relates it to the variability between participants. If the measurement error is small relative to the participant variation, then the ICC will approach 1, however if the error is relatively large, then the ICC will be smaller (De Vet 2006). This also means that the ICC as a parameter can be heavily influenced by the heterogeneity of the participant population.

As discussed previously, the measurements made in this study are on a continuous scale, and are therefore most suitable for analysis with ICC's (De Vet 2006). If the data were either nominal or ordinal however, then a Kappa coefficient would have been more appropriate (Sim 2005). Kappa statistics can provide valuable information about categorical data, but like the ICC, many different types are available, and so careful selection is required to avoid the misinterpretation of results (Knottner et al. 2011). The limits of agreement were proposed by Bland and Altman,

and were designed to assess the agreement between two separate measurement methods (Bland 1986). The calculation requires the mean difference between two measurements, and the standard deviation of these differences. It is suggested that approximately 95% of these differences will lie between the mean differences ± 1.96 standard deviations (Bland 1986). Costa-Santos et al. (2012) compared the interpretation of limits of agreement to the ICC when analysing neonatal outcome variables (Costa-Santos et al. 2011). They found that the two statistics provided inconsistent results, and therefore recommend that each be interpreted with their limitations in mind. In a QF agreement and reliability study investigating lumbar intervertebral translation, Van Loon et al. (2012) used both limits of agreement and ICC's to analyse their results (Van Loon et al. 2012). As Costa-Santos et al. (2012) discuss, the main limitation of limits of agreement is the subjective nature of their interpretation (Costa-Santos et al. 2012), which is highlighted in this example. Van Loon refers to the limits of agreement as 'best agreement' (smallest range) and 'least agreement' (largest range), but there is no mention of the relevance of either. Limits of agreement therefore can be a useful parameter, but only when understanding of the ranges is sufficient to interpret their meaning. In conclusion however, according to Myles (2001), the limits of agreement were originally developed for two sets of independent data and are not suitable for repeated measures data (Myles 2001). They were therefore not used in the current study's analysis.

In this study the agreement and reliability of measurements will be quantified by reliability (ICC) and agreement (SEM) as recommended by Barlett and Frost (2008) (Barlett and Frost 2008). The selection process for each of these statistics is outlined below.

5.1.6.2 ICC Selection

Selecting an appropriate ICC is an important part of method design, as the type of ICC used will affect the ways in which it can be interpreted. In order to justify the ICC choices made for the study, the selection procedure has been outlined.

If the structure of the data is considered as rows and columns, the participants ($n = 10$) are the rows and the columns represent the different measurements made, for example, for the intra-marker study, 1st and 2nd mark-ups, and for the inter-marker study, 1st and 2nd markers. The row data were considered a source of systematic variance, as it is expected that there will be differences amongst the participants. In this study, the column data is also considered a source of systematic variance, as there is variability in the skill levels of the independent markers. Therefore, as there are two sources of systematic variability, a two-way ANOVA model was required.

When a two-way model is selected, another important consideration is whether the column variable represents a random or fixed effect (McGraw 1996). This decision does not affect the ICC calculation, but does alter its interpretation. For both intra- and inter-marker studies fixed column variables were chosen, as the population of available markers varied in experience and skill level, any change in marker would likely have an effect.

The difference in marker skill level also helps determine the choice between ‘consistency’ and ‘absolute agreement’ measures. If the systematic variability due to the difference in marker skill level is not considered relevant, then consistency measures are used. However if the difference in marker skill level is considered relevant, then absolute measures are most appropriate. In the case of the intra-marker study, it is the author’s belief that experience gained from the marking-up of images for the first time would influence their ability to mark-up on the second occasion. Similarly, the inter-marker study was conducted by an experienced marker, and a relative novice. Therefore in both situations, the difference in marker skill would most likely influence systematic variability, and so absolute agreement measures were considered the most appropriate.

The following formulae show the difference between absolute agreement and consistency measures.

$$ICC_{\text{agreement}} = \frac{\sigma_p^2}{\sigma_p^2 + \sigma_{pt}^2 + \sigma_{\text{residual}}^2}$$

$$ICC_{\text{consistency}} = \frac{\sigma_p^2}{\sigma_p^2 + \sigma_{\text{residual}}^2}$$

σ_p^2 denotes variance in the participants under study, σ_{pt}^2 denotes variance in the markers, and $\sigma_{\text{residual}}^2$ denotes measurement error (Ailliet et al. 2015; De Vet 2006).

When marker variability is considered relevant, it is included in the denominator of the estimated ICC (ICC agreement). If it is considered irrelevant, it is simply not included (ICC consistency).

The final stage of the selection process is to decide whether the ICC should apply to single or average measurements. In this study the data produced represents the participant’s individual

IV-RoMmax or initial attainment rate measurements, and are therefore considered single measurements. Although the computer software uses the average of 5 templates to process each individual measurement (Section 4.2.8), average measurements would only be appropriate if the results were an average taken from more than one marker.

The ICC selected for both intra- and inter-marker studies was ICC_{absolute agreement} (3, 1). In SPSS this is represented as a two-way mixed, single measure. Statistical analysis was performed using IBM SPSS (Version 21).

5.1.6.3 Agreement

The formula used to derive the SEM is calculated by dividing the standard deviation of the mean differences between the two measurements (SD_{diff}) by $\sqrt{2}$ (De Vet 2006). It is suggested by De Vet et al., that the factor $\sqrt{2}$ is included as it concerns the difference between two measurements, and that errors can occur in both. The formula used is shown below.

$$SEM = SD_{diff}/\sqrt{2}$$

5.1.7 Results

5.1.7.1 IV-RoMmax: Reliability

A total of 20 image sets were obtained from the ongoing normative database study (Section 4.2.1). Convenience samples of 10 participants were used for both the intra-marker and inter-marker studies. Two different sets were used, as initially only an intra-marker study was planned, and therefore the subsequent inter-marker study was conducted using the author as the second marker. These images were taken from participants who had their template marking done previously, and so two separate groups were required. The intra-marker group were aged 22-29 years (mean 26, SD 2.3). The inter-marker group were aged 25-66 years (mean 47, SD 14.6). There were no tracking failures in either group, and so results were produced for all levels (L2-S1) in both.

The ICC's (reliability) and SEM's (agreement) for both intra- and inter-marker IV-RoMmax studies are shown in table 12. The results show acceptable reliability with the smallest ICC being 0.93 and 0.83 for the intra- and inter-marker studies respectively. When comparing Intra- and inter-marker groups, it was expected that the intra-marker would demonstrate better reliability and agreement (Mellor F.E. et al. 2014; Yeager et al. 2014). This trend was observed in the recumbent group for all inter-vertebral levels apart from L4-L5, which showed marginally better reliability in the inter-marker group. In the weight-bearing group however, ICC's were the same

or slightly better in the inter-marker group at 2 out of the 4 inter-vertebral levels. It should be noted however that the differences between these groups is small, and that all ICC's represent excellent reliability.

When taking the 95% confidence intervals into account however, a clearer difference between intra- and inter-marker groups is seen. In both recumbent and weight-bearing orientations, the inter-marker study demonstrates a larger range (with the exception of L5-S1 weight-bearing). The recumbent group demonstrated the lowest level of reliability in terms of the confidence intervals, with a lower limit of 0.45 at L5-S1 in the inter-marker group. In an attempt to categorise the ICC, Shrout suggests that this result would be considered 'fair', however the ICC's at all other levels and groups would be either 'moderate', or predominantly 'substantial' (Shrout 1998). When comparing the ICC's of recumbent and weight-bearing groups, for both intra- and inter-marker studies, it is notable that reliability is comparatively better at all levels for weight-bearing.

5.1.7.2 IV-RoMmax: Agreement

Agreement was found to be better than 1° at all levels, for both intra- and inter-marker studies and in both recumbent and weight-bearing groups (Table 12). In the recumbent group intra- and inter-marker SEM's were very similar, but there was a more marked difference in weight-bearing, which demonstrated comparatively increased error at all inter-vertebral levels in the inter-marker study. Recumbent and weight-bearing groups demonstrated no clear differences, with the exception of L2-L3 in the inter-marker study (recumbent SEM 0.31°, weight-bearing SEM 0.76°). The smallest SEM was 0.17° at the level of L3-L4 in the recumbent intra-marker study. The largest SEM, as previously mentioned was 0.76° at the level of L2-L3 in the weight-bearing inter-marker study. This result aside, in terms of comparing inter-vertebral levels, generally L5-S1 most consistently demonstrated the greatest disagreement, ranging from an SEM of 0.54° in the intra-marker weight-bearing group, to 0.69° in inter-marker recumbent.

Table 12: Intra- and inter-marker reliability and agreement for IV-RoMmax recumbent and weight-bearing n=10

Inter-vertebral level	Intra-marker ICC (95% CI)	Inter-marker ICC (95% CI)	Intra-marker SEM (°)	Inter-marker SEM (°)
Recumbent				
L2-L3	0.94 (0.76-0.99)	0.86 (0.54-0.97)	0.26	0.31
L3-L4	0.98 (0.94-1.0)	0.95 (0.83-0.99)	0.17	0.25
L4-L5	0.94 (0.78-0.98)	0.97 (0.90-0.99)	0.42	0.41
L5-S1	0.93 (0.74-0.98)	0.83 (0.45-0.96)	0.64	0.69
Weight-bearing				
L2-L3	0.98 (0.92-1.0)	0.94 (0.80-0.99)	0.45	0.76
L3-L4	0.99 (0.96-1.0)	0.99 (0.67-1.0)	0.23	0.24
L4-L5	0.99 (0.97-1.0)	0.98 (0.93-1.0)	0.39	0.59
L5-S1	0.96 (0.82-0.99)	0.99 (0.94-1.0)	0.54	0.61

5.1.7.3 Initial attainment rate

The ICC's (reliability) and SEM's (agreement) for both intra- and inter-marker weight-bearing initial attainment rate studies are shown in table 13. The reliability of QF initial attainment rate measurements was 'substantial' (Shrout 1998) being more than 0.81 in both intra- and inter-marker studies at all inter-vertebral levels (Table 13). The smallest ICC was 0.84 at the level of L3-L4 in the inter-marker study, and the largest was 0.98 in the intra-marker study at the same level. The intra-marker study demonstrated consistently better reliability (including narrower confidence intervals) than that of the inter-marker study. Generally the lower limits of the 95% CI were 'moderate' – 'substantial' (Shrout 1998), however in the inter-marker study, at the inter-vertebral levels of L3-4 and L5-S1, the lower limits were 'fair' (0.49 and 0.53) respectively.

The agreement of initial attainment rate measurements is also acceptable in both intra- and inter-marker studies. In the upper inter-vertebral levels (L2-3 and L3-4) SEM's are comparatively lower in the intra-marker study, however in the lower levels (L4-L5 and L5-S1) SEM's are comparatively higher.

Table 13: Intra- and inter-marker reliability and agreement for initial attainment rate weight-bearing n=10

Inter-vertebral level	Intra-marker ICC (95% CI)	Inter-marker ICC (95% CI)	Intra-marker SEM _{ratio}	Inter-marker SEM _{ratio}
L2-L3	0.95 (0.78-0.99)	0.95 (0.80-0.99)	0.026	0.036
L3-L4	0.98 (0.92-1.0)	0.84 (0.49-0.96)	0.020	0.033
L4-L5	0.92 (0.71-0.98)	0.91 (0.70-0.98)	0.032	0.018
L5-S1	0.95 (0.81-0.99)	0.88 (0.53-0.97)	0.023	0.019

5.1.8 Discussion

5.1.8.1 IV-RoM

The widely used ICC statistic has no standard values for acceptable reliability (Mieritz et al. 2012); however there have been many attempts to quantify its meaning in the literature. The ICC is expressed as a value between 0 and 1, and a value of >0.7 is generally accepted as reliable; however there remains no definite consensus (Ailliet et al. 2015). For example, it has been suggested that values above 0.75 are indicative of good reliability (Portney and Watkins 2009), but Aaronson et al (2002) recommend coefficients of >0.70 for group comparisons, and >0.90 for individual measurements as a minimal standard (Aaronson et al. 2002). The interpretation of the ICC is therefore somewhat subjective in terms of what is acceptable or not. Shrout (1998), attempts to address this problem by providing adjectives that describe the different ranges of reliability values (Shrout 1998).

(0.00-0.10) - virtually none;

(0.11-0.40) - slight;

(0.41-0.60) - fair;

(0.61-0.80) - moderate;

(0.81-1.0) - substantial.

This has been adopted by a previous QF weight-bearing study in the cervical spine (Branney 2014) and serves as a reference point for the results in this chapter.

The interpretation of QF agreement and reliability studies is also made difficult by the use of different ICC types. For example Teyhen and Mellor use an ICC (2,1) (Teyhen et al. 2005, Mellor F.E. et al. 2014), whereas Yeager et al (2014) and Branney and Breen (2014) use an ICC (3,1) with

in some cases only brief explanations of the ICC selection process (Branney and Breen 2014; Yeager et al. 2014).

The agreement and reliability of IV-RoM and initial attainment rate measurements using QF have previously been assessed in recumbent participants (Mellor F.E. et al. 2014). As anticipated, the reliability and agreement of IV-RoM measurements during recumbent sagittal flexion were found to be similar to those found in the current work, with 'substantial' reliability (Shrout 1998), and acceptable error (i.e. $<1^\circ$) demonstrated at all levels for both intra- and inter-marker studies. It has been demonstrated that reliability and agreement are typically decreased in the inter-marker group (Mellor F.E. et al. 2014, Yeager et al. 2014), however these differences were shown to be minimal in the current study, and there were notable exceptions to the trend (Table 12). Although ICC's were very similar between intra- and inter-marker groups, generally the width of the CI's and the SEM's were larger in the latter. It appears that whilst errors arising from the use of different markers do have a small impact, inter-marker agreement and reliability is still acceptable.

Inter-marker agreement and reliability was assessed using two independent markers. At this stage the only examiners of interest are those with expertise of using the technology and of image marking. If the QF protocol were ever incorporated on a larger scale (within the NHS for example), then further agreement and reliability studies using 3 or more markers may be advisable, as a larger population of template markers would require assessment.

Interestingly, when comparing the findings of Mellor et al. (2014) to the current study, 95% confidence intervals are generally wider in both for recumbent inter-marker results at the level of L2-L3 (0.037-0.891) (Mellor F.E. et al. 2014) and (0.54-0.97) respectively. Mellor et al. did not include L5-S1 in their study and so no comparison can be made, however as demonstrated, the current study mirrored this finding at L2-L3 and also showed comparatively wider confidence levels at L5-S1 (Table 12). It would appear therefore, that there is a marginal decrease in reliability of measurements at inter-vertebral levels closer to the edge of the image field (i.e. L2-3 and L5-S1). The anticipated difficulty (due to superimposition) in template marking/tracking of L5-S1 was the reason cited by Mellor et al for its exclusion (Mellor et al. 2014), and it makes sense that tracking problems are more likely to occur in templates that partially leave the image field i.e. L2-3 and L5-S1). The current study's results have shown however, that reliability (using the criteria of Shrout 1998) and agreement of QF measurements at all levels, including L2-L3 and L5-S1, can be achieved in both recumbent and weight-bearing protocols, at an acceptable level. The sample size was restricted to 10 due to time and resource constraints, however many other

studies have also used this number (Branney and Breen 2014; McGregor et al. 1995; Mellor F.E. et al. 2014). If a larger sample was possible, narrower CI's would be expected. A more detailed justification for the sample size chosen can be found elsewhere (Appendix T).

For the weight-bearing group's intra-marker results, the closest comparisons can be made with the work of (Teyhen et al. 2005). Reliability was similar to the current study with ICC's >0.96 at all inter-vertebral levels, however the IMRCI QF protocol for IV-RoMmax measurement appears to show a modest improvement. Intra-marker agreement was marginally better in the current study, with comparative SEM ranges of (0.4°-0.7° and 0.23°-0.54°). Teyhen did not conduct an inter-marker study, and so no direct comparisons can be made with this group. In terms of participant numbers, Teyhen et al. (2005) recruited double the number of the current study (n = 20) and so it may be expected that there would be narrower confidence intervals, unfortunately these were not reported (Teyhen et al. 2005). Interestingly, the current study demonstrated larger ICC's in the weight-bearing groups than those found in the recumbent. These results confirm the agreement and reliability of measurements made using the weight-bearing protocol, however it should be noted that SEM's were generally larger in this group.

An advantage of the QF methodology is the analysis of continuous data, therefore if IV-RoMmax occurs before or after full flexion, it will not be missed. Other QF protocols simply use images at pre-designated points of the flexion cycle. Teyhen et al., used the single upright and fully flexed image, and Ahmadi et al used sample points at 0, 25, 50, 75 and 100% of the cycle ROM (Ahmadi et al. 2009, Teyhen et al. 2005). It is possible that both methods will therefore have missed the true measurement of IV-RoMmax.

The current study does not include ICC's and SEM's for pooled inter-level data. It was decided that pooling all participants at all levels may give an incorrect impression of results from 40 participants and not 10. To do so, would also conflate the 'between subject variation' and the 'between inter-vertebral level variation', thus obscuring the relevance of the ICCs. A possible solution would be to calculate the ICC using the sum of all inter-vertebral values for each participant, effectively reverting the participant number back to 10. This however was also deemed inappropriate, as there would be an accumulation of the individual errors from each inter-vertebral level. The concept of cumulative errors occurring in studies that do not report individual inter-segmental levels is a possible criticism of Yeager et al, who base their conclusions on overall ICC's (Yeager et al. 2014).

The results have shown that the agreement and reliability of IV-RoMmax measurements made using the weight-bearing sagittal plane QF protocol are acceptable at all inter-segmental levels, including L2-3 and L5-S1. The concern that weight-bearing images may be of inferior quality, and subsequently adversely affect the agreement and reliability of measurements was unfounded. ICC's between intra- and inter-marker groups were broadly similar; however there were typically wider CI's and larger SEM's in the inter-marker group.

5.1.8.2 Initial attainment rate

This is the first time initial attainment rate has been reported using image data from the QF lumbar sagittal weight-bearing protocol, and so the agreement and reliability of measurements is thought never to have been investigated. Testing of the accuracy of the parameter is not currently feasible, as it is a novel concept; there are no existing reference standards to compare it to. Initial attainment rate as described using the QF protocol is an idea that is being developed by the IMRCI group, and the advancement of its use as a research parameter is in the early stages. In a recent study however, Breen et al. (2015), demonstrate how the variable can be viewed as analogous to the neutral zone in lateral flexion (Breen et al. 2015).

Initial attainment rate measurements were highly repeatable with ICC's ranging from (0.84-0.98) and SEM's from (0.018-0.036). An unexpected result was the relative decrease in measurement error shown in the lower levels (L4-L5 and L5-S1) in the inter-marker study when compared to the intra-marker. There is no obvious explanation for this result; however as template marking experience increased, a shift in technique that increased the alignment between 1st and 2nd markers may have occurred.

Two previous QF studies have reported results using initial attainment rate data from sagittal plane (also referred to as attainment rate), and their findings are broadly similar to those of this study. Mellor (2014), showed in the lumbar spine (L2-L5) that like IV-RoM, initial attainment rate measurements were also repeatable using a recumbent protocol, however the results demonstrated particularly good intra-marker agreement and reliability, with the largest CI for ICC being (0.766-0.982 at L3-L4), and their largest SEM_{ratio} being (0.009) at the same level (Mellor 2014). These results represent an improvement over those seen in the current study, although comparisons between recumbent and weight-bearing were broadly similar. The improved intra-marker results seen in Mellor's study may be a reflection of their greater template-marking experience at the time of the investigation (Mellor 2014). If the templates are unlikely to change between the first and second mark-ups, then intra-marker agreement and reliability will increase.

The agreement and reliability of initial attainment rate measurements was shown to be comparatively reduced in the upper most and lowest inter-vertebral segments (L2-3 and L5-S1, mirroring the trend seen in the IV-RoMmax study. Branney (2014) demonstrated the agreement and reliability of QF initial attainment rate measurements in the cervical spine (Branney 2014). Their results also demonstrated comparable initial attainment rate measurements to those found in the current study, and again the widest CI's were shown to be at the inter-vertebral levels at the edge of the image field (i.e. C1-C2 and C5-C6). This highlights the impact of human error when templates are marked at these levels. Both the Branney and Mellor studies also analysed 10 participants, which makes their results directly comparable to those reported in this study.

5.1.8.3 Summary

An investigation into the agreement and reliability of weight-bearing IV-ROM and initial attainment rate measurements was required for the following reasons.

- To assess the competency and skill of the author with regards the marking-up of image templates.
- To date, only recumbent image data has been published using the QF measurement system, and have excluded the level of L5-S1. Weight-bearing measurements have never been analysed using the IMRCI QF system, and the QF studies that have previously investigated agreement and reliability of these measurements have either not been conducted at an inter-segmental level (Yeager et al. 2014), have not included all the inter-segmental levels required for use in this study (Mellor F.E. et al. 2014; Teyhen 2005), have not used continuous data to find the IV-RoMmax (Ahmadi et al. 2009; Teyhen 2005) or not conducted an inter-marker study (Ahmadi et al. 2009; Teyhen 2005).
- The preliminary studies (Chapter 3) demonstrated that IV-RoM motion patterns are more variable during weight-bearing protocols than recumbent. The associated occurrences of phenomena such as double peaks, and paradoxical motion, may affect the determination of kinematic variable measurements.
- Weight-bearing protocols typically require a greater radiation dose to achieve the same image quality as those obtained with recumbent examinations. If radiation exposure is kept to a minimum, weight-bearing images are characteristically poorer, which can subsequently affect image quality. Analysis of weight-bearing measurements was therefore particularly important as agreement and reliability

could be compromised when compared to measurements processed from recumbent protocols.

- Superimposition of bony structures (e.g. iliac crest, L5 vertebral body) at the base of the lumbar spine, make marking-up and tracking of the L5-S1 level more difficult than the others (L2-L5). As the measurements of IV-RoM and initial attainment rate at L5-S1 are required in this study, and that it is hypothesised that there may be a higher incidence of marking-up and tracking difficulties at this level, L5-S1 measurement agreement and reliability required assessment.
- The agreement and reliability of lumbar weight-bearing initial attainment rate measurements has never been investigated. Initial attainment rate agreement and reliability has been determined in the cervical spine (Branney 2014), and in the recumbent lumbar spine (Mellor 2009), but never at the level of L5-S1.

Given the highly repeatable IV-RoMmax and initial attainment rate measurement results, the author's marking ability was of an adequate standard. Weight-bearing measurements of both parameters were achieved with acceptable agreement and reliability at all levels, including those nearest the edge of the image field (L2-3 and L5-S1). Superimposition problems at the level of L5-S1, increased occurrences of paradoxical motion and double peaks, and image quality issues associated with weight-bearing images, did not markedly affect agreement and reliability.

5.1.9 Conclusion

IV-RoMmax and initial attainment rate measurements made using the weight-bearing sagittal plane QF protocol demonstrated acceptable agreement and reliability. In chapter 6 relationships between these parameters during weight-bearing sagittal flexion are explored. IV-RoMmax is also used in chapter 7 where correlations between maximum angular range and muscle activity variables are investigated.

Note: All of the agreement and reliability investigations described above were conducted by the author as a direct part of their PhD research. The results may however be incorporated into the ongoing normative database study (4.2.1) in the future. Due to resource and ethical constraints, weight bearing flexion test-retest reliability studies were not conducted. This has since been investigated by the IMRCI group however, showing the inter-session (i.e. 6 weeks apart) ICC for weight bearing flexion IV-RoMmax as 0.82 (0.73-0.88). All other directions (i.e. extension, right and left lateral flexion) were also reported as above 0.7. At this point this data is unpublished.

5.2 Part 2: Surface electromyography (sEMG) of the lumbar and thoracic paravertebral muscles during the weight-bearing sagittal plane QF protocol: An Intra- and inter-session sEMG agreement and reliability study.

5.2.1 Introduction

The weight-bearing sagittal plane QF protocol has never before been combined with sEMG, and the agreement and reliability of sEMG parameters during the examination has not been assessed. When measurements of muscle activity are made for research purposes, the methods used must be both reliable and repeatable, as inaccurate measurements can result in error being greater than the changes occurring in the muscle (Stokes 1985). The present study protocol was designed to reduce the effects of biological and systematic variations as much as possible, however its impact on the agreement and reliability of sEMG amplitude measurements remains unknown. Another important question in terms of methodological design was the decision to use the sEMG data from left and right sides individually, or whether to combine them, and use an average of both for each spinal level T9 (TES), L2 (LES) and L5 (LMU).

Historically, the literature has placed little emphasis on the reliability of sEMG amplitude parameters (Daneels et al. 2001), and in terms of the paraspinal muscles, the evidence suggests that agreement and reliability can be poorer than in other muscle groups (Ahern et al. 1986; Stokes et al. 1988). Therefore if meaningful relationships are to be found between sEMG amplitudes and intervertebral kinematic variables, their validity will depend heavily on the agreement and reliability of both sEMG and QF parameters, and so the different paraspinal muscles of interest (i.e. TES, LES and LMU) require assessment. This is especially important as EMG reliability has been shown to vary between different paraspinal muscles (Biederman et al. 1990). Biederman et al. (1990) also investigated the reliability of the RMS parameter during weight holding tasks, and showed it to be more reliable for the multifidus than over more cephalad muscles such as iliocostalis lumborum par thoracis (Biedermann et al. 1990). It was proposed that the difference may have been the result of the arms being raised during the test, and as the QF protocol also requires participants to have raised arms, the reliability of all muscle levels should be tested.

Objectives

- To determine whether there are significant differences in the mean normalised sEMG amplitudes between left and right sides.
- To determine the inter- and intra-session agreement and reliability of normalised sEMG amplitudes during the weight-bearing sagittal flexion and return QF protocol.

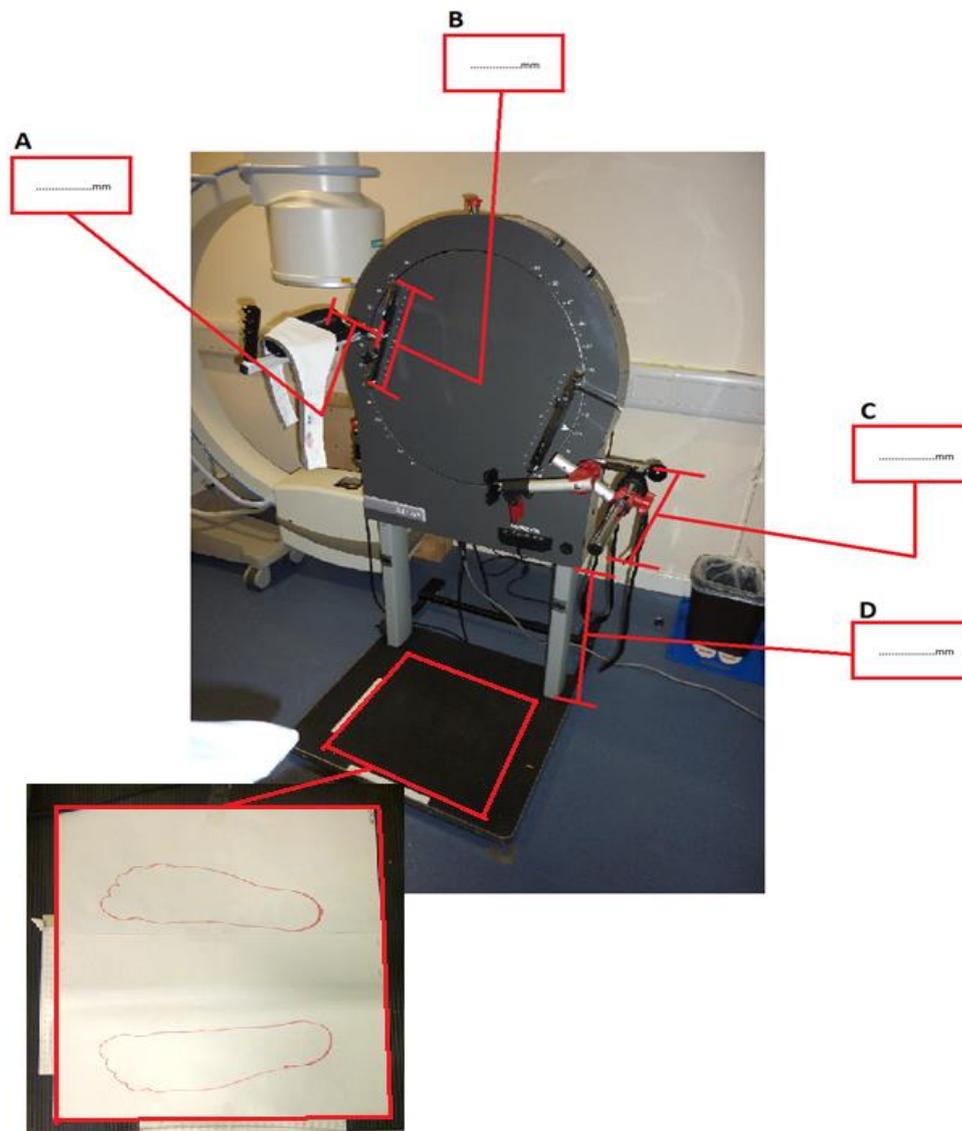
5.2.2 Methods

10 healthy males (mean age SD) were recruited to participate in the lumbar weight-bearing sagittal plane sEMG protocol. The details of the method have been outlined previously (Sections 4.2.8 and 4.2.14), however the following modifications were incorporated for the agreement and reliability study.

1. Participants received no radiation during the procedure, however lead shielding was worn to recreate QF test conditions.
2. The sEMG protocol was carried out twice by a single examiner. All follow up recordings took place within 1 week of the baseline, with a minimum of 2 days separation.
3. The acquisition cycle was repeated 4 times (several minutes apart) at baseline and follow up. Intra-session results compared cycles 1 and 2 (of the 4), whereas inter-session results were calculated as an average of the 4 mean normalised amplitudes recorded over the cycle duration.
4. In an attempt to ensure that the electrodes were re-applied in the same position at follow-up, the baseline electrode positions were recorded using an indelible ink tracing around each of their borders, and by using a transparent electrode positioning map.
5. In order to replicate the baseline participant positioning at follow-up, foot positioning was recorded by tracing around each participant's feet on A3 paper at baseline, and motion-frame apparatus positions were recorded (Figure 45).
6. The sMVC was determined once only for each participant at baseline (the same sMVC was used at follow-up).
7. The initiation of the movement of the motion frame and the beginning of sEMG recordings were synchronised by pressing 2 start buttons simultaneously. (This system was replaced by an automated switch in the main study) (Section 4.2.20).

The sEMG tracing was recorded for approximately 20 seconds during each cycle. The mean normalised amplitudes over this period were used for analysis.

Figure 45: Motion frame apparatus and foot positioning



5.2.3 Ethical Approval

The AECC Research Ethics Sub-Committee granted approval for the sEMG sub-study (Appendix E).

5.2.4 Data analysis

Test re-test agreement and reliability were analysed using ICC's (reliability) and SEM's (agreement). Two different ICC types were required for the intra- and inter-session studies respectively. The intra-session study compared the mean amplitudes between cycles that took place only minutes apart, without the need to remove or replace electrodes, or change participant positioning. The intra-session results were therefore analysed using a 2 way-mixed consistency ICC using the single measures outputs. The inter-session results however needed to account for potential sources of error due to participant repositioning and electrode application

positioning. The inter-session results were therefore analysed using a 2 way-mixed absolute agreement ICC using average measures outputs. Average measures outputs were used as the results were the comparison of the mean of 4 cycles from the baseline and follow-up sessions. The formula for calculation of the SEM's is the same as for the intra- and inter-marker studies (Section 5.1.6.3).

As the author was the only member of the IMRCI research group with the ability to conduct the sEMG experiment, it was not possible to conduct an inter-examiner study.

5.2.5 Results

There were no statistically significant differences between left and right normalised sEMG amplitudes at any level (Table 14). There were examples of notable differences in certain participants however, (for example participant 6 (Table 15)).

The decreased reliability and agreement expected due to participant repositioning, electrode re-application, and variations in an individual's bending movement between trials was minimal. Although there were distinct differences between the 95% CI's and the SEM's between the intra- and inter-session studies, the ICC's for both at all muscles levels were in the 'substantial' category i.e. >0.81 (Table 16). CI's were however comparatively wider at all muscle levels in the inter-session study, the widest range being 0.508-0.968 for LES. The muscle amplitudes of LMU were most consistently reliable, with the CI's lowest range being >0.9. In terms of agreement the largest error was found for LES in the inter-session study, and the smallest was found for TES in the intra-session study. TES demonstrated less error than both of the lower muscle levels, for intra- and inter-session studies (Table 11).

Table 14: t-tests comparing the mean normalised sEMG amplitudes (% of sMVC) of left and right sides n=10

	n	Mean	SD	t	df	p
TES Left vs Right	10	1.9	7	0.849	9	0.418
LES Left vs Right	10	-4.3	7.7	-1.753	9	0.113
LMU Left vs Right	10	-5.8	11.7	-1.567	9	0.152

Table 15: Left and Right normalised amplitudes (% of sMVC) cycles averaged from the baseline group n=10

Participant	TES	TES	LES	LES	LMU	LMU
	left	right	Left	Right	left	right
sEMG01	12.9	11.2	15.2	16.9	28.8	31.1
sEMG02	9.1	9.4	15.7	18.4	31.6	30.7
sEMG03	16.1	16.7	17.8	15.8	32.5	30.0
sEMG04	10.2	17.3	21.2	21.5	31.1	30.9
sEMG05	5.3	5.0	8.8	16.5	23.7	30.6
sEMG06	38.7	20.3	28.8	53.8	31.9	70.0
sEMG07	11.6	8.9	12.7	12.5	8.5	10.0
sEMG08	3.9	4.4	5.1	7.9	6.0	10.4
sEMG09	16.1	7.9	12.1	13.5	11.4	17.0
sEMG10	18.4	22.2	21.9	25.3	36.2	39.1
Shapiro-Wilk	0.025	0.378	0.990	0.002	0.030	0.040
Mean	12.9	11.2	14.6	18.4	24.2	30.0
S.D	10.2	7.0	7.9	13.5	11.2	17.1
Median	12.3	10.3	15.5	16.7	30.0	30.6
Upper Q	16.1	17.2	20.4	20.8	31.8	31.0
Lower Q	9.3	8.1	12.2	14.0	14.4	20.2

Table 16: Intra- and inter-session reliability and agreement for sEMG normalised amplitudes during weight-bearing plane QF protocol n=10

	Intra-session ICC (95% CI)	Inter-session ICC (95% CI)	Intra-session SEM (%)	Inter-session SEM (%)
TES	0.996 (0.986-0.999)	0.895 (0.606-0.974)	0.5	2.7
LES	0.984 (0.939-0.996)	0.872 (0.508-0.968)	1.2	3.9
LMU	0.990 (0.961-0.998)	0.974 (0.902-0.993)	1.4	2.8

5.2.6 Discussion

5.2.6.1 Comparing muscle activity between sides

The comparison between mean normalised sEMG amplitudes recorded from TES, LES and LMU, revealed no significant differences between left and right sides (Table 14). When comparing ipsilateral to contralateral, previous studies (Ahern et al. 1988; Lariviere et al. 2005; Lariviere et al. 2000; Oddsson and De Luca 2003) all report significant differences between left and right side paraspinal activity during dynamic movements. These studies investigated both LBP patients and healthy controls, whereas this study population consisted of only healthy controls, which may partially explain why a closer similarity between sides was achieved, although Reeves et al (2006), demonstrated no significant difference in the imbalance in sides between participants with a history of low back pain, and those without (Reeves et al. 2006).

There were however individual examples of clear differences between sides, such as in participant 6 (Table 15). These results will have increased the overall difference between left and right, although not to a significant level for the sample as a whole. It would seem therefore that a difference from side to side may be normal in a population of healthy participants. With this in mind, the decision to use the mean of both sides combined (as utilised by Reeves et al. 2006) for main study analysis was taken.

The standardisation of the QF protocol results in each participant bending at the same rate and over the same range. Even without the standardised motion, the sagittal plane is perhaps most suitable for kinematic assessments of the spine, as relative to the frontal plane, there are a lack of coupled movement patterns (Keessen 1984).

In order to avoid excessive deviation from the sagittal plane, other safeguards were put in place. In a recent lumbar spine kinematics study, Tafazzol et al. described how they instructed participants to remain in the sagittal plane (Tafazzol et al. 2014). As much like common sense as this may sound, the instruction to remain facing forwards (avoiding unnecessary rotation) was also given to this study's participants. The radiographer was also able to visually and radiographically assess each participant's movement. During the range of motion tolerance trials, the radiographer advised the participants against any excessive rotation of the head and shoulders, and also recorded a single frame at the end of each practice range (e.g. 60°). This allowed the visual assessment of any rotation that had occurred in the lumbar vertebrae, and participants were instructed to alter their movement pattern if required. These safeguards may

have contributed to a more equal share of muscle activity between sides, and the overall high agreement and reliability seen between sessions.

Lariviere et al. (2009) demonstrated that providing visual feedback of all out of plane exertions, could reduce unwanted out of plane moments, and significantly alter activation amplitudes (Lariviere et al. 2009). This was found to be the case particularly in the frontal and transverse planes, but feedback had less of an effect in the sagittal plane. In a follow up study however, they also demonstrated that the use of a 3-D visual feedback system did not actually decrease the within-subject variability, and that learning how to use the system (over three assessment days) had negligible effects on both coupled moments and EMG variables (Lariviere et al. 2014). Therefore no additions to existing safeguards were thought necessary for the main study data collection of this project.

5.2.6.2 Reliability

It is recommended that any procedures to be used in EMG studies should undergo reliability testing (Soderberg 2000). A common problem with sEMG studies is the great variability in their findings (Geisser et al. 2005; Van Dieen et al. 2003), and so the high reliability shown in this study is reassuring.

It is usual for a proportion of variability to be attributed to a lack of standardisation, and the method by which EMG variables are normalised (Lariviere and Arsenault 2008). The results however show 'substantial' reliability for both intra-day and inter-day sessions, indicating that the standardisation of movement range, speed and direction provided by the QF protocol may have played an important role in reducing the impact of variability resulting from these causes. It is difficult to compare reliability with other studies as the protocols are invariably very different (Thuresson et al. 2005), however this study produced comparable intra-session results to two similar investigations. Daneels et al (2001) showed the reliability of amplitude (averaged EMG) of lumbar multifidus and iliocostalis lumborum pars thoracis during flexion and return to be 'moderate' to 'substantial' ($ICC > 0.75$) (Daneels et al. 2001), as did Lariviere et al. (2000), when they investigated EMG amplitude reliability of the TES and LES muscles during a flexion and extension task (Lariviere et al. 2000). All intra- and inter-session results showed ICC's > 0.85 , an improvement on these previous studies (Daneels et al. 2001; Lariviere et al. 2000), suggesting that the standardisation of the QF protocol may increase reliability when compared to sagittal flexion in an uncontrolled environment.

There is a clear comparative decrease in reliability in the inter-session group, most markedly seen in the wider CI's (Table 16), yet reliability as a whole remains high. Other studies show a distinctly lower inter-session reliability compared to intra-session groups, especially for amplitude components (Daneels et al. 2001; Jobson et al. 2013; Kollmitzer et al. 1999). The 'substantial' reliability (ICC's >0.81) seen in the current study's inter-session group, may be an indication therefore that the procedures and standardisations put in place to recreate test conditions between sessions worked well. EMG remains however a very sensitive parameter (Daneels et al. 2001), and it has been shown that even small changes in electrode position can have a significant effect on recorded sEMG amplitudes (De Nooij et al. 2009) also see (Section 3.3). The attention to accurate participant re-positioning (Figure 45) and electrode re-application (Figure 42) may therefore be important factors in keeping the influence of these sources of error to a minimum.

As with the kinematic reliability studies, a common problem encountered when critiquing the literature is that the authors often fail to provide enough detail about the statistical tests they employ. Williams et al. (2013) for example, show that highly reliable peak magnitude is achievable, with intra-session ICC's of 0.97 and 0.96 for acute and chronic low back pain groups respectively (Williams et al. 2013b). However, the authors do not refer to the type of ICC test used, so evaluation of these results is difficult. The Williams et al. (2013) study also includes no reference to agreement, a common omission in the sEMG agreement and reliability literature making comparisons within individuals problematic (De Vet 2006).

5.2.6.3 Agreement

As referred to above, it is difficult to compare sEMG amplitude agreement with other studies as it is not typically mentioned (Daneels et al. 2001; Lariviere et al. 2000; Williams et al. 2013b), and the methods and normalisation techniques vary between studies. The sEMG amplitude agreement in this study is acceptable with the largest intra- and inter-session error being <1.5% and <4% respectively (Table 16). As seen in the inter-vertebral kinematic IV-RoM results, the inter-session group demonstrated typically greater error than the intra-session group, and agreement was also consistently the best for TES. These seem therefore to be linked. In a study that did attempt to quantify the measurement error of surface EMG, Thuresson et al. (2005), included the standard error of measurement, which was calculated as the within-subject standard deviation (Thuresson et al. 2005).

5.2.6.4 Summary

All intra- and inter-session preparation and testing were conducted by the author, and the consistency this provided is reflected in the high agreement and reliability found in both groups, but particularly in the intra-session study. The method represents a more regulated forward flexion and return protocol than that seen elsewhere in the literature, and produces highly repeatable sEMG amplitude measurements. The intra-session agreement and reliability was found to be better than the inter-session, most likely because there was no requirement to remove and re-apply electrodes between cycles. Despite this requirement for the inter-session study, agreement and reliability was still found to be 'substantial' (Shrout 1998).

The standardisation of the rate, range and direction of movement, keeping testing periods to the same time of day, keeping the environment at the same temperature, and using accurate electrode mapping techniques, are all factors that may have contributed to the high agreement and reliability of the results. These results can only be applied to the specified population; however the sample is reflective of the age group and gender of those recruited into the main study.

5.2.7 Conclusion

There was no significant difference in mean normalised sEMG amplitude over the flexion and return cycle between left and right sides, and therefore the mean of left and right sides will be used in the main study. The results also indicate that normalised sEMG RMS amplitudes are a repeatable measure of muscular activity during the weight-bearing QF sagittal forward bending and return protocol, and as such are suitable for use in the main study. The stability of sEMG amplitude measurements demonstrated by both the intra- and inter-session results, suggest that the methods are suitable for both cross-sectional and longitudinal studies conducted over a similar time-frame. Relationships between sEMG and kinematic parameters (IV-RoMmax) are investigated in chapter 7.

Note: The agreement and reliability results from sections 5.1 and 5.2 have been published in peer reviewed journals (Du Rose and Breen 2016a, Du Rose and Breen 2016b) (Appendices Q and R). In addition, the above studies conform to the Quality Appraisal of Reliability (QAREL) Checklist (Lucas 2010) (Appendices N and O).

Chapter 6

6.1 Relationships between lumbar inter-vertebral motion and lordosis in adult males: a cross sectional cohort study

6.1.1 Introduction

The bulk of the following chapter is a peer reviewed paper published in BMC Musculoskeletal Disorders (Du Rose and Breen 2016b) by the author and this study's lead supervisor¹⁰, and conforms to the Strengthening the Reporting of Observational Studies in Epidemiology (STROBE) guidelines (Von Elm et al. 2008). The paper is not presented in its entirety as the agreement and reliability findings were presented in the previous chapter. In the general discussion section however there have been substantial additions, and there are some methodological details that replicate previous aspects of the thesis. The paper addresses the study's secondary aims (Section 2.7.2.4) by exploring the relationships between IV-RoMmax and lordosis, initial attainment rate and the IV-RoMmax at other lumbar levels.

6.1.2 Background

Movement of the lumbar spine requires the participation of multiple motion segments and the relative contributions of these segments are a function of their own mechanical properties (Sahrmann 2002). Aberrant spinal movement patterns are widely thought to be related to musculoskeletal pain and dysfunction (Iguchi 2004; Kanemura 2009; Spinelli et al. 2015), and as such they are used to inform surgical and conservative clinical decision making (Fritz et al. 2007; O'sullivan 2005; Sahrmann 2002; Steiger et al. 2014), and as indicators of spinal stability (Fritz et al. 1998; Kanemura 2009; Panjabi 1992a, 1992b). As a consequence of their wide variation in both low back pain and healthy populations however, the clinical importance of factors such as inter-vertebral range of motion (IV-RoM) remains unclear (McGregor et al. 1997), and the identification of biomechanical factors that may contribute to low back pain, remains a challenge (Mellor et al. 2014). Information about how IV-RoM may interact with other biomechanical factors may therefore help provide a better understanding of how variations in lumbar inter-vertebral kinematics may affect prognosis and treatment outcomes.

The starting point for this should be the collection of detailed normative quantitative data with respect to *in vivo* inter-vertebral motion and morphologic parameters (Li 2009). Quantitative

¹⁰ Professor Alan Breen

fluoroscopy (QF) has been shown to be an accurate and reliable 2D method of doing this (Mellor et al. 2014, Teyhen et al. 2005, Yeager et al. 2014). Recent technological advances have enabled the acquisition of 3D lumbar kinematic data *in vivo* (Aiyangar et al. 2014, Harvey et al. 2015), however it has been demonstrated that there is only minimal axial rotation and lateral bending associated with movements in the sagittal plane (Ellingson and Nuckey 2015; Harvey 1998; Pearcy 1985), and in terms of QF inter-vertebral measurements, out of plane motion of up to 10° does not significantly affect accuracy (Breen et al. 2006). Therefore, the greater expense and dose associated with current 3D techniques weighted against the potentially negligible clinical and research benefits, justify the use of 2D QF technology, particularly in the sagittal plane. Indeed, the investigation of spinal mechanical behaviour has been outlined as a priority for future QF research (Breen et al. 2012), which begins with the relationships between IV-RoM and other kinematic variables in healthy, pain-free control populations. Such normative information should provide insights into the possible biomechanical consequences of changes within each.

Previous dynamic studies using fluoroscopy have highlighted contrasting ranges and patterns of angular rotation between the upper and lower lumbar motion segments (Ahmadi et al. 2009; Kanayama 1996; Lee 2002; Li 2009; Okawa 1998; Wong 2006; Wong 2004; Wu et al. 2014; Xia 2009), which make different contributions to movements such as sagittal flexion. There is also evidence to suggest that lordosis may relate to an individual's spinal flexibility (Been and Kalichman 2014). Indeed, a recent MRI study that investigated the intrinsic shape of the lumbar spine concluded that lumbar spinal shapes may be related to an individual's risk of injury (Pavlova et al. 2014).

IV-RoM is the most commonly reported measure of inter-vertebral motion (Mellor 2014; Pearson et al. 2011; Teyhen 2005) and attainment rate (defined as the velocity with which IV-RoM is reached), has been identified as a reflection of intervertebral restraint (Mellor et al. 2014, Teyhen et al. 2007, Wong et al. 2004). Initial attainment rate is a refinement of this which measures the slackness of an inter-vertebral motion segment in its initial phase of rotation (Breen et al. 2012, Mellor et al. 2009, Mellor et al. 2014). This parameter has been shown to correlate with the dynamic neutral zone (Breen et al. 2015), and is therefore also believed to be of importance when considering the stability of motion segments. Relationships between these and other kinematic and morphologic variables have not been investigated previously. This study examined the relationships between IV-RoMmax at lumbar inter-vertebral levels from L2

to S1 and lordosis, initial attainment rate and IV-RoMmax at other lumbar spine levels during forward bending in healthy controls. It was hypothesised that:

- A. There will be a direct relationship between the size of lordosis and IV-RoMmax in the upper lumbar segments, and vice versa
- B. There will be an inverse relationship between IV-RoMmax in the upper and IV-RoMmax in the lower lumbar segments
- C. There will be a direct relationship between initial attainment rate and IV-RoMmax at the same level

6.1.3 Methods¹¹

6.1.3.1 Study design

This was a cross-sectional, laboratory based cohort study of the relationships between L2-S1 IV-RoMmax and lordosis, initial attainment rate and IV-RoMmax at other levels (e.g. relationships between L2-L3 and L4-L5 IV-RoMmax).

6.1.3.2 Participants

The eligibility criteria for the study are shown in Table 17. Twenty male participants from the Anglo-European College of Chiropractic (AECC) student population were recruited. National Research ethics Service (NRES) approval was gained for the study (Bristol 10/H0106/65) and written informed consent was obtained from all participants prior to data collection. A participant number of 20 was selected, as a sample size ≥ 12 has been recommended as sufficient for the precision around the measurement to be used in an exploratory study (Julious 2005).

¹¹ Note: As this chapter is based on a published paper, the methodology is presented in its entirety. This means that there is some duplication in terms of the methodology (previously outlined in Chapter 4). This decision was taken to maintain the natural flow of the work, and for the ease of the reader.

Table 17: Eligibility criteria (duplication of the information in table 10)

Inclusion	Exclusion
Males aged 20-40 years	Inadequate understanding of English
An ability to understand written information	Currently receiving treatment for osteoporosis
Willing to participate and able to give informed consent	A history of recent abdominal or pelvic surgery
Consent to General Practitioner being informed	A history of previous lumbar spine surgery
A BMI < 30	A BMI > 30
No history of low back pain that prevented normal activity for at least 1 day in the previous year	Any medical radiation exposure in the past year or exposure in the past 2 years with a dose greater than 8 mSv
	Involvement in any other ongoing research

6.1.3.3 Data collection and processing

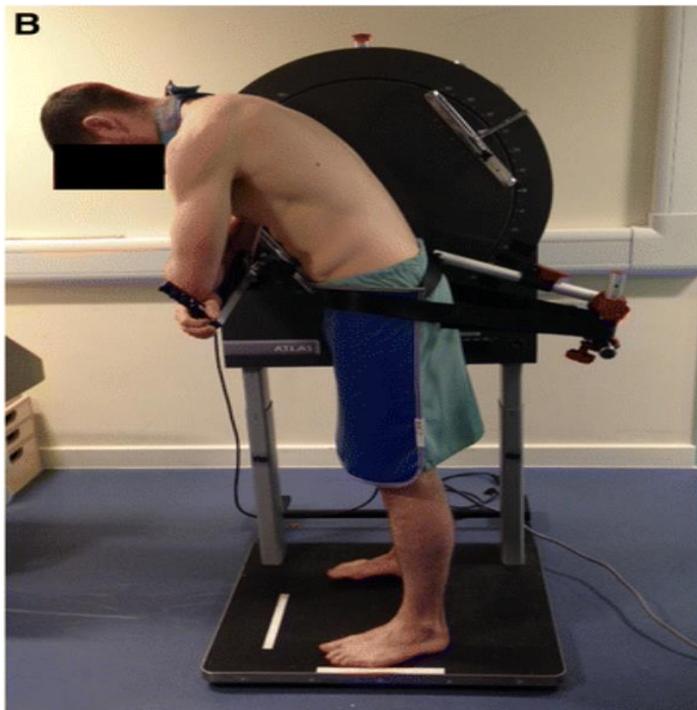
All data collection was conducted at the radiology department of the AECC. Fluoroscopic images of the lumbar spine were collected at 15 Hz using a Siemens Arcadis Avantic VC10A digital fluoroscope (CE0123) and a motion frame which acted to both stabilise the participants and guide their bending motion. Participants were asked to stand in a neutral upright position with their right side against the motion frame (Figure 46), and shadow the movement of a rotating arm rest which guided them during continuous fluoroscopic imaging, through a standardised range of 60° of forward flexion and return to upright, over a period of approximately 20 seconds. A review of spinal ranges of motion in controls proposed that the lumbar spine has an overall range (inclusive of both flexion and extension components) of approximately 80°, with 60° of this attributable to the flexion component (Dvorak et al. 1991). It was therefore theorised that the majority of each participant's lumbar inter-vertebral rotation would be completed within this range.

Prior to image acquisition, participants were taken in 20° stages through to the full 60° to safeguard that they were able to tolerate the movement. The movement of the motion frame was recorded by electronic feedback from its motor drive, and synchronised with the fluoroscopic imaging. To minimise bending from the hip joints, the pelvis was stabilised (Pearcy et al. 1984, Dvorak et al. 1991, Mellor et al. 2014, Du Rose and Breen 2016a, Du Rose and Breen 2016b) using a strap secured around the anterior superior iliac spine bilaterally, and attached to an appendage of the motion frame directly posterior to the participant (Figure 47).

Figure 46: Fluoroscope and weight-bearing motion frame

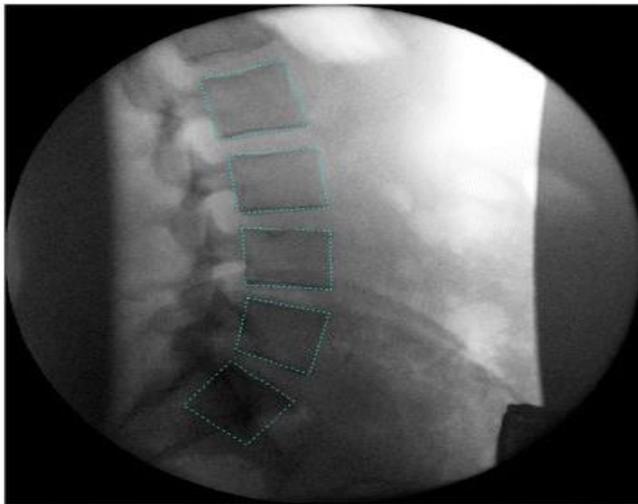


Figure 47: Weight-bearing motion frame during flexion including pelvic restraint mechanism and lead protection



A lead apron was worn to shield the gonads, and participants were verbally reminded to maintain a neutral bending position during the flexion cycle. The position of the central ray was targeted at L4 to make sure that all vertebrae (L2-S1) were included in the image field (Figure 48).

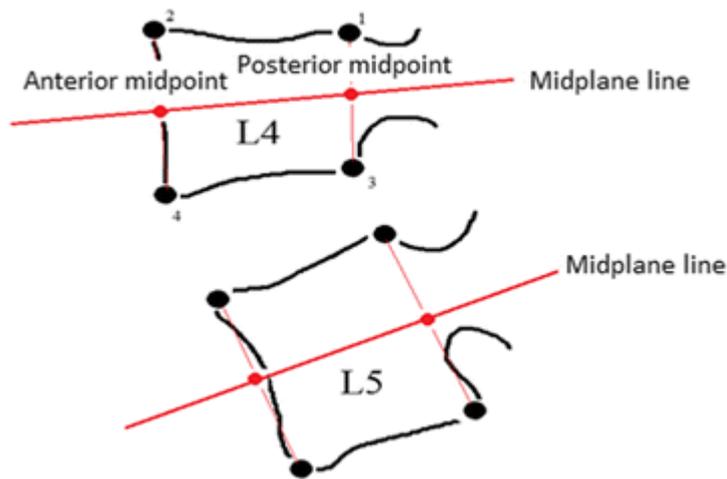
Figure 48: Fluoroscopic image of the lumbar spine. Templates placed around the lumbar vertebrae (L2-S1) on the first frame of the QF sequence



The fluoroscopic sequences were then transferred to a desk top computer for analysis using bespoke image processing codes written in Matlab (The Mathworks, Cambridge). Using the screen cursor, the outlines of each vertebra from L2-S1 in the first image of each sequence were marked-up manually with an electronic template. In order to increase precision, this process was replicated five times for each sequence and the results were averaged. In all subsequent image frames the software tracked each vertebra automatically, creating a continuous measurement of its movement throughout the flexion and return bending sequence. To ensure that template tracking was maintained throughout the sequence, visual checks were made using video playback.

The data collected comprised of range of motion (IV-RoM), initial attainment rate, and lordosis and the reliability and agreement of the first two of these were assessed as part of the study (De Vet 2006) (see chapter 5). The technique used to measure changes in inter-vertebral angle was established elsewhere (Frobin 1996), and is shown in figure 49.

Figure 49: Frobin method to measure the change in inter-vertebral angle. Rotation is calculated as the angle between the two mid-plane lines



IV-RoMmax for each inter-vertebral level (L2-S1) was calculated as the maximum angular range reached at any point throughout the 60° flexion and return cycle (Figure 50). Initial attainment rate for each level was calculated as the ratio of the slopes of motion frame movement and the inter-vertebral rotation over the first 10° immediately following the onset of inter-vertebral motion. The calculation of this variable has been outlined elsewhere (Mellor et al. 2009), and is also shown in figure 51. Lordosis was measured as the sum of all inter-vertebral angles (L2-S1), from the first image in the sequence. All participant data were anonymised.

Figure 50: Calculation of the maximum angular range reached during flexion (IV-RoMmax)

Maximum angle of rotation reached by each inter-vertebral motion pair (A); Maximum motion frame rotation (B) (always 60° during the QF sagittal flexion examination). Note: Maximum inter-vertebral range of motion may not always be found at the end of motion frame movement range

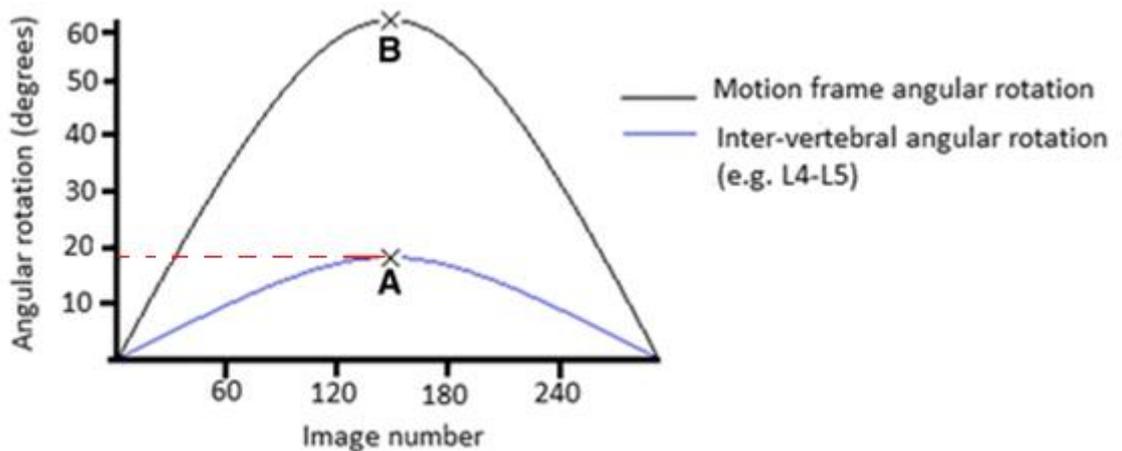
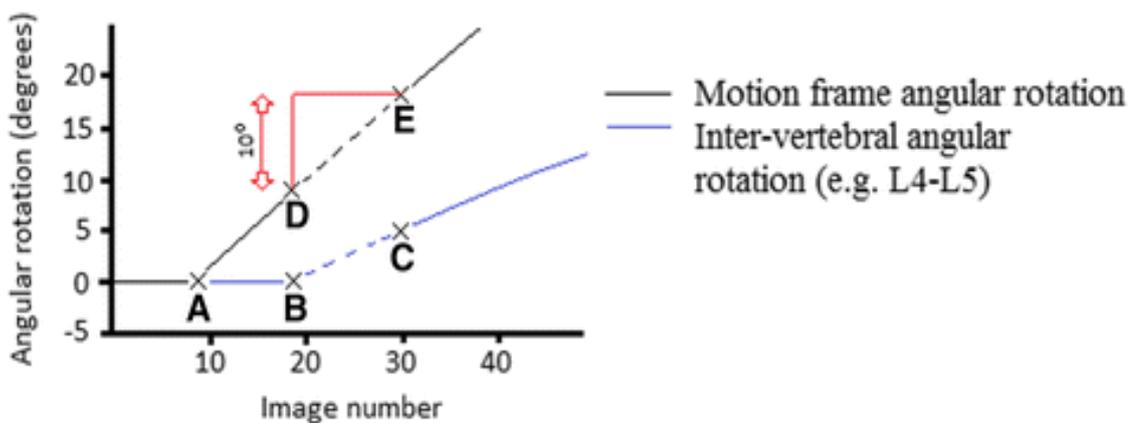


Figure 51: Calculation of initial attainment rate

The dotted lines represent the lines of best fit for motion frame movement (black) and inter-vertebral motion (blue), from which gradients can be calculated. Point at which the motion frame begins movement (A); Point at which inter-vertebral motion begins (B); Dotted line between (B) and (C) = the area under the curve from which the line of best fit is drawn to calculate inter-vertebral movement gradient; Dotted line between (D) and (E) = the area of the curve from which the line of best fit is drawn to calculate the motion frame movement gradient. Initial attainment rate is the calculated as the slope of BC/slope of DE



6.1.3.4 Data analysis

The normality of all data were tested using the Shapiro-Wilk test. Relationships between IV-RoMmax and other biomechanical variables, from normally distributed data were analysed using the Pearson product-moment correlation coefficient, and non-normal data using the Spearman's rank correlation. Any significant relationships (p values < 0.05) were also analysed using simple linear regression. Statistical analysis was performed using IBM SPSS (version 21).

6.1.4 Results

Twenty males with no history of low back pain over the previous year consented to participate. Failed template tracking occurred in 2 participant's sequences, and their data were discarded. The mean (SD) age, height, and body mass Index (BMI) were 27.6 (4.4) years, 1.8 (0.06)m, and 24 (2.2) respectively. Average radiographic exposure factors for the group were 79.7 (5.4)kV and 55.4 (3.4)mA. The mean effective dose was calculated using ICRP103 conversion software PCXMC (Monte Carlo Simulation Package), as 0.143 mSv. A complete motion sequence of the lumbar spine therefore requires less radiation than a single traditional radiograph (Breen et al. 2012). No participants described any fear of pain during the protocol.

The IV-RoMmax (All levels between L2 and S1), Initial attainment rate and lordosis measurements for each participant are shown in tables 18, 19 and 20 respectively. Table 18

shows that the mean IV-RoMmax of levels L2-3, L3-4 and L4-5 are similar (approximately 10°), however the mean IV-RoMmax at L5-S1 is notably smaller (i.e. 6.4°). The standard deviation appears to be higher for inter-vertebral levels in the lower half of the lumbar spine compared to the upper (i.e. >3 for L4-5/L5-S1 and <3 for L2-3/L3-4).

Table 18: Angular range (IV-RoMmax) data

Participant	Angular range (degrees)			
	L2-L3	L3-L4	L4-L5	L5-S1
RS082	14.8	11.8	2.8	1
RS083	11.9	11.6	4.4	6.4
RS084	8	11.7	11.6	6.1
RS085	8.4	11.2	15.3	7.9
RS086	11.3	10.7	8.7	2
RS087	9.1	9.9	13	5.2
RS088	10	11	11.2	5.4
RS089	9.9	11.9	8.8	7.6
RS091	7.4	10.6	7.4	4.6
RS092	4.6	6.4	9.7	9.5
RS093	7.7	10.3	9.7	9.2
RS094	10.4	13.4	6.2	11.4
RS095	4.3	9.3	16.6	0.8
RS096	12	10.2	9.1	5
RS097	5.1	7.7	11	10.5
RS098	7.7	9.7	13.3	9.8
RS099	9.8	8.9	4.1	4.3
RS100	8.3	9.3	14	8.7
Mean	8.9	10.3	9.8	6.4
SD	2.7	1.6	3.9	3.2

Mean initial attainment rate was highest at the level of L2-3, and lowest at L4-5 (Table 19). Standard deviation was highest for measurements at the level of L5-S1.

Table 19: Initial attainment rate data

Participant	Initial attainment rate (ratio)			
	L2-L3	L3-L4	L4-L5	L5-S1
RS082	0.2168	0.0076	0.0425	0.0202
RS083	0.2954	0.2883	0.008	0.1006
RS084	0.3595	0.1779	0.0526	0.0505
RS085	0.1765	0.1304	0.0748	0.0697
RS086	0.1916	0.0429	0.0179	0.1373
RS087	0.2069	0.074	0.132	0.0211
RS088	0.1074	0.0476	0.0444	0.2587
RS089	0.2149	0.0873	0.0163	0.1441
RS091	0.1155	0.0875	0.0797	0.1161
RS092	0.1926	0.2004	0.1565	0.0843
RS093	0.3544	0.1143	0.039	0.0727
RS094	0.1238	0.1768	0.026	0.094
RS095	0.07	0.0344	0.1507	0.0077
RS096	0.2658	0.1519	0.1193	0.1015
RS097	0.1945	0.2239	0.0791	0.13
RS098	0.1675	0.1118	0.0699	0.641
RS099	0.1087	0.0428	0.0404	0.0453
RS100	0.237	0.4892	0.1776	0.3293
Mean	0.1999	0.1382	0.0737	0.1346
SD	0.0815	0.1150	0.0524	0.1498

Lordosis measurements ranged between 34° and 67° with a standard deviation of 9° (Table 20).

Table 20: Lordosis data

Participant	Lordosis (Angle° between L2 and S1)
RS082	56.5
RS083	53.6
RS084	51.3
RS085	52.1
RS086	46
RS087	60.8
RS088	66.6
RS089	61.4
RS091	50.3
RS092	50.2
RS093	52.9
RS094	58
RS095	33.9
RS096	54
RS097	41.8
RS098	44.1
RS099	63.9
RS100	37.6
Mean	51.9
SD	8.84

6.1.4.1 Correlations

A summary of the correlations between all biomechanical variables and IV-RoMmax is given in Table 21. Significant correlations were found between IV-RoMmax and at least one other variable at all inter-vertebral levels. These were consistently of mid-level strength (r - values ranging from -0.64 to 0.73). Lordosis was positively correlated with IV-RoMmax at L2-L3 and negatively with L4-5 ($r = 0.54$ and -0.52 respectively). In terms of IV-RoMmax at one level versus IV-RoMmax at other levels, correlations were found between all levels except L5-S1. L2-L3 range was shown to be positively correlated with that of L3-4, but negatively correlated with L4-5. Initial attainment rate showed examples of strong correlations (both positive and negative) with range at all levels, the strongest being the relationship found between initial attainment rate at L3-4 and L5-S1 IV-RoMmax ($r = 0.73$).

Table 21: Correlations between kinematic variables and IV-RoMmax at all inter-vertebral levels $n = 18$ (Significant relationships are highlighted in bold)

Kinematic variable	L2-L3 IV-RoMmax		L3-L4 IV-RoMmax		L4-L5 IV-RoMmax		L5-S1 IV-RoMmax	
	r	p	r	p	r	p	r	p
Lordosis	0.54	0.021	0.401	0.099	-0.52	0.026	-0.02	0.973
L2-L3 IV-RoMmax	-		0.65	0.003	-0.64	0.004	-0.35	0.157
L3-L4 IV-RoMmax	0.65	0.003	-		-0.29	0.234	-0.12	0.636
L4-L5 IV-RoMmax	-0.64	0.004	-0.29	0.234	-		0.15	0.558
L5-S1 IV-RoMmax	-0.35	0.157	-0.12	0.636	0.15	0.558	-	
Initial attainment rate								
L2-L3	0.20	0.419	0.14	0.58	-0.09	0.713	0.21	0.403
Initial attainment rate								
L3-L4	-0.18	0.465	-0.11	0.668	0.17	0.512	0.73	0.001
Initial attainment rate								
L4-L5	-0.53	0.023	-0.64	0.004	0.59	0.009	0.02	0.949
Initial attainment rate								
L5-S1	0.05	0.852	-0.02	0.938	0.07	0.776	0.42	0.079

Significant relationships are highlighted in bold

6.1.4.2 Simple linear regression analysis

The coefficients of determination (r^2) for each of the significant correlations are shown in Figure 52(A-H). The values range from (0.28 to 0.42) and demonstrate that IV-RoMmax at specific levels, can be influenced by lordosis, the IV-RoMmax at other lumbar levels, and initial attainment rate. Figure 52A for example shows that 41% of the variability in L4-L5 IV-RoMmax can be accounted for by the range of L2-L3 IV-RoMmax.

Figure 52A - 52H: Scatter plots and linear regression values for all significant correlations

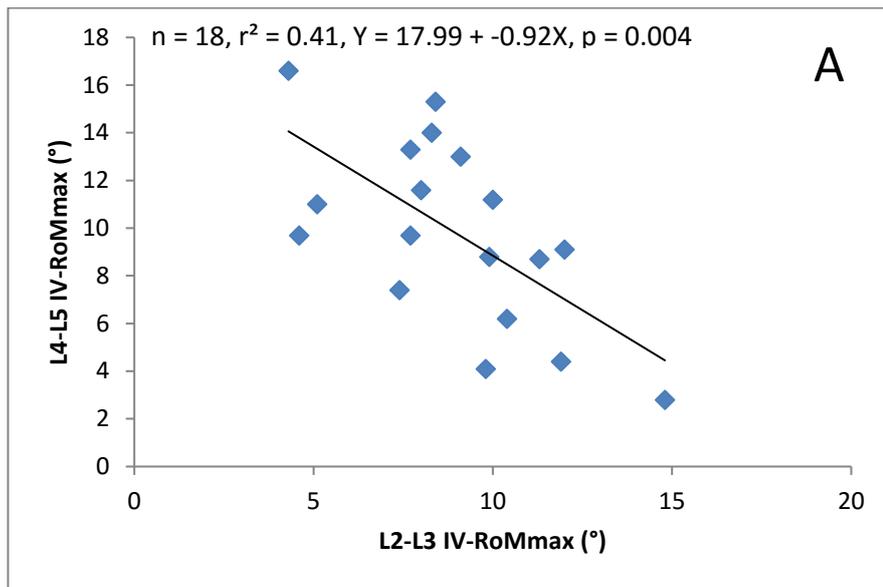


Figure 52A shows the inverse relationship between L2-L3 IV-RoMmax and L4-L5 IV-RoMmax (i.e. as L2-L3 IV-RoMmax increases, L4-L5 IV-RoMmax decreases).

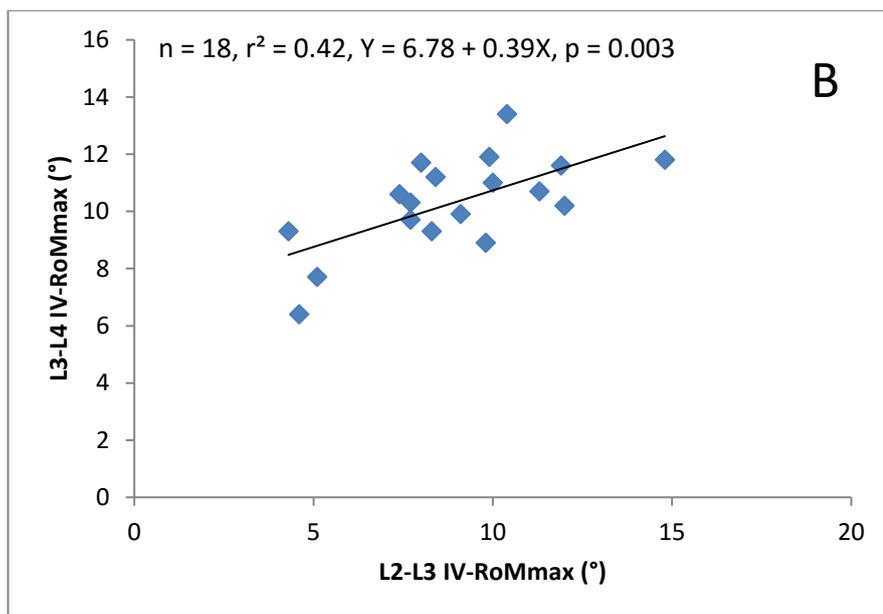


Figure 52B shows the direct relationship between L2-L3 IV-RoMmax and L3-L4 IV-RoMmax (i.e. as L2-L3 IV-RoMmax increases, L3-L4 IV-RoMmax increases).

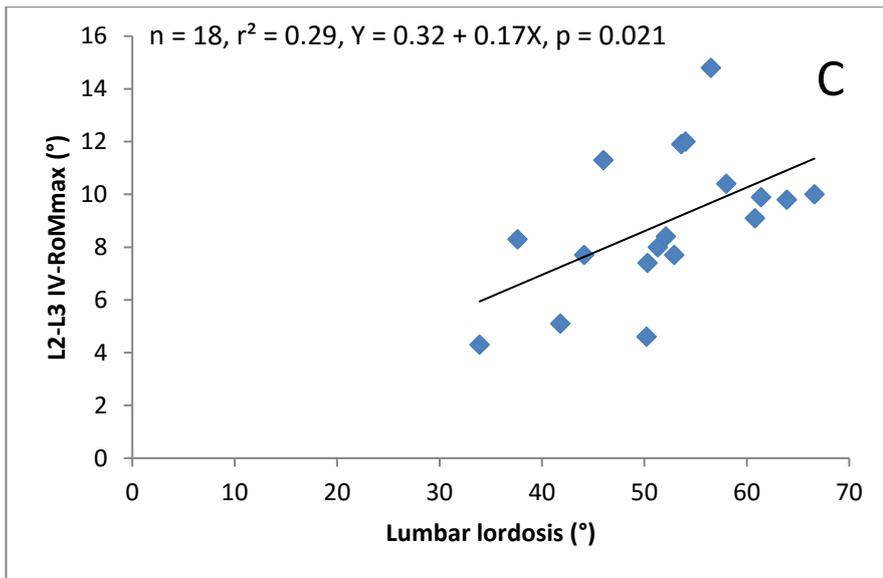


Figure 52C shows the direct relationship between lumbar lordosis (L2-S1) and L2-L3 IV-RoMmax (i.e. as lumbar lordosis increases, L2-L3 IV-RoMmax increases).

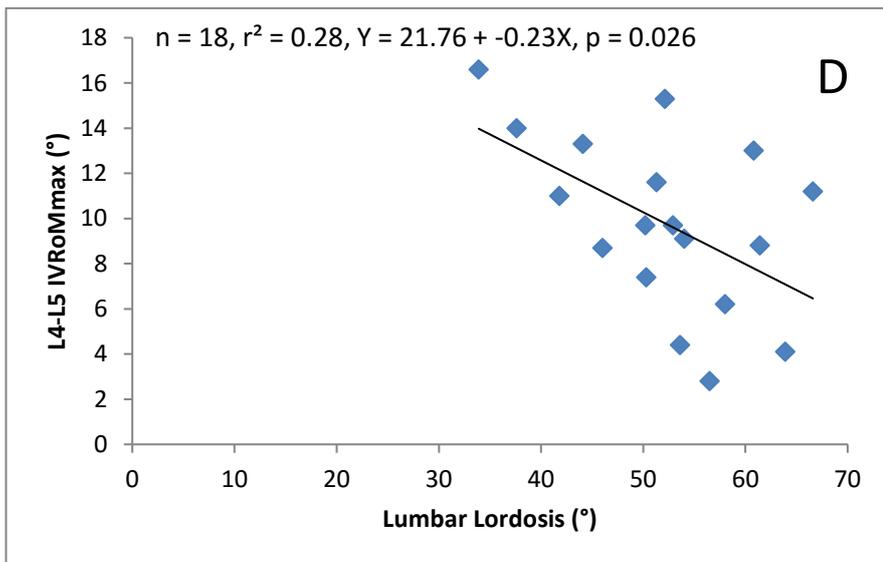


Figure 52D shows the inverse relationship between lumbar lordosis (L2-S1) and L4-L5 IV-RoMmax (i.e. as lumbar lordosis increases, L4-L5 IV-RoMmax decreases).

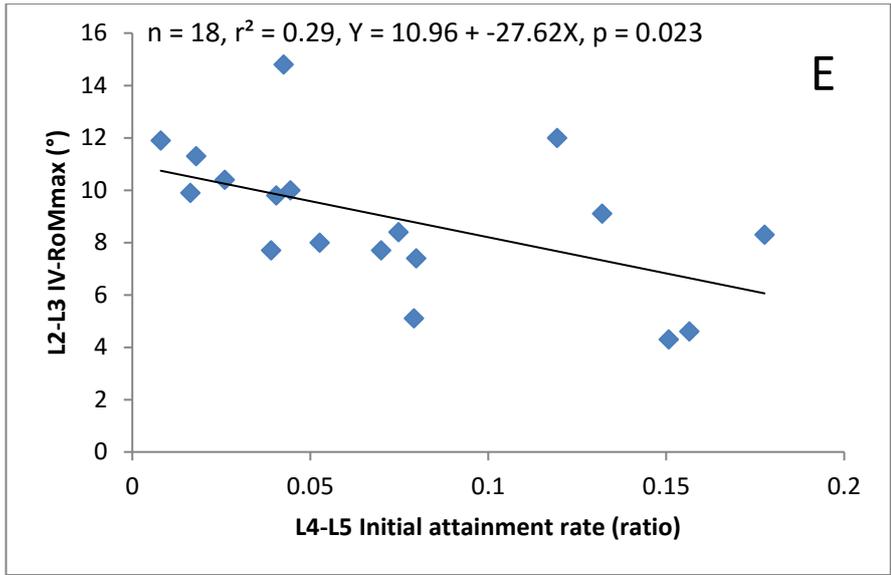


Figure 52E shows the inverse relationship between L4-L5 initial attainment rate and L2-L3 IV-RoMmax (i.e. as L4-L5 initial attainment rate increases, L2-L3 IV-RoMmax decreases).

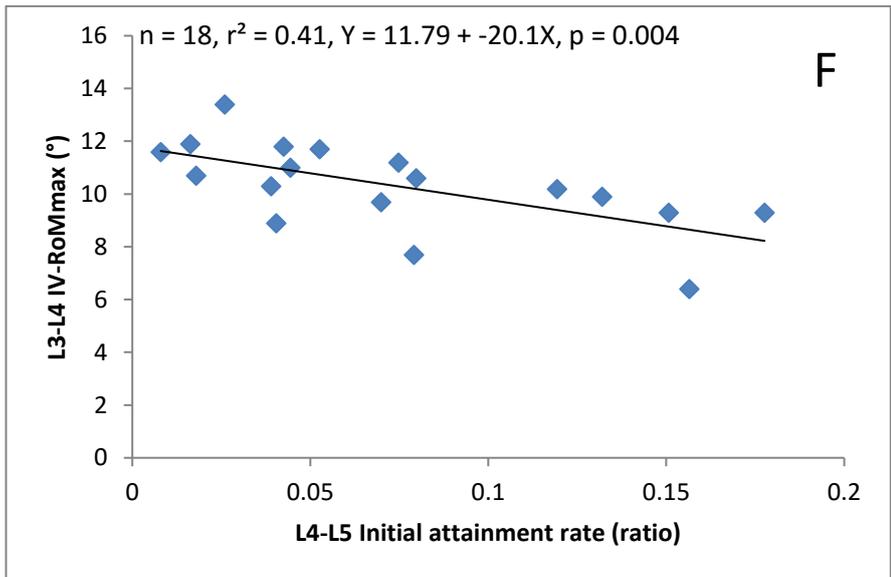


Figure 52F shows the inverse relationship between L4-L5 initial attainment rate and L3-L4 IV-RoMmax (i.e. as L4-L5 initial attainment rate increases, L3-L4 IV-RoMmax decreases).

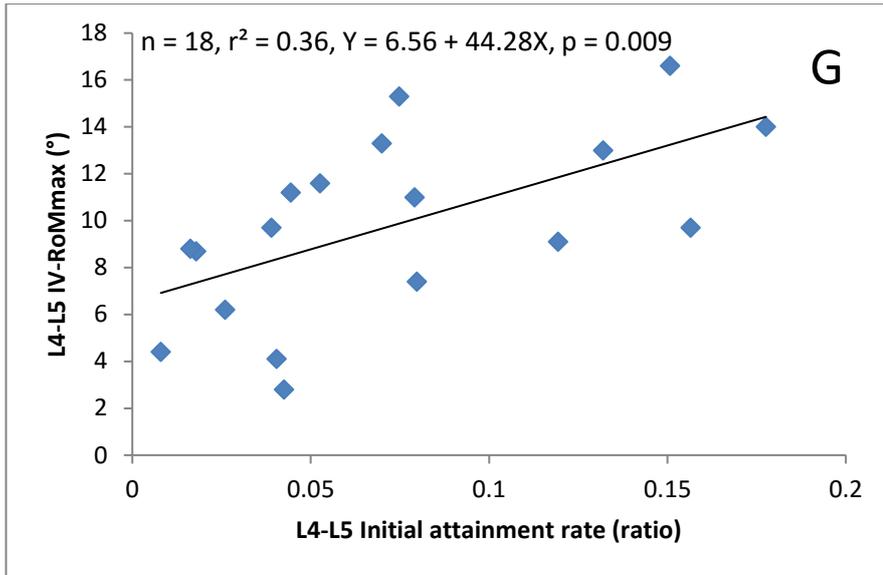


Figure 52G shows the direct relationship between L4-L5 initial attainment rate and L4-L5 IV-RoMmax (i.e. as L4-L5 initial attainment rate increases, L4-L5 IV-RoMmax increases).

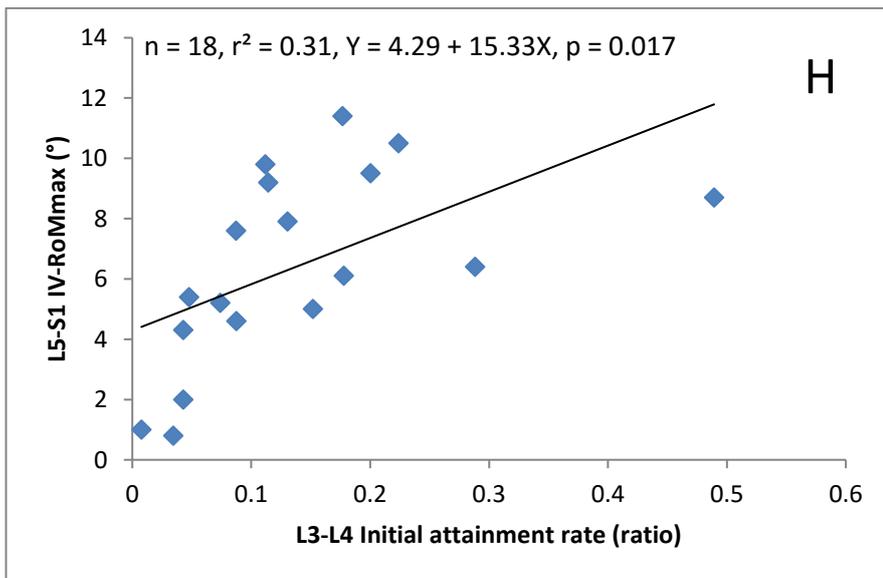


Figure 52H shows the direct relationship between L3-L4 initial attainment rate and L5-S1 IV-RoMmax (i.e. as L3-L4 initial attainment rate increases, L5-S1 IV-RoMmax increases).

Significant relationships between IV-RoMmax and IV-RoMmax at other levels (A and B), lordosis (C and D), and initial attainment rate (E-H). n = sample size, r² = coefficient of determination, Y = linear regression equation, p = p value for the regression coefficient

Several relationships show a fair trend, but do not reach statistical significance. These include a direct relationship between L5-S1 initial attainment rate and L5-S1 IV-RoMmax, and a direct relationship between lordosis and L3-L4 IV-RoMmax (Table 21).

6.1.5 Discussion

The results show evidence of relationships between kinematic variables at multiple levels of the lumbar spine. IV-RoMmax at all inter-vertebral levels was significantly correlated, positively or negatively, with at least one other kinematic or morphological variable, and there appear to be trends in these relationships in terms of the regions of the lumbar spine. The following discussion addresses the latter three hypotheses outlined previously (Section 2.7.2.2), however due to the apparent inter-dependency between lordosis, IV-RoMmax and initial attainment parameters, there is some inevitable overlap between areas.

6.1.5.1 Lordosis vs IV-RoMmax

Hypothesis: There will be a direct relationship between the size of lordosis and IV-RoMmax in the upper lumbar segments, and vice versa

In agreement with the hypothesis, the results suggest that the degree of lordosis has a direct influence on inter-vertebral rotation and that individuals with a relatively larger curvature will move more from the upper lumbar segments (L2-L3 and L3-L4) and those with a flatter lordosis will move more from the lower segment of (L4-5) see table 21 and figure 52C (It should be noted however that L5-S1 IV-RoMmax did not show any relationship with lordosis). This pattern suggests a pivot point at L4, above which individuals with a relatively greater lordosis move initially from the upper segments to flatten the spine, and also subsequently move furthest. This supports the view that a degree of lordosis may allow a more even sharing of motion throughout the lumbar spine, offering a degree of protection to the L4-5 segment during bending (Pavlova et al. 2014), and that lordosis itself has an important role in spinal biomechanical behaviour (Aspden 1989). These findings may have implications for prognosis in patients with L4-5 pain generation, a segment commonly involved in lumbar degeneration (Wu et al. 2014), especially if there is both hypo-lordosis and motion restriction in the upper lumbar spine.

This phenomenon can be visualised by looking at examples of the motion graphs. In Figure 53, the participant is known to have the largest lordosis (67°) in the sample. The graph shows that movement begins at L2-3 and cascades sequentially to L5-S1, and in this instance the sequence of movement appears to relate to the degree of initial lordosis. This is not always the case (see appendix M), and as is becoming more evident, other biomechanical factors are also of influence. It should be noted that in this example the upper lumbar segments (L2-L3 and L3-L4)

actually have a similar IV-RoMmax to L4-L5, and so the inverse relationship observed between the angular ranges of upper (L2-L3) and lower (L4-L5) segments is not apparent in this participant. Figure 54 shows the motion graph of a participant with a relatively hypolordotic curvature, and demonstrates how the movement in this instance appears to initiate at L4-L5, although phase lag between all segments is more difficult to discern. In this example however, the difference between L2-3 and L4-5 IV-RoMmax can be seen.

Figure 53: A participant with a lordosis of 67°. Movement initiates at L2-L3

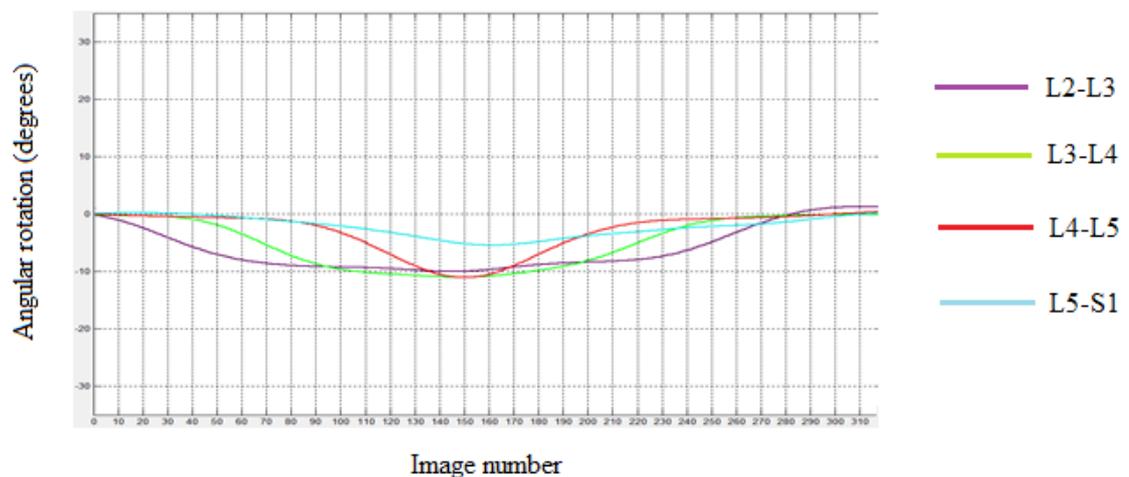
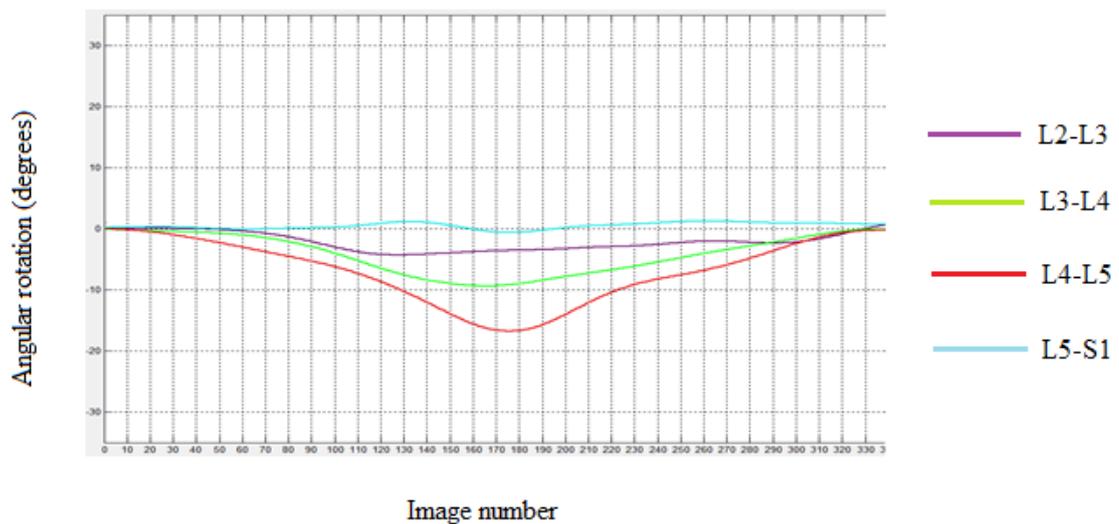


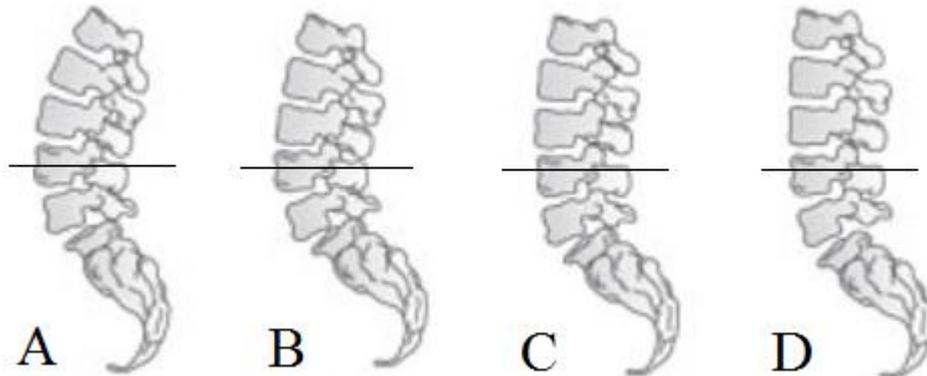
Figure 54: A participant with a lordosis of 34°. Movement initiates at L4-L5



The results suggest therefore that the more lordotic spine will cascade from the upper lumbar vertebrae (and move further), over a pivot point at L4, a segment that typically represents the apex of the curve in both flattened and lordotic lumbar spines (Figure 55). If the individual were to move from below this point then undue stress may be placed on the lower lumbar structures.

In individuals who move more from the lower lumbar spine, there is restraint of the upper levels, a mechanism likely to maintain a degree of sagittal balance (Barrey et al. 2013).

Figure 55: The position of L4 vertebral body in different 4 different types of curvature



Note: A = Hyperlordosis B = High normal lordosis C = Low normal lordosis D = Hypolordosis. The horizontal black line dissects the L4 vertebral body in each type of curvature

This may go some way to explaining why there is contention in the literature regarding patterns of segmental cascade. It is undecided as to whether lumbar segments begin their movement simultaneously (Ahmadi et al. 2009; Lee 2002; Wong 2006; Wong 2004), sequentially (Kanayama 1996) or a mixture of the two (Okawa 1998; Takayanagi 2001), which led Ahmadi et al. (2009) to comment that a 'normal' movement pattern of the lumbar spine during flexion is yet to be determined (Ahmadi et al. 2009). Despite differences between such studies in the interpretation of rotation initiation, the relationships observed between lordosis and IV-RoMmax suggest that more focus should be placed on spinal curvature, as it directly relates to inter-vertebral range, possibly due to the influence of cascade patterns, themselves related to the lordosis. The pattern of movement will also be influenced by how the individual chooses to bend, and so consideration as to how to achieve uniformity between participants (i.e. standardisation of movement) was an important part of this study's design.

6.1.5.2 Implications for stabilisation surgery

If these relationships were reproduced in wider populations, they could have implications for lumbar spinal surgery. For example, if the result of a spinal fusion is to flatten the curvature of the lumbar spine, it is likely that this will place further stress on the lower lumbar levels (i.e. L4-L5), potentially leading to an increased rate of failure at this level (Le Huec et al. 2015). The restriction of movement at L4-L5 may also be compensated for superiorly by an increased proportion of total ROM taken by the upper lumbar, further exacerbating the stress on these levels. This is in agreement with findings in the ASD literature (Lee and Langrana 1984; Untch et

al. 2004; Lee et al. 1988; Xia et al. 2013; Chow et al. 1996; Scannell and McGill 2003), and provides a possible biomechanical explanation for studies that conclude that a diminished lordosis increases the risk of ASD (Rothenfluh et al. 2015). In patients with previous spinal fusions, Rothenfluh et al. (2015) showed that a combination of a high pelvic incidence (pelvic incidence calculated as “the angle between the line perpendicular to the sacral plate at its midpoint and the line connecting this point to the femoral heads axis” (Boulay et al. 2006), and a diminished lordosis, are predisposing factors for the development of ASD (Rothenfluh et al. 2015). If the normal relationships shown in this study are an indication of how behaviours may change with an imposed mechanism of restraint such as a segmental fusion, then it is logical that more stress will be placed on L4-L5 itself, but also the upper lumbar post surgical fusion of L4-L5.

The present study could not investigate pelvic incidence due to an inability to include the femoral head in the x-ray image field, which may be considered a limitation of study design. Future studies may therefore wish to adapt the protocol to incorporate this measurement, and also consider measurements beyond the lumbar spine (i.e. thoracic kinematics) (Claus et al. 2009; Hemming et al. 2015).

6.1.5.3 Inter-segmental versus regional motion studies

The results also suggest that studies that divide the lumbar spine into regions (Dankaerts et al. 2006; Hemming et al. 2015) should consider the normal kinematic behaviour of specific segments. Indeed, whilst L2-L3 and L3-L4 behave in a similar way (Du Rose and Breen 2016b; Kozanek 2009), L4-5 and L5-S1 perhaps need to be considered separately in such studies, or at least use the effective pivot point of L4 as a point of division. This is in agreement with Roussouly et al. (2005), who divided the lumbar lordosis into two arches separated at L4, suggesting that the majority of total lordosis resides between L4-S1, and that the size of this angle influences the segments above (Roussouly et al. 2005). Even this may not be adequate however, as in the example of the participant in Figure 54, L5-S1 shows negligible movement, but L4-5 rotates almost 20 degrees. If these angles are combined as a region it represents a modest contribution from both levels effectively cancelling each other out.

In this study, L5-S1 typically moved the least (mean 6.4° SD 3.2°) and so may be considered as the most restrained segment, perhaps due to specific anatomical adaptations (i.e. the lliolumbar ligaments, facet orientation etc., or as a result of the pelvic restraint protocol) (see also sections 9.11.2 and 9.11.3), and it has been shown that in terms of IV-RoMmax, L4-L5 has an inverse relationship with the levels above. These findings should therefore be considered in studies that

investigate regional lordosis kinematics in relation to LBP. Hemming et al. (2015), used the sub-grouping criteria developed by O'Sullivan et al. (2005) to investigate differences in regional spinal kinematics between flexion pattern (pain provoked during flexion), extension pattern (pain provoked during extension) and healthy controls during functional tasks, including forward bending. Their results showed significant differences between all groups during the forward bending task in upper lumbar and lower thoracic curvature, but this was not the case for the lower lumbar spine (Hemming et al. 2015). Considering the inter-vertebral kinematic findings of the present study, the way in which the spine is divided into regions for such studies may again be of importance. Hemming et al. divided the lumbar spine using the level of L3 as the cut off between upper and lower regions, which may have influenced results as it has been shown that L2-L3 and L3-L4 are directly related to each other, and inversely related to L4-L5 in terms of their angular range. It is feasible therefore that although only one segmental level away, the movements of L3-L4 and L4-L5 may effectively counteract each other. This is perhaps a reason why no differences were found between groups in relation to lower lumbar curvature (Hemming et al. 2015). It is also possible that the differences associated with the upper lumbar spine may be altered if lumbar spinal regions were defined differently (i.e. using L4 and not L3 as a point of division). In another example, in partial agreement with the results of this study, Pavlova et al. (2015) showed that "curvier" individuals (i.e. larger lordosis) tend to have more movement in their upper lumbar regions, but did not find that more kyphotic lumbar spines had greater movement in the lower lumbar (Pavlova et al. 2015). This may again be partly due to the method, as skin markers were used to measure movement of L1-L3 and L3-L5, and so did not specifically include L4-L5.

In contrast to these findings, Dankaerts et al. (2006) who used the same lumbar division system as described by Hemming et al. 2015, found that patients classified into flexion and extension pain provocation groups (O'Sullivan et al. 2005), had respectively kyphotic and lordotic lower lumbar curvatures when measured during sitting (Dankaerts et al. 2006). Therefore, attempts to determine relationships between the kinematics and curvature of regions of the lumbar spine and LBP are of interest, some contradictions have been shown in their outcomes. It could be argued that moving from a sitting based protocol (e.g. Dankaerts et al. 2006) could prevent normal pelvic movement and decrease the influence of the lower limb musculature on lumbar stabilisation, or that biomechanical differences exist between the study groups. It is clear however, that these are areas of research where inter-vertebral information would be valuable, and so should be a consideration for future investigations.

A current limitation of QF inter-vertebral measurements are their restriction to spinal regions such as the lumbar (Durose A and Breen 2016, Du Rose and Breen 2016b, Mellor et al. 2014) and cervical spines (Branney et al. 2015). Whilst typically used to measure regional kinematics, surface marker systems can provide kinematic information from a wider area (e.g. include the thoracic and lumbar regions) (Hemming et al. 2015). The absence of kinematic data collection beyond the lumbar spine is therefore a limitation of this work, and the combined use of QF and surface marker technologies should be considered for future studies.

6.1.5.4 IV-RoMmax vs IV-RoMmax

Hypothesis: There will be an inverse relationship between IV-RoMmax in the upper and IV-RoMmax in the lower lumbar segments

The hypothesis was accepted as L2-L3 and L3-L4 IV-RoMmax were both inversely correlated with the IV-RoMmax of L4-5, however no significant relationships were found between L5-S1 and the upper lumbar segments.

This is the first time that an inverse relationship between the IV-RoMmax of L2-L3 and the IV-RoMmax at L4-5 has been shown, and it suggests a direct compensation mechanism occurring between the two regions. The results also show that the IV-RoMmax of L2-3 and L3-4 were strongly positively correlated, suggesting that they tend to work in tandem. This was in agreement with Kozanek et al. (2009), who showed that the behaviour (in terms of IV-RoM) of L2-3 and L3-4 were similar, and that this was different to that of the lower lumbar spine (specifically L4-L5), however they attributed these movement patterns to facet orientation (Kozanek 2009), which provides a feasible structural explanation for the patterns observed. The combined effect of increased upper lumbar rotation is most pronounced in more lordotic lumbar spines, and will be reflected in the motor control of these segments, i.e. a strategy that allows movement in the upper lumbar, but restricts movement at L4-L5.

If it is accepted that instability results from reduced restraint, then it may be suggested that reduced motion at these upper levels could promote relative L4-5 instability as a consequence of motion stress transfer. The reverse of this pattern has been shown in spinal surgery patients, as when stiffness is induced in the lower segments via fusion, there is increased mobility in superior segments (Lee and Langrana 1984, Untch et al. 2004, Lee et al. 1988, Xia et al. 2013, Chow et al. 1996, Scannell and McGill 2003). However, this has never been shown to be an adaptive mechanism in healthy controls. This suggests that individuals with specific biomechanical features may be pre-disposed to increased stresses through the lower lumbar segments (especially L4-L5). The results suggest that the L4-L5 motion segment behaves

differently during flexion than the rest of the lumbar spine, and is consequently an important segment for further research. It has also been shown however, that relationships exist between L4-L5 IV-RoMmax and kinematic parameters at other spinal levels, and so L4-L5 should perhaps not be considered in isolation.

Taghipour-Darzi et al. (2012) suggest that inter-vertebral rotation information, particularly end of range information does not typically provide useful information regarding diagnosis of instability. They do however suggest that mid-range kinematic measurements may be important (Taghipour-Darzi et al. 2012). Indeed their study reports that when using the criteria developed by Hicks et al. (2005) to determine sub-groups of NSLBP patients with instability (Hicks 2005), the L4-5 motion segment in this group was actually hypomobile during mid-range flexion in their segmental instability group relative to the same level in healthy controls. This finding appears contradictory to what should be expected in a lumbar segmental instability (LSI) group, and the authors suggest that the hypomobility may be due to a restriction mechanism caused by a muscular reflex adaptation to the patient's pain (Panjabi et al. 1994). However, given the findings in the present study, it is also feasible that the lack of motion at this level may in fact be a compensation mechanism for relative hypermobility at segments elsewhere in the lumbar spine, or a relatively high mean lordotic curvature within the sample (Du Rose and Breen 2016b). Indeed, the Hicks (2005) criteria are somewhat generic, in that the positive prone instability test used is not level specific, and therefore hypermobility may actually have been expected to be found at levels other than L4-L5. In addition, although the Taghipour et al. (2012) protocol does measure mid-range rotation, it does not take continuous measurements throughout the cycle, and if some form of vertebral cascade is assumed (Kanayama 1996) (which would be expected in a group with large lordosis), it is possible that the true maximum angular range will be missed, as they are reached at different stages of the bend. It is also a consideration therefore, that the population diagnosed with LSI, have pre-existing biomechanical behaviours that predispose them to less movement at L4-5. As discovered in this study, these may include those with a larger lordosis, those that have increased relative movement in the upper lumbar (i.e. L2-L3), and those with minimal laxity at the L4-L5 segment. This highlights an example of where replicating a previous study (e.g. Taghipour-Darzi et al. 2012) using continuous inter-vertebral data measurements would be of interest.

6.1.5.5 Initial attainment rate (laxity) and IV-RoMmax

Hypothesis: There will be a direct relationship between initial attainment rate and IV-RoMmax at the same level

This relationship was not found throughout every level of the lumbar spine, however a moderate and significant direct relationship was found between the initial attainment rate and the IV-RoMmax of L4-L5 (Table 21), and so the hypothesis was accepted.

Hodges et al. (2013) suggested that a lax segment will be associated with a relative increase in IV-RoM at the same level (Hodges et al. 2013). Despite this being the case for L4-5, it was not the case for other levels, although there was also a strong positive relationship between L3-L4 initial attainment rate and L5-S1 IV-RoMmax. The reasons for this relationship are not clear, however as no other parameters (i.e. lordosis or IV-RoMmax) were significantly correlated with L5-S1 IV-RoMmax, in this case muscle activity may be more directly influential. It has been shown that L2-3 and L3-4 behave in a similar way in terms of angular range, and therefore in terms of overall stability, muscular control strategies may be required to counteract the combined effect of increased laxity and the associated increased range occurring simultaneously at two levels. This, left unchecked, would result in excessive movement in the upper lumbar region, and may therefore be prevented by other mechanisms (see chapter 9 further analysis: Initial attainment rate versus muscle activity changes). It is possible that laxity at L3-4 relating to an increase in IV-RoM at L5-S1 is a compensation mechanism for a lack of movement at L4-5, however this cannot be substantiated. Indeed, it is difficult to find research that supports or opposes these speculations, as there are few *in vivo* inter-vertebral kinematic studies in healthy controls to compare with.

6.1.5.6 An intra-operative comparison

The relationship between the IV-RoM at L4-5 and the initial attainment rate at L4-5 suggests that increased range relates to an increased segmental laxity, and therefore decreased stiffness. In a novel study that used an intra-operative system to determine stiffness and neutral zone measurements from load deformation data, Hasegawa et al. (2009) showed that unstable segments (i.e. segments with degenerative spondylolisthesis) have reduced stiffness and larger neutral zones compared to healthy motion segments (Hasegawa et al. 2009). The study did not however find a significant relationship between IV-RoM and stiffness or the neutral zone ($r^2 = 0.021$ $P = 0.336$ and $r^2 = 0.000$ $P = 0.988$) respectively. A criticism of Hasegawa's study, is that it pools data from multiple levels, which considering the present study's results may not be appropriate. It has been shown here that initial attainment rate (laxity) has different relationships with IV-RoMmax dependent on the specific motion segment measured, and therefore pooling of results from such levels may lead to inaccurate conclusions. Hasegawa et al's findings are in contrast to those of this study (for the level of L4-L5), and may be explained by the lack of level specific information. It should also be noted that the degree of disc

degeneration within their study sample may have been confounding. According to Kirkaldy-Willis and Farfan (1982), disc degeneration will progress from normal, to dysfunctional, to unstable and eventually to a restabilisation phase (Kirkaldy-Willis and Farfan 1982), and Hasegawa et al. state that the degenerative cases in their study were a mixture between those between unstable and restabilisation phases (Kirkaldy-Willis and Farfan 1982). Assuming this to be the case then a sample of both more and less mobile segments will counteract the influence of each other in terms of range, perhaps making its measurement questionable in this instance.

6.1.5.7 Compensation by adjacent segments

Conversely, while attainment rate and IV-RoMmax at L4-5 were positively correlated, L4-5 attainment rate was negatively correlated with the IV-RoMmax at L2-3 and with L3-4 above (Table 21). As both attainment rate and IV-RoMmax are expressions of intervertebral restraint, these relationships can be regarded as compensatory, contributing to the attenuation of stress throughout the lumbar spine linkages. Thus there are indications of interactions and effects between kinematic and morphological variables at different levels. This is of importance in terms of musculoskeletal modelling, as traditionally, parameter changes have been modelled in a uniform manner throughout the lumbar spine. In a recent study, Putzer et al. (2016) modelled the effect of changes in ligament stiffness on lumbar inter-vertebral movement, and showed that a uniform increase in ligament stiffness throughout the lumbar spine results in increased loading and movement in the lower lumbar segments (Putzer et al. 2016). The current study has shown however that inter-vertebral stiffness is not uniform throughout the lumbar spine, and that there are apparent compensation mechanisms to such parameter changes. Indeed, in terms of the purported risk of increased lumbar ligament stiffness to lower lumbar structures (Putzer et al. 2016), it has been shown that a healthy lumbar spine may adapt to an increase in stiffness (i.e. decreased initial attainment rate) in the upper lumbar (i.e. L2-L3 and L3-L4), by a decrease in stiffness at L4-L5 (Du Rose and Breen 2016b). This level of information will therefore at some stage need to be incorporated into musculoskeletal models of the spine.

Note: Several other relationships approached significance and may therefore also be important. L5-S1 IV-RoMmax and its initial attainment rate was positively correlated, suggesting that typically, if lower lumbar segments are lax, they will move further. This was not found for the upper segments. L3-L4 IV-RoMmax was also directly related to lordosis (i.e. the same relationship as L2-L3 IV-RoMmax with lordosis), the lack of significance possibly due to the typically more neutral starting position of this segment.

6.1.6 Limitations

The study's results are only representative of a small, young, healthy, male population and replication with larger and more extensive populations would be required to explore the relationships in wider age groups and in females. In light of this, any discussions relating to the investigation and management of wider LBP populations warrant careful consideration. Furthermore, it was also not possible to address the impact of loading on spinal behaviour, although every effort was made to standardise the population sample and study protocol for body mass index. In this research all measurements were made during weight-bearing, and therefore the effect of muscle activity is also a consideration. Chapter 7 examines the relationships between lumbar paraspinal muscle activity and the kinematic and morphological variables described here (Du Rose and Breen 2016a). Future studies may also wish to consider the use of dynamic stereo x-ray imaging (Aiyangar et al. 2014), especially if investigation of rotation in the transverse or coronal planes is required, where associated out of plane movements are more prominent.

6.1.7 Conclusions

Significant correlations were found between IV-RoMmax, IV-RoMmax at different inter-vertebral levels, lordosis and initial attainment rate, and the study demonstrated weak to moderate effects of these variables on IV-RoMmax. There is an increasing awareness of the importance of sagittal parameters when planning surgical strategy, correcting sagittal balance, or when considering more conservative treatment options (Barrey and Darnis 2015; Doulgeris et al. 2015), therefore the ability to accurately assess and measure sagittal kinematic and morphological parameters is important, as we attempt to understand their potential clinical utility (Mehta et al. 2012). The existence of intrinsic links between morphological variables such as lordosis have been described before (Roussouly and Pinheiro-Franco 2011), however we are the first to use continuous in vivo inter-vertebral motion to investigate its links with IV-RoMmax and initial attainment rate. These results provide clues as to what may happen when kinematic or morphological changes are imposed through conservative treatment or surgery, both as local and regional effects. The apparent inter-dependency may assist in building rationales for treatments, and highlights the need to account for factors such as lordosis when conducting kinematic studies. If the results are re-affirmed by multivariate investigations in larger samples, future longitudinal studies are recommended to investigate the effect of interventions in low back pain populations, that have been informed by the relationships described in this study. It should be noted however that this was an investigation into normal biomechanical behaviour,

and no comparisons were made with a clinical population. Therefore the study findings cannot be translated directly into clinical practice in relation to NSLBP groups.

There is evidence to suggest that the protocol used in this study may produce kinematic patterns that are different to what might be expected when free-bending. It should be re-iterated however, that when comparing individuals, standardisation is an essential part of study design, and the protocol developed within this study provides an acceptable method for doing so.

“If detailed and standardised measures of spinal posture could be applied in studies of posture behaviour, the potential to compare and combine data from multiple studies (i.e. meta-analysis) would be greatly improved. Such standardisation and meta-analysis would provide foundation for conclusive determination of relationships between posture and pain” (Claus et al. 2016).

The use of inter-vertebral kinematic measurements is one area where such standardisation could feasibly be achieved. This would not only benefit kinematic research fields, but when such standardisation is applied in combination with motor control investigations, this would progress the field of spinal control research in general. This study has demonstrated how kinematic and morphological parameters can influence the restraint of motion segments during forward bending, but this has been done in the absence of information about the active control system (Panjabi 1992a; Panjabi 1992b). Relationships between muscle activity and IV-RoMmax are investigated in the next chapter.

Chapter 7

7.1 Relationships between paraspinal muscle activity and lumbar inter-vertebral range of motion

7.1.1 Introduction

As per chapter 6, the bulk of the following chapter is a peer reviewed paper, this time published in the journal *Healthcare* (Du Rose and Breen 2016a) by the author and this study's lead supervisor¹². The paper is also not presented in its entirety as the agreement and reliability findings were also presented in the chapter 5. In the general discussion section there have also been substantial additions, and again there are some methodological details that replicate previous aspects of the thesis, although the majority is referenced to previous chapters to avoid unnecessary duplication. The paper addresses the primary thesis aim (Section 2.7.2.3) by investigating the relationships between lumbar inter-vertebral motion and lumbar spinal muscle electrical activity in healthy adults during standardised weight-bearing forward bending.

7.1.2 Background

Optimal control of the spine during voluntary trunk bending requires fine-tuned coordination of numerous trunk muscles (Reeves et al. 2007). This dynamic control is believed to be modulated by communication between three sub-systems, the passive (vertebrae, discs, and ligaments), the active (muscles and tendons), and the control (central nervous system and nerves) systems (Panjabi 1992a; Panjabi 1992b). Investigating the interplay between sub-systems however is difficult, as the spine is a complex structure; and a hidden kinematic chain. Several different technologies are therefore typically required, each with their own limitations.

In order to directly investigate the passive and active sub-systems of the spine, there have been many efforts to concurrently measure spinal kinematics and muscle activity (Sanchez-Zuriaga et al. 2012; Kim et al. 2013; Claus et al. 2009; Hashemirad et al. 2009; Burnett et al. 2004; McGill et al. 1997; Kaigle et al. 1998; Callaghan et al. 1998; Peach et al. 1998; Dankaerts et al. 2009). The majority of these studies have used surface electromyography combined with skin surface kinematic measurement techniques such as Fastrak (Burnett et al. 2004;

¹² Professor Alan Breen

Dankaerts et al. 2009), Isotrak (McGill et al. 1997; Callaghan et al. 1998), or cameras (Sanchez-Zuriaga et al. 2015; Kim et al. 2013; Hashemirad et al. 2009). These are typically limited to the investigation of gross spinal motion. To include segmental data usually requires invasive techniques such as the surgical insertion of intra-osseous pins. In this way Kaigle et al. (1998) investigated the reduction in lumbar muscular activity during full flexion (flexion relaxation) and spinal kinematics at an inter-vertebral level (Kaigle et al. 1998). However, typically only single motion segments were considered, and EMG was also only recorded from one level (e.g., lumbar longissimus thoracis) (Kaigle et al. 1998).

7.1.2.1 Contemporaneous monitoring of inter-vertebral passive and active systems

Study of the integrated function of the joints and muscles of the spine requires contemporaneous multi-level kinematic and electromyographic monitoring throughout the motion. This is necessary to incorporate timing, magnitude, and segmentation in the two systems to characterise control. Multi-level surface electromyography fulfils these requirements for muscle activity and quantitative fluoroscopy measures a range of continuous inter-vertebral motion variables (Breen et al. 2012). Contemporaneous recording of these measures therefore provides an integrated assessment of the passive and active systems of the spine, and it is proposed that this may be useful when assessing patients with low back pain (LBP) (Sanchez-Zuriaga et al. 2015; D'hooge et al. 2013). This study therefore deployed quantitative fluoroscopy (QF), and surface electromyography (sEMG) of the lumbar spine together for the first time. The study investigated the biomechanics of the lumbar spine in a healthy control population in order to potentially better understand the significance of biomechanical changes in LBP populations.

7.1.2.2 Variable selection

In order to investigate relationships between segmental kinematics and local muscle activity, suitable variables from each must be identified. While responses to perturbation (Hodges et al. 2009), and the flexion relaxation phenomenon (an absence of paraspinal muscle activity during full sagittal flexion (FRP)) have been investigated (Luhring et al. 2015; McGorry and Lin 2012), few studies have included sEMG amplitude changes throughout the cycle, be they increases or decreases. This study therefore addressed these parameters. QF measures continuous intervertebral rotation and translation in the coronal and sagittal planes during weight-bearing or recumbent motion and can also extrapolate the instant axis of rotation (IAR) and rotational range attainment rate from this. However, the need to also compare intervertebral range of motion (IV-RoM) with sEMG in the present studies, dictates the need

for continuous motion information. Therefore IAR rotation and attainment rate were not likely to be so useful. In addition, the small ranges of translation make this measure unsuitable for numerical comparisons, leaving maximum rotational motion as the preferred measure.

To investigate the relationships between lumbar muscle activity and inter-vertebral restraint during bending requires access to the maximum IV-RoM (IV-RoMmax). Continuous intervertebral rotation data allows both temporal comparisons with other variables and the actual maximum IV-RoM (IV-RoMmax), rather than IV-RoM at the limit of voluntary trunk bending, to be extracted. Recording in the standing orientation allows these comparisons.

7.1.2.3 Enhanced functional assessment

Sanchez-Zuriaga et al. (2015) suggested that whilst an increase in ES EMG activity during flexion was observed in LBP groups when compared to healthy controls, no difference was found in gross lumbar ROM (Sanchez-Zuriaga et al. 2015). This would suggest that either muscle activity has no effect on the range of motion, or that the detail of what is happening at individual levels is being missed. For example it may be that when there is an increase in paraspinal activity in LBP patients during flexion, but no difference in RoM, the share of RoM may have shifted between levels at different stages in the motion. Indeed, the primary role of the paraspinal muscle during flexion is to resist inter-vertebral motion (Bogduk 2012) and so it may be that the motion is restricted at a specific level, and compensated for elsewhere, be this at other lumbar levels, or in the thoracic spine or pelvis. In support of Bogduk (2012), *in vitro* experiments have also shown that increased multifidus activity decreases the range of inter-vertebral motion (Wilke et al. 1995), and an increase in the activity of locally acting paraspinals relative to the globally acting (i.e. an increased LES/TES ratio) is purported as a spinal stiffening strategy (Van Dieen et al. 2003). Further insight is however required at an inter-vertebral level *in vivo*. This study therefore investigates the relationships between paraspinal muscle activity and lumbar spinal kinematics in healthy controls.

7.1.3 Aim of the study

The purpose of this study was to quantify the relationships between IV-RoMmax during flexion of the lumbar spine and the accompanying paraspinal muscle activity.

7.1.3.1 Specific objectives

To determine whether mean lumbar paraspinal sEMG amplitudes are related to the IV-RoMmax at lumbar inter-vertebral levels.

To determine whether ratios of inter-level lumbar paraspinal sEMG amplitudes are related to the IV-RoMmax at lumbar inter-vertebral levels.

To determine whether changes in lumbar paraspinal sEMG amplitudes during different phases of the forward bending cycle are related to IV-RoMmax at lumbar inter-vertebral levels.

In particular it was hypothesised that

- A. There will be an inverse relationship between muscle activity and the IV-RoMmax
- B. There will be an inverse relationship between the LES/TES ratio and IV-RoMmax
- C. There will be a direct relationship between the size of muscle deactivation and IV-RoMmax

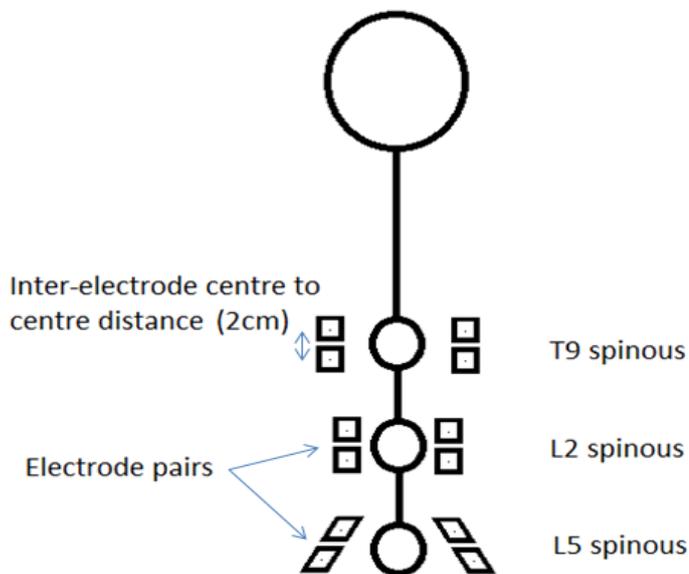
7.1.4 Methods

Although this chapter represents a different publication to chapter 6, the data collection for both was performed concurrently. Therefore the methodological details are exactly the same as the previous chapter (please see section 6.1.3), with the addition of the electromyography protocol outlined below. The kinematic data collection and processing section was therefore removed to avoid unnecessary duplication. The methodology relating to the sEMG data collection however was retained.

7.1.4.1 Electromyography

Prior to the commencement of the weight-bearing data collection (see section 6.1.3.3), participants lay prone in order for 12 electrode sites to be marked on their backs with a skin pencil. In preparation for this, the skin over their lower backs was prepared for sEMG electrode application by light abrasion, cleaning with an alcohol swab, and when necessary, shaving of the area. Disposable pre-gelled self-adhesive Ag-AgCl electrodes were then applied over three bilateral muscle groups with a 20 mm centre-to-centre inter-electrode distance as follows: Thoracic erector spinae (TES) (5 cm lateral to the T9 spinous process) (Peach et al. 1998; Nelson-Wong and Callaghan 2010), the lumbar erector spinae (LES), and lumbar multifidus (LMU) (2 cm lateral to the L2 and L5 spinous processes) (McGorry and Lin 2012; O'Shaughnessy et al. 2013) whilst the participant was in slight flexion (Figure 56).

Figure 56: Electrode positioning sites. (Note: T9 spinous refers to the spinous process of the ninth thoracic vertebra, L2 to the second lumbar vertebra and L5 to the fifth lumbar vertebra.)

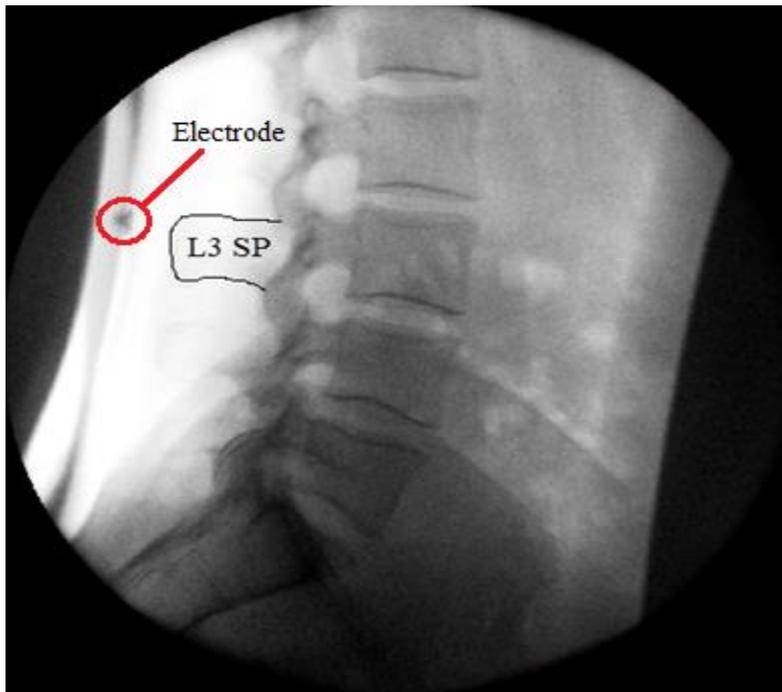


Although cross talk from multiple muscles will inevitably contribute to the signal recorded at each electrode site, cross-sections of the spine at each electrode site showed that the muscles that will predominate at T9 (TES) and L2 (LES) is longissimus thoracis, and at L5 (LMU) multifidus. Three Biopac wireless transmitters (Bionomadix Dual Channel Wireless EMG) were then placed on the lower back attached by self-adhesive Velcro pads. There was no significant difference between the normalised mean sEMG amplitudes recorded over left and right sides during the flexion and return cycle. Therefore, an average of the mean amplitudes from both sides was used for all analysis (D’hooge et al. 2013).

7.1.4.2 Electrode positioning accuracy

Electrode application accuracy is dependent on the subjective identification of bony anatomical landmarks, and current methods used are therefore limited by human subjectivity and variation in individual anatomy (Kim et al. 2007; Chin et al. 2006; Billis et al. 2003; Chakraverty et al. 2007). It has been suggested however that accuracy can be improved significantly when techniques are combined (Merz et al. 2013). This investigation was integrated into a larger ongoing normative database study, which required recumbent QF imaging before weight-bearing imaging commenced. In order to improve electrode positioning accuracy, an electrode was placed over the spinous process of L3 during the recumbent protocol. This provided an improved anatomical reference point for the application of the electrodes (Figure 57).

Figure 57: An electrode placed over the spinous process of L3 (duplication of figure 42)



7.1.4.3 The sEMG equipment set-up and signal processing

The sEMG signal data were recorded at a sampling rate of 2000 Hz using a common-mode rejection ratio (CMRR) of 110 dB and an input impedance of 1000 MOhms. The six signals were band pass filtered at 10–500 Hz and full wave rectified. The root mean square (RMS) amplitude was calculated for individual participant cycles and normalised during post-processing to sub-maximal voluntary contractions expressed as a percentage of the sMVC.

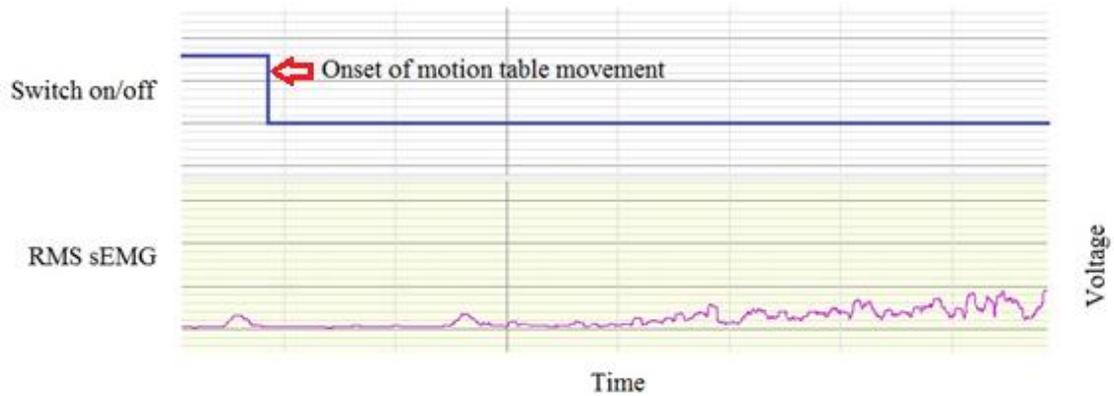
7.1.4.4 The reference contraction

When data collection had been completed, and in order to provide a sub-maximal reference contraction (sMVC) (Demoulin et al. 2006), participants were asked to lie prone on a padded bench with their hands behind their head. They were then required to raise their torso off the couch and hold this position for five seconds whilst their legs and pelvis were stabilised. This process was repeated three times and the average sMVC was used as a reference. This technique was selected over a normalisation to a peak, primarily due to the even loading of the investigated muscle groups, but also to avoid the problem of variations in participant's muscle activation patterns in order to produce the same movement.

7.1.4.5 Synchronisation

The QF motion frame controller recording and the sEMG data recording were co-ordinated using a trip switch attached to the motion arm of the frame. This registered a data point on the sEMG timeline (Figure 58).

Figure 58: Synchronisation of the motion frame movement and sEMG recordings



7.1.4.6 Fear of movement

It has been shown that sEMG measurements can be affected by psychological factors, such as a fear of movement (kinesiophobia) and re-injury (Geisser et al. 2005; Vlaeyen 1999), which have also been linked to reduced lumbar spinal movements (Geisser et al. 2004; Thomas and France 2008). Whilst these were not expected to factor in a healthy population group, they nevertheless required consideration in the study design. Therefore the final part of the study protocol was for the participant to answer the following question “Were you fearful of the forward bending and return phases, or the sMVC causing you low back pain?” If any participants described a fear of pain during any of the movements required of them, then their data would be discarded.

7.1.5 Data analysis

sEMG ratios (Van Dieen et al. 2003; Reeves et al. 2006) were calculated from the mean left-right normalised sEMG (RMS) amplitudes during the flexion phase only as follows, LMU/LES, LES/TES and LMU/TES. In order to calculate sEMG changes at different stages of the flexion cycle, the forward bending phase was divided into five epochs for each participant (D’Hooge et al. 2013; Dankaerts et al. 2009). As the entire flexion cycle was approximately 10 seconds, each epoch represented about 2 seconds of EMG signal recording and 12° of motion frame movement. The change in mean sEMG between epochs was then calculated (e.g., the change

during the early stage of flexion was calculated as (epoch 1–2) for each of TES, LES, and LMU). This was repeated to determine changes between all epochs at all levels.

All data were tested for normality using the Shapiro-Wilk test. Relationships between IV-RoMmax and sEMG ratios and changes from normally distributed data were analysed using the Pearson product-moment correlation coefficient, and non-normal data using the Spearman’s Rank Correlation. Significant relationships (p values < 0.05) were further analysed using simple linear regression. Statistical analysis was performed using IBM SPSS (version 21).

7.1.6 Results

Descriptions of the study population and radiation exposure factors have been outlined previously (section 6.1.4). As failed template tracking occurred in 2 participant’s sequences, all their data (including sEMG data) were discarded. Mean normalised RMS sEMG during the flexion cycle ranged between 3% and 21% for the TES, 2% and 31% for the LES and 13% and 40% for the LMU (Table 22). No participants reported any fear of movement.

Table 22: Mean normalised sEMG amplitude during entire flexion phase of cycle

Participant	Mean normalised sEMG amplitude (% of sMVC)		
	TES	LES	LMU
RS082	11.05	17.49	30.09
RS083	3.38	7.67	19.61
RS084	5.56	3.06	23.31
RS085	11.67	8.17	21.59
RS086	6.03	7.73	21.27
RS087	6.71	6.26	21.76
RS088	2.58	3.54	17.58
RS089	14.52	8.31	18.17
RS091	21.16	30.80	35.96
RS092	6.35	15.74	39.51
RS093	6.24	6.82	18.50
RS094	6.08	8.89	32.18
RS095	8.49	13.07	18.91
RS096	11.58	9.73	17.15
RS097	5.27	2.10	13.06
RS098	11.39	9.44	27.20
RS099	9.32	24.24	34.73
RS100	4.30	4.13	13.83
Mean	8.43	10.40	23.58
SD	4.57	7.52	7.85

7.1.6.1 A general description of sEMG activation patterns for LMU, LES and TES during the forward bending cycle

Typically the activity of LMU increased upon initiation of forward flexion before reaching a peak typically in epochs 3 or 4 of the flexion cycle. This is shown in the data with decreasing negative mean values (representing a relative increase in amplitude in the latter epoch) for epochs 1-2, 2-3 and 3-4. The positive mean amplitude change between epochs 4-5 represents a relative decrease in amplitude (never to the point of complete deactivation) over the latter stages of the flexion movement (Table 23). There were exceptions to this rule however, as some participants demonstrated a continued rise in activity to the full 60° of flexion. This general pattern was mirrored by the activity of LES, except the peak was reached earlier (typically in epochs 2 or 3) and the relative size of normalised RMS sEMG activity was always smaller than LMU. This pattern is shown by the negative mean value between early epochs, and positive mean values between the latter two (Table 24). The TES signal was typically the smallest (in terms of normalised amplitude) of the 3 muscles examined, and demonstrated the smallest changes. There was a notable trend however for an increase in TES activity that would begin between epochs 3-5 and continue to the full 60° of flexion. This again can be seen in the negative mean values seen over the latter epochs, representative of an increase in TES activity over this period of the cycle (Table 25).

Table 23: LMU sEMG amplitude changes throughout the flexion cycle

Participant	Normalised EMG of LMU (epoch 1-epoch2)	Normalised EMG of LMU (epoch 2-epoch3)	Normalised EMG of LMU (epoch 3-epoch4)	Normalised EMG of LMU (epoch 4-epoch5)
RS082	-11.67	-8.48	-4.20	0.04
RS083	-7.14	-1.54	0.21	0.32
RS084	-5.44	-2.68	-1.36	1.17
RS085	-5.74	-6.25	-1.30	9.52
RS086	-7.99	-4.70	-2.15	0.72
RS087	-7.13	-6.32	-3.10	-4.93
RS088	-7.24	-1.92	1.84	-0.86
RS089	-6.41	-1.70	0.99	0.35
RS091	-8.48	-7.63	-7.71	-5.73
RS092	-20.90	-21.08	-7.64	1.90
RS093	-7.93	-1.02	0.87	2.34
RS094	-11.65	-8.16	-3.11	3.02
RS095	-6.77	-4.43	-0.59	-0.28
RS096	-4.93	-5.44	-5.58	-6.60
RS097	-5.83	-2.90	-2.07	-2.35
RS098	-6.28	-5.72	5.42	13.78
RS099	-8.31	-6.89	-5.29	-5.46
RS100	-7.21	-0.42	5.63	9.26
Mean	-8.17	-5.40	-1.62	0.90
SD	3.67	4.67	3.80	5.47

Table 24: LES sEMG amplitude changes throughout the flexion cycle

Participant	Normalised EMG Of LES (epoch 1- epoch 2)	Normalised EMG at of LES (epoch 2- epoch 3)	Normalised EMG of LES (epoch 3- epoch 4)	Normalised EMG of LES (epoch 4- epoch 5)
RS082	-7.38	-4.16	2.35	9.36
RS083	-3.88	4.10	3.95	1.20
RS084	-0.11	-0.12	-0.24	-0.21
RS085	-3.58	2.47	6.75	1.45
RS086	-3.70	0.97	3.41	2.69
RS087	-3.31	-2.21	4.54	0.61
RS088	-0.06	2.66	0.71	-0.18
RS089	-4.27	-0.67	4.15	1.90
RS091	-6.08	-4.93	-2.44	-2.00
RS092	-15.38	-2.67	7.93	3.79
RS093	-2.49	4.08	2.12	0.27
RS094	-1.88	2.16	2.19	2.17
RS095	-2.55	-1.26	1.78	2.51
RS096	-5.17	-2.56	-1.52	0.38
RS097	-0.52	0.25	0.14	0.18
RS098	-1.04	6.09	6.12	0.58
RS099	-8.59	-4.89	-0.40	-0.35
RS100	-0.50	-0.16	0.17	0.23
Mean	-3.92	-0.05	2.32	1.37
SD	3.77	3.21	2.88	2.41

Table 25: TES sEMG amplitude changes throughout the flexion cycle

Participant	Normalised EMG Of TES (epoch 1- epoch 2)	Normalised EMG of TES (epoch 2- epoch 3)	Normalised EMG of TES (epoch 3- epoch 4)	Normalised EMG of TES (epoch 4- epoch 5)
RS082	1.14	1.09	0.31	-3.29
RS083	-0.34	0.19	0.15	-0.52
RS084	-1.39	-1.84	-1.05	-1.81
RS085	1.02	1.87	-3.70	-4.02
RS086	-1.32	2.63	-0.98	-1.60
RS087	0.42	3.42	0.08	-1.89
RS088	-0.05	0.05	0.12	-0.29
RS089	0.01	0.76	-1.60	-4.24
RS091	-2.68	2.19	2.84	-2.46
RS092	-1.03	-0.43	-0.17	-1.90
RS093	-0.98	-1.73	-1.94	-2.19
RS094	-0.08	-0.07	-0.01	0.01
RS095	-2.93	2.39	2.94	0.29
RS096	-2.87	-0.14	-0.01	-0.92
RS097	-1.00	-0.55	-0.72	-0.55
RS098	-0.50	2.75	0.07	-2.51
RS099	-0.70	2.70	-0.43	0.65
RS100	-0.68	-1.54	-0.77	-2.16
Mean	-0.78	0.76	-0.27	-1.63
SD	1.18	1.68	1.51	1.41

The patterns in the data can also be clearly visualised in the recorded sEMG signals. Figures 59A and 59B, represents a recording from a participant's LES, and shows the typical paraspinal activation pattern for a forward bending and return exercise. The flexion phase (i.e. the first ten seconds) shows an increase in amplitude that diminishes towards the end of the bend, followed by a larger increase in activity as the trunk is raised back to the standing position (i.e. the last ten seconds). It should be noted that in no participant was the FRP demonstrated, this was most likely due to the standardised range of 60° preventing full sagittal flexion.

Figures 59A and 59B: Examples of raw (Figure 59A) and rectified and filtered (Figure 59B) LES sEMG signal recorded during entire flexion and return cycle

Figure 59A

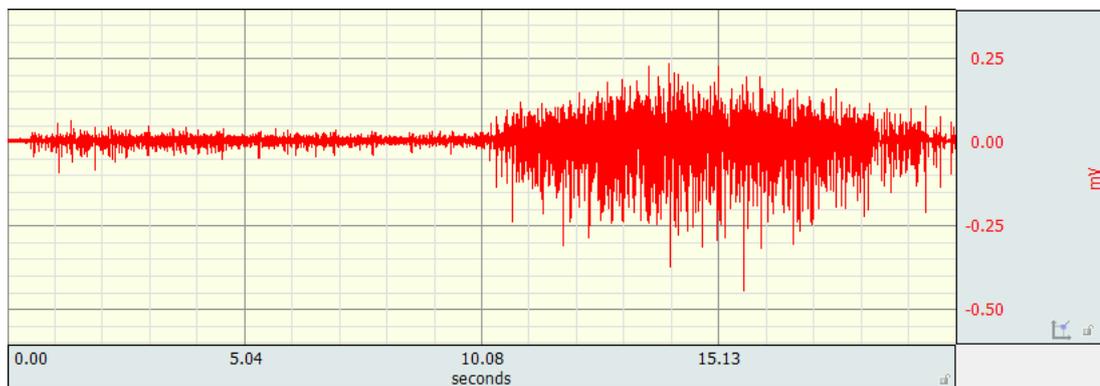
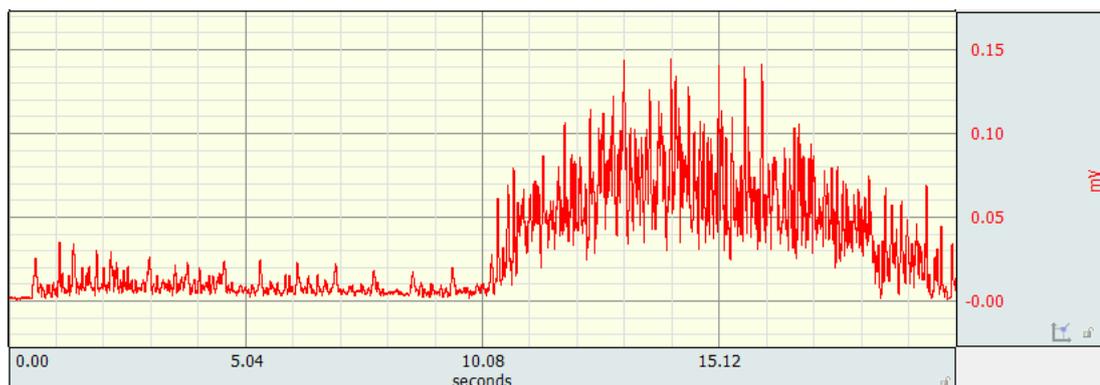


Figure 59B



Figures 60A and 60B show an example of how the activity of LMU gradually increases during flexion, until a point in the latter stages of the bend, at which deactivation begins.

Figures 60A and 60B: Examples of raw (Figure 60A) and rectified and filtered (Figure 60B) LMU sEMG signal recorded during flexion

Figure 60A

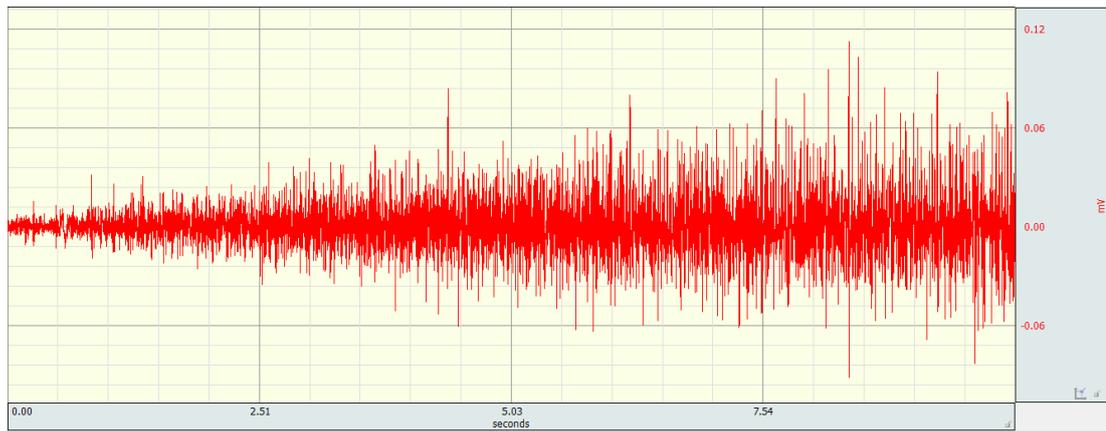
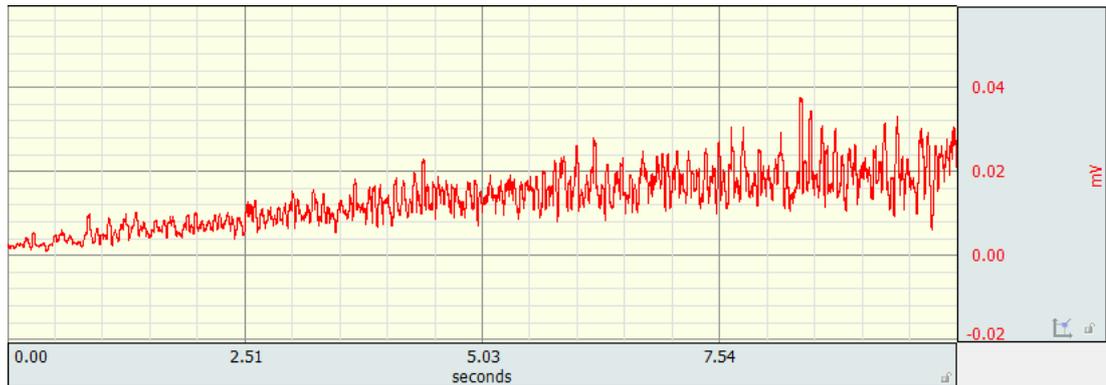


Figure 60B



The slightly earlier peak typically demonstrated by the LES muscle can be seen in the example shown in figures 61A and 61B. It should be noted that again, although myoelectric deactivation begins during the flexion cycle, complete deactivation does not occur.

Figures 61A and 61B: Examples of raw (Figure 61A) and rectified and filtered (Figure 61B) LES sEMG signal recorded during flexion

Figure 61A

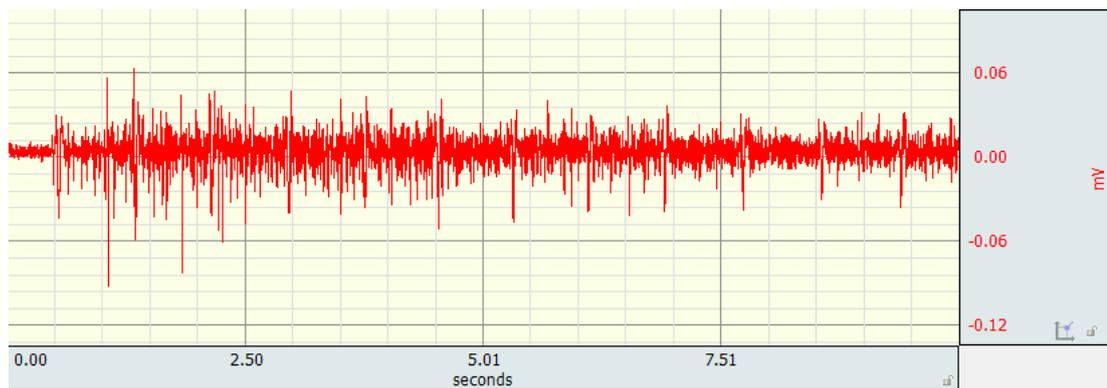
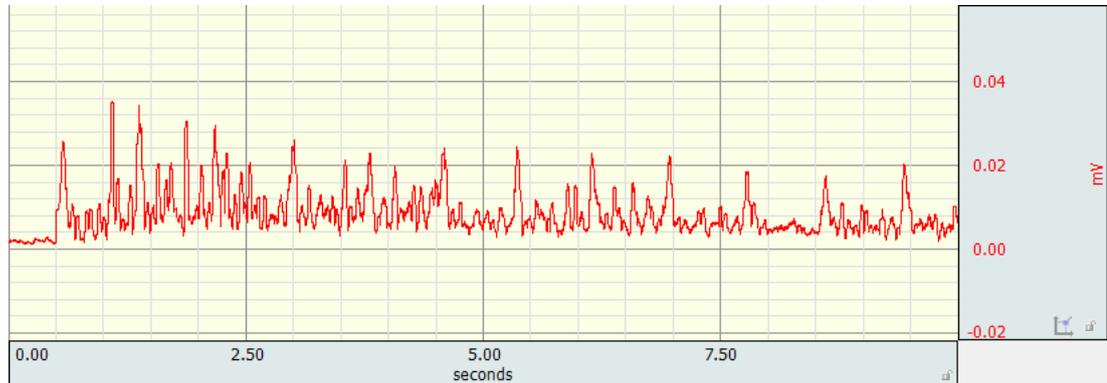


Figure 61B



Finally, figures 62A and 62B show an example of TES activity during the flexion phase of the cycle. It is clear from the signal in this example, that although limited activity occurs during the majority of the bend, activity does begin to increase during the latter stages.

Figures 62A and 62B: Examples of raw (Figure 62A) and rectified and filtered (Figure 62B) TES sEMG signal recorded during flexion

Figure 62A

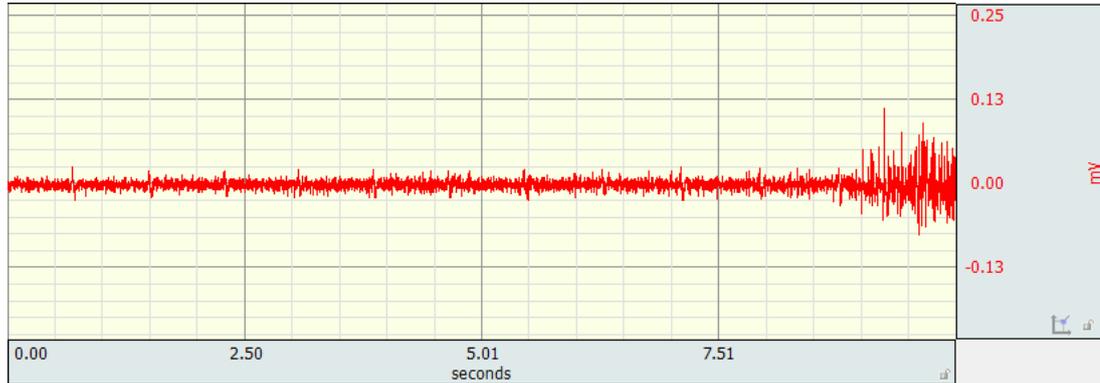
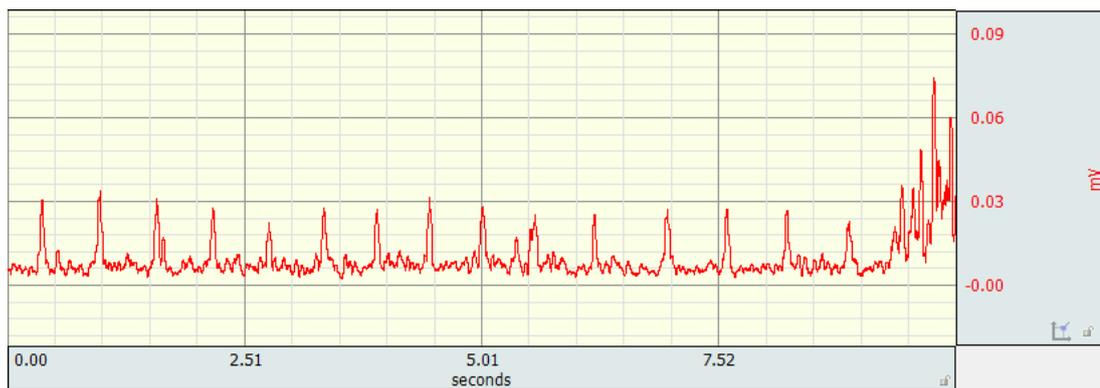


Figure 62B



7.1.6.2 Correlations between mean muscle activity and IV-RoMmax

No significant relationships were found between the mean sEMG amplitude of any muscle and the IV-RoMmax at any level (Table 26). There was a trend shown however, in that LES and LMU activity was inversely related to the IV-RoMmax of L4-L5.

Table 26: Correlations between mean normalised sEMG (% of sMVC) across the entire flexion cycle and IV-RoMmax at all inter-vertebral levels (n = 18)

Mean sEMG amplitude (% of sMVC)		Inter-vertebral level			
		L2-L3	L3-L4	L4-L5	L5-S1
TES	r	0.24	0.119	-0.116	-0.198
	p	0.926	0.639	0.647	0.430
LES	r	0.022	-0.120	-0.448	-0.371
	p	0.931	0.630	0.062	0.129
LMU	r	-0.014	-0.080	-0.455	-0.028
	p	0.955	0.751	0.058	0.912

7.1.6.3 Correlations between Muscle Activity Changes and IV-RoMmax

A summary of all correlations between changes in muscle activity and IV-RoMmax is given in (Table 27). Significant correlations were only found with lower lumbar segmental motion (L4-5 and L5-S1). These were consistently of mid-level strength (r -values ranging from -0.48 to 0.59), and include inter-vertebral relationships with all three muscle levels. The results also demonstrate a number of correlations that approach significance; these did include relationships with motion at upper inter-vertebral lumbar levels (L2-3 and L3-4). All significant correlations were further analysed using simple linear regression. The effects of muscle activity changes on IV-RoMmax are shown in (Table 28). The table shows that r^2 values range from 0.177 to 0.247 .

Table 27: Correlations* between muscle activity changes (three groups, five epochs) and IV-RoMmax at all inter-vertebral levels (n = 18)

		Inter-vertebral level			
Muscle activity change		L2-L3	L3-L4	L4-L5	L5-S1
TES epoch 1-2	r	0.404	0.316	-0.164	0.224
	p	0.097	0.201	0.516	0.371
TES epoch 2-3	r	0.083	-0.02	0.036	-0.477
	p	0.743	0.938	0.888	0.045
TES epoch 3-4*	r	-0.059	-0.077	-0.171	-0.434
	p	0.817	0.760	0.496	0.072
TES epoch 4-5	r	-0.124	-0.194	-0.134	-0.103
	p	0.625	0.441	0.596	0.683
LES epoch 1-2*	r	-0.203	0.070	0.595	0.391
	p	0.418	0.782	0.009	0.108
LES epoch 2-3	r	-0.045	0.257	0.295	0.497
	p	0.86	0.303	0.234	0.036
LES epoch 3-4	r	-0.117	-0.118	0.211	0.266
	p	0.645	0.642	0.4	0.286
LES epoch 4-5*	r	0.228	0.215	-0.088	-0.055
	p	0.362	0.392	0.729	0.829
LMU epoch 1-2	r	0.14	0.334	0.314	-0.144
	p	0.58	0.176	0.204	0.567
LMU epoch 2-3*	r	0.021	0.062	0.317	0.139
	p	0.935	0.807	0.200	0.581
LMU epoch 3-4	r	-0.039	0.164	0.455	0.273
	p	0.877	0.517	0.058	0.272
LMU epoch 4-5	r	-0.159	0.067	0.429	0.461
	p	0.53	0.793	0.076	0.027

Significant correlations are highlighted in bold italic. Correlations that approach significance are highlighted in bold. * Indicates a row that includes non-parametric data and therefore a Spearman's Rank Correlation was used. All other normally distributed data was analysed using the Pearson product-moment correlation coefficient. r = correlation co-efficient, p = p-value (95% confidence level). Note: A negative correlation relates to a relative increase in muscle activity between epochs.

Of particular note is that changes in LMU, LES and TES at different stages of the cycle, can all influence the IV-RoMmax of L5-S1. The significant correlations indicate that a decrease in

LMU activity during the final stages of sagittal flexion relates to an increase in L5-S1 IV-RoMmax, as does a decrease in mid-cycle LES activity, and an increase in mid-cycle TES activity. A decrease in early cycle LES activity also relates to an increase in L4-L5 IV-RoMmax.

Table 28: Simple linear regression analysis: significant correlations

Variable	Inter-vertebral			
	level	r	p	r ²
LMU Epoch 4-5	L5-S1	0.461	0.027	0.212
LES Epoch 2-3	L5-S1	0.497	0.036	0.247
TES Epoch 2-3	L5-S1	-0.477	0.045	0.227
LES Epoch 1-2*	L4-5	0.595	0.009	0.177

* Indicates a row that includes non-parametric data and therefore a Spearman's Rank Correlation was used. All other normally distributed data was analysed using the Pearson product-moment correlation coefficient. r = correlation co-efficient, p = p-value and r² = the co-efficient of determination. Note: A negative correlation relates to a relative increase in muscle activity between epochs.

7.1.6.4 Correlations between sEMG Ratios and IV-RoMmax

The sEMG ratio data for all three muscle ratios, showed similar mean and SD values for LMU/TES and LMU/LES, however the SD for the LES/TES values was notably smaller (Table 29). The correlations between sEMG ratios and IV-RoMmax at all inter-vertebral levels are also shown (Table 30). The only significant relationship was found between the ratio of LES/TES and the IV-RoMmax at L4-5, and is demonstrated by the scatter plot in (Figure 63). This plot highlights the negative correlation between the LES/TES ratio and L4-L5 IV-RoMmax, and shows that when the muscle activity of the LES increases relative to that of the TES, there is a decrease in the IV-RoMmax at L4-L5. The only other correlation to approach significance was between LMU/LES ratio and the IV-RoMmax at L5-S1 (r = 0.37, p = 0.13).

Table 29: sEMG ratio data for LMU/TES, LMU/LES and LES/TES during flexion phase of cycle

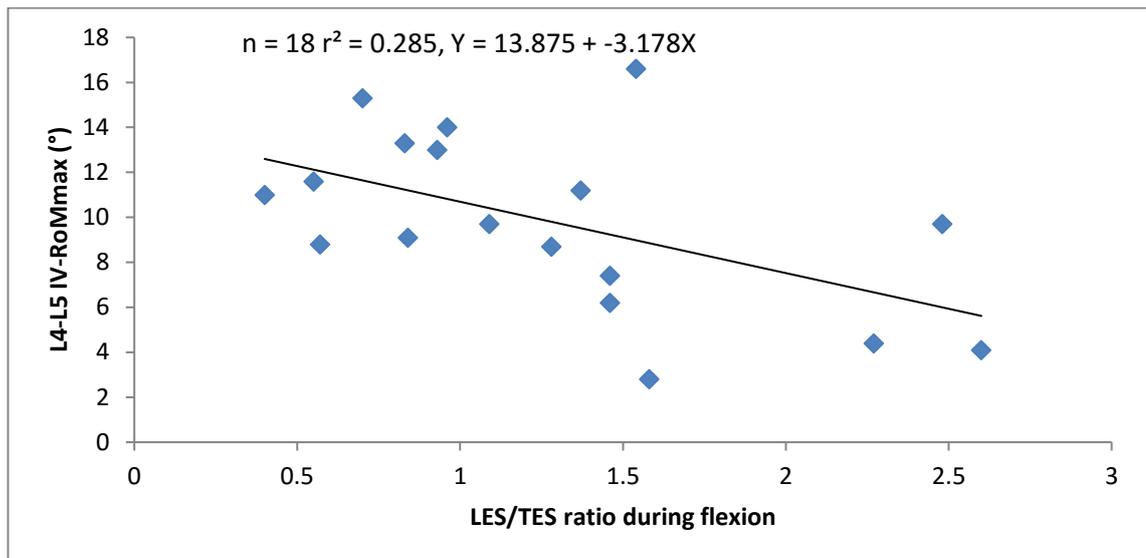
Participant	sEMG ratios		
	LMU/TES	LMU/LES	LES/TES
RS082	2.72	1.72	1.58
RS083	5.81	2.56	2.27
RS084	4.19	7.61	0.55
RS085	1.85	2.64	0.70
RS086	3.53	2.75	1.28
RS087	3.24	3.47	0.93
RS088	6.82	4.96	1.37
RS089	1.25	2.19	0.57
RS091	1.70	1.17	1.46
RS092	6.22	2.51	2.48
RS093	2.96	2.71	1.09
RS094	5.30	3.62	1.46
RS095	2.23	1.45	1.54
RS096	1.48	1.76	0.84
RS097	2.48	6.23	0.40
RS098	2.39	2.88	0.83
RS099	3.72	1.43	2.60
RS100	3.21	3.35	0.96
Mean	3.39	3.06	1.27
SD	1.67	1.70	0.65

Table 30: Correlations between muscle activity ratios and IV-RoMmax at all inter-vertebral levels (n = 18)

Ratio		Inter-vertebral level			
		L2-L3	L3-L4	L4-L5	L5-S1
LMU/TES	r	0.046	-0.013	-0.236	0.152
	p	0.856	0.958	0.345	0.548
LMU/LES	r	-0.209	0.04	0.263	0.37
	p	0.405	0.875	0.292	0.13
LES/TES	r	0.095	-0.217	-0.533	-0.242
	p	0.708	0.387	0.023	0.333

r = the Pearson product-moment correlation coefficient, p = p-value. Note: A negative correlation relates to a relative increase in muscle activity ratio.

Figure 63: The relationship between the LES/TES ratio and the IV-RoMmax at L4-5



Relationship between LES/TES ratio and IV-RoMmax at L4-L5. n = sample size, r^2 = coefficient of determination, Y = linear regression equation

In summary, although there were trends between mean LES and LMU activity over the entire flexion cycle and L4-L5 IV-RoMmax, no significant relationships were found. Significant correlations were found however between the LES/TES ratio and L4-L5 IV-RoMmax, and between sEMG activity changes and the IV-RoMmax of lower lumbar levels. These findings are discussed in following section.

7.1.7 Discussion

The primary aim of this thesis was to investigate the relationships that exist between lumbar inter-vertebral motion and lumbar spinal muscle electrical activity in healthy adults during standardised weight-bearing forward bending. A visual analysis of muscle activation patterns (Appendix L) suggested that activity of locally acting LMU and LES muscles whilst beginning simultaneously, reach a point at which deactivation can begin to occur at different points of the cycle. TES on the other hand was usually still increasing in activity during the latter stages, suggestive of a possible compensatory mechanism for the decrease in LES and LMU activity (Andersson et al. 1996).

Although no significant relationships were found between the normalised mean RMS sEMG of LMU, LES or TES during the entire flexion cycle and IV-RoMmax at any lumbar level, analysis did reveal a significant relationship between the ratio of LES/TES and the IV-RoMmax of L4-L5, and several significant relationships between changes in normalised RMS sEMG at

specific stages of the flexion cycle, and the IV-RoMmax of lower lumbar levels (i.e. L4-L5 and L5-S1). These findings and their interpretation are discussed further below.

7.1.7.1 Relationships between mean muscle activity amplitudes over the entire flexion cycle and IV-RoMmax

In a study comparing paraspinal EMG between LBP patients and healthy controls, Ahern et al. (1988), concluded that significantly lower muscle activity observed in the LBP group was due to the reduced degree of flexion achieved in the group (Ahern et al. 1988). This is contrary to the conclusions of other studies which suggest that an increase in paraspinal EMG in patients is a stabilisation strategy that effectively limits their range of movement (Sanchez-Zuriaga et al. 2015; Sihvonen et al. 1991; Kuriyama et al. 2005). The Ahern et al. (1988) study only recorded EMG from the local lumbar paraspinal musculature however, so their findings could perhaps be explained by unrecorded compensatory activity of globally acting muscles, or adaptations in movement patterns beyond the lumbar spine. On balance therefore, it was anticipated that muscle activity (especially of muscles located anatomically close to the motion segment involved i.e. LES and L2-L3 or LMU and L5-S1) would relate inversely to the amount of angular rotation at these segments, and the following hypothesis was formulated.

Sub hypothesis

- There will be an inverse relationship between muscle activity and the IV-RoMmax

No significant relationships were found however (Table 26), and so the hypothesis was rejected. Therefore within the spectrum of normal physiological inter-vertebral movement during flexion, mean TES, LES and LMU activity throughout the flexion cycle was not shown to significantly influence angular range at any inter-vertebral level. This is in agreement with others findings (Reeves et al. 2007), and suggests a large degree of complexity in terms of segmental control in the lumbar region, and that the co-ordination of multiple muscles is likely required in the control of healthy motion segments. The potential for such relationships cannot be dismissed completely however, as only three muscle groups (all paraspinal muscles) were investigated, and the correlations between LES and LMU activity and L4-L5 IV-RoMmax did approach significance (Table 26). Future investigations using a larger sample size or investigating other trunk muscles are therefore warranted. The analysis of muscle activity ratios, and muscle activity changes during different stages (epochs) of the flexion cycle did however reveal significant relationships, which are discussed below.

7.1.7.2 Relationships between muscle activity ratios and IV-RoMmax

It is suggested that in order to maintain the functional stability of the spine, there needs to be an interplay between the local and global muscles (Hodges and Moseley 2003), and this study's results provides a degree of evidence to support this statement. Of particular relevance was the ratio of LES/TES, which was shown to have a statistically significant negative relationship with the range of motion at L4-L5 (Figure 63 and Table 30).

This ratio of lumbar erector spinae over thoracic erector spinae activity has been investigated in several previous studies (Van Dieen et al. 2003; Reeves et al. 2006; Cholewicki and McGill 1996; Van den Hoorn et al. 2012). In a musculoskeletal trunk model based on the EMG data collected from two healthy participants during various dynamic tasks, Cholewicki and McGill (1996) suggested that preferential recruitment of the LES over the TES may be a strategy to increase spinal stiffness (Cholewicki and McGill 1996). A further study comparing the muscle recruitment patterns in healthy controls to those of LBP patients, found higher LES/TES ratios in the latter (Van Dieen et al. 2003), which led to the conclusion that the differences found between groups were likely to be an adaptation designed to enhance spinal stability. This theory was further supported by Van Den Hoorn et al. (2012), who also demonstrated a significantly higher LES/TES ratio in LBP patients during gait (Van Den Hoorn et al. 2012). Reeves et al. (2006) also investigated this muscle activation imbalance in varsity athletes, and while maintaining that there was indeed a relationship between muscle imbalance between levels and LBP, also found that in some individuals with a history of LBP, TES activity could be dominant (Reeves et al. 2006). The authors contend that this may be explained by pathology, e.g. the CNS optimising activation to minimise compression, or by a difference in muscle fibre types between groups in order to compensate for fatigue related pain (Reeves et al. 2009). Interestingly, and with ramifications in terms of this study's conclusions, they also discuss the possibility of the reported activity patterns being the result of different types of posture or lordosis, and that further studies may account for this effect.

On balance however, the weight of the previous literature suggested that an increase in the LES/TES ratio acts to increase stability in the lumbar spine, and the following hypothesis was formulated.

Sub hypothesis

- There will be an inverse relationship between the LES/TES ratio and IV-RoMmax

Whilst the results of this study highlight that the ratio of LES/TES can vary in a population with no long term history of low back pain, such variations would appear to relate to differences in the inter-vertebral mechanics in such a population. A significant inverse relationship between the LES/TES ratio and the IV-RoMmax of L4-5 was discovered, and so the hypothesis was accepted. The relationship was not significant at other levels however, which would suggest that the coordinated activity of local and global muscles (from the electrodes sites selected) have an influence over L4-L5 IV-RoMmax specifically.

In a population of young and healthy adult males therefore, it has been shown that an increase in the ratio of LES/TES provides increased restraint at the level of L4-L5. It could be argued that L4-L5 is of particular clinical importance, as it is frequently cited as a suspected pain generator, and a segment commonly targeted for surgical fusion (Le Huec et al. 2015). The fact that this ratio appears to affect the stiffness of this particular segment in the absence of pain is therefore of importance, as it provides a possible biomechanical reason for the conflicted findings of previous studies (Van Dieen et al. 2003; Reeves et al. 2006; Lariviere et al. 2000; Van Den Hoorn 2012), and should therefore be considered in addition to theories of motor control responses to pain or dysfunction. For example, as discussed in the literature review, studies that suggest that an increase in LES activity relative to TES is a strategy to increase stiffness in LBP groups (Van Dieen et al. 2003), should also consider that individuals with a high LES/TES ratio may either have a muscle recruitment strategy designed to stabilise a specific segment (in this case L4-L5), or that there are a high proportion of individuals within the sample with these normally occurring biomechanical behaviours (i.e. a smaller IV-RoMmax at L4-L5 in association with an increased LES/TES ratio, that is not associated with the LBP).

This study has consistently suggested that the IV-RoMmax of a single inter-vertebral segment should not be considered in isolation, due to the demonstration of interactions with biomechanical elements elsewhere in the lumbar spine. Therefore, whilst a high LES/TES ratio restrains movement at L4-L5, it has also been shown that movement at this level is inversely correlated with movement in the upper lumbar (L2-L3 and L3-L4), and so it would be logical that a smaller LES/TES ratio (i.e. relatively more TES and less LES activity) would relate to restricted movements in these segments. No such relationships were found however, indicative of different stabilisation strategies relating to the upper lumbar levels.

7.1.7.3 Fibre types and sizes

When considering the possible roles (i.e. stabilisers, movement initiators or mechanisms for sensory feedback) of globally (TES) and locally (LES and LMU) acting muscles, a further consideration is the fibre type composition at these levels. In terms of both longissimus and multifidus, it has been shown that fibres of both muscles are larger in the thoracic spine than the lumbar (Mannion et al. 1997). This study found that TES typically contributed the least (compared to other paraspinal levels) in terms of muscle activity during sagittal bending, which is suggestive that either control of forward flexion is not its primary function, or that TES is simply more efficient, possibly due to its relatively larger fibre size.

There is disagreement in the literature in terms of fibre type distribution in thoracic and lumbar regions. Mannion et al. (1997) showed that there was no difference between regions in terms of the 3 main fibre types (Type I, IIA and IIX), and concluded therefore that the thoracic and lumbar regions are likely to have similar functions and act synergistically during movements (Mannion et al. 1997). In an earlier fibre composition study however, Sirca and Kostevc (1985) concluded that the thoracic region consists of a greater proportion of type I fibres than the lumbar, suggesting a postural and movement stabilisation function (Sirca and Kostevc 1985). A limitation of both studies was a relatively small population sample, and therefore fibre type and size differences between regions remains incompletely understood. The results from this study do suggest however that TES and LES have clearly different roles during forward bending, which may also be reflected in their associated fibre types, as with the findings of Sirca and Kostevc.

7.1.7.4 The exploration of other muscle activity ratios

Previous work has indicated a clear distinction between the kinematic behaviour of the upper and lower sections of the lumbar spine (Pavlova et al. 2015; Mitchell et al. 2008), which is in agreement with the compensatory relationship shown between the upper and lower lumbar regions (i.e. L2-L4 IV-RoMmax inversely correlated with L4-L5 IV-RoMmax) shown in the current study. It was anticipated therefore that there may be correlations between the IV-RoMmax of upper and lower lumbar segments and the muscle activity ratio of LMU/LES (LMU recorded adjacent to L5 in the lower lumbar spine, and LES adjacent to L2 in the upper). Whilst the ratios of both LMU/LES and LMU/TES were explored, neither demonstrated any significant relationship with IV-RoMmax at any level, and so any interactions in terms of a synergistic stabilising function between these groups is not clear. Whilst it has been suggested that LMU and LES have similar functional roles (Stokes et al. 2003) which would make compensatory behaviours more likely, the current findings would suggest that LMU is

likely to have a fundamentally different role to the thoracic and lumbar longissimus muscles, and functions independently to them. The fact that no significant relationships were found is perhaps also indicative of the different control strategies required for L4-L5 and L5-S1, as unlike L2-L3 and L3-L4, the two lower lumbar segments do not function in a uniform manner, and so the specific role of the superficial LMU appears more closely linked to the movement of L5-S1. Indeed, the results would suggest that LMU activity has an important role in the segmental control of movement at L5-S1 specifically, which is discussed further in a following section (Section 7.1.7.9). Future studies investigating the potential role of upper vs lower lumbar muscles in inter-vertebral movement control, may therefore benefit from the use of electrode array systems that can record from multiple adjacent muscle levels, to ensure that significant relationships are not missed.

In addition, given that the larger the range of ratio values within a group the more likely it is that associations will be found, ratio ranges were compared between muscle pairs. The range in values for both LMU/TES and LMU/LES ratios were found to be greater than the range for LES/TES, and so a relatively smaller range was not a reason for the absence of relationships.

7.1.7.5 The importance of consistent electrode positioning

The lack of correlations between IV-RoMmax and LMU/TES may partially explain why Reeves et al. (2006) and Van Dieen et al. (2003) showed contrasting results in terms of relating the lumbar ES/Thoracic ES ratio to LBP. The electrode positions used by Reeves et al. (2006) were effectively the same as this study's LMU and TES sites, which did not reveal any significant relationships. Van Dieen et al. on the other hand used electrode positions more similar to the LES and TES sites used in this study, providing a possible reason for their conflicting results. Indeed, there is disagreement in the literature over which muscles comprise local and globally acting groups. This study has interpreted LES as a locally acting muscle in line with Bergmark et al's 1989 original paper, and other subsequent studies (Bradl et al. 2005; O'sullivan 2000; Van Dieen 2003). Although the lumbar longissimus has both segmental and regional attachments, some authors define the muscle as globally acting (Kim et al. 2015), which will have consequences for the interpretation of these studies due to conclusions based on groupings (i.e. muscle allocation to local or global groups) that are not consistent throughout the literature. Differences between methodologies are especially common in EMG based studies, and this highlights the difficulty of comparing studies that use different protocols, and why meta-analysis of study findings in this area is uncommon. Further universal standardisation of these elements is therefore recommended.

7.1.7.6 Relationships between muscle activity amplitude changes and IV-RoMmax

It has been suggested that intersegmental forces maintain or decrease inter-vertebral motions (Panjabi et al. 1984; Kaigle et al. 1995). It would seem logical then, that if the role of the posterior muscles is to resist sagittal flexion, in order for inter-vertebral movement to occur, there must be a deactivation of this supporting musculature. In light of this theory, the following hypothesis was formulated.

Sub-hypothesis

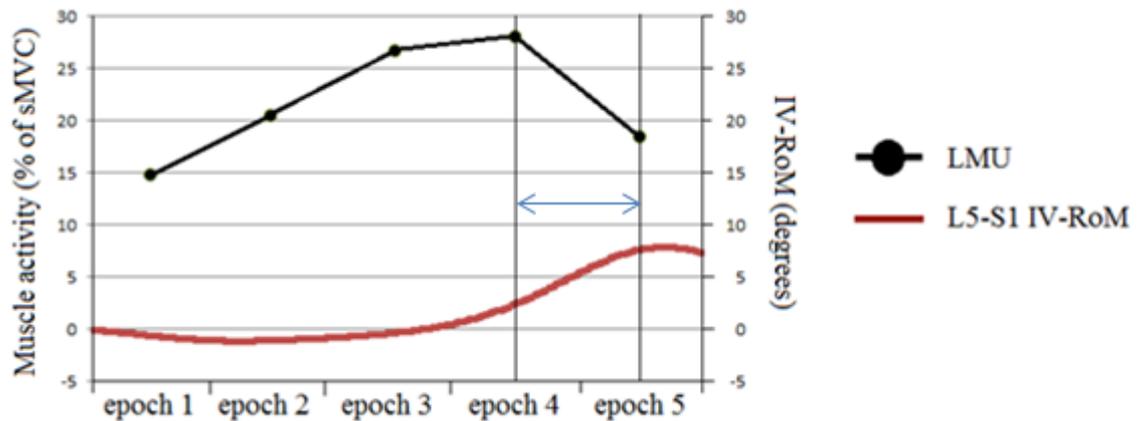
- There will be a direct relationship between the size of muscle deactivation and IV-RoMmax

The results demonstrate that changes in activity of TES, LES, and LMU at various stages of the forward bending cycle, can all be to some degree related to the IV-RoMmax at lower lumbar levels (L4-5 and L5-S1)¹³. These changes were not however uniformly just increased deactivation relating to increased IV-RoMmax, as there were also examples of increased activity relating to increased IV-RoMmax. Therefore, although the hypothesis can be accepted, the posterior muscles demonstrate diversity in terms of their stabilisation roles.

Figure 64 shows an example of how the muscles most local to the L5-S1 inter-vertebral segment (i.e. LMU) demonstrate a significant decrease in activity during the final stage of flexion in a healthy control subject. This appears to correspond with the phase lag (Kanayama et al. 1996) in the initiation of movement at the adjacent inter-vertebral level (i.e. L5-S1) which can be visualised using the motion graphs (e.g. Figure 26). Indeed it was shown that the larger the change in activity between epochs 4-5, (i.e. deactivation in the latter stages of the flexion cycle) the larger the IV-RoMmax at L5-S1 (Table 27). This is suggestive of a degree of direct localised control by LMU, however, other influences such as the stabilisation of the pelvis cannot be ruled out as a possible external cause. This direct relationship between corresponding levels was not apparent between the LES and the upper inter-vertebral lumbar motion segments (Table 27), and may be suggestive of anatomically specific control at this level.

¹³ Note: There were also many correlations that approached significance (Table 27), and therefore future studies with a larger sample size may well reveal more statistically important relationships, potentially with upper lumbar inter-vertebral levels.

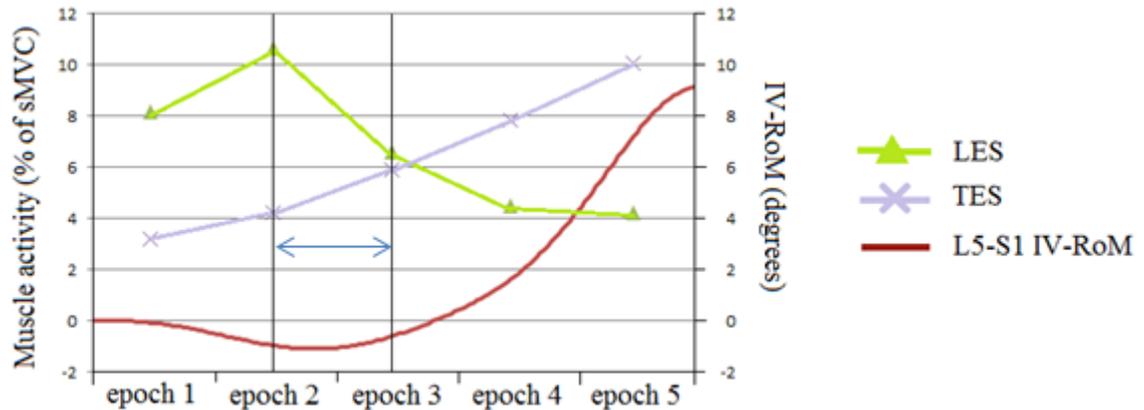
Figure 64: An example of LMU activity and lumbar IV-RoM during sagittal flexion



The highlighted section (between arrow heads) represents the change in muscle activity and IV-RoM between epoch 4 and 5. In this example it is shown that a decrease in LMU corresponds with an increase in angular range during the final stages of flexion.

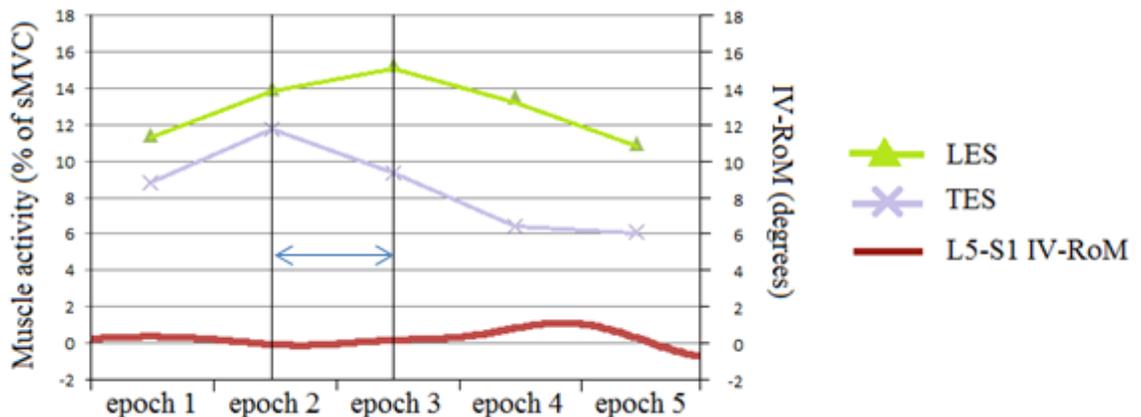
That is not to say LES or TES do not have a role in stabilising L5-S1 rotation. Of particular interest is the apparent shift in effect between TES and LES on the IV-RoMmax of L5-S1 (Figures 65 and 66). As LES activity decreases between epochs 2 and 3 of the cycle (early mid stage) there is an associated increase in L5-S1 IV-RoMmax, whilst at the same stage of the cycle TES changes (decrease) are significantly associated with a decrease in L5-S1 IV-RoMmax (Figures 65 and 66). This indicates possible different roles for TES and LES (and possible interaction) in terms of the control of the range of motion at a distal motion segment. If there is more movement at L5-S1 there may be less activity of LES, more TES, and vice versa when there is less movement.

Figure 65: An example of LES and TES activity and L5-S1 IV-RoM during sagittal flexion (An example of a greater IV-RoMmax). Please note that the scales of the Y-axis are slightly different to those seen in Figure 66.



The highlighted section (between arrow heads) represents the change in muscle activity and IV-RoM between epoch 2 and 3. In this example it is shown that a decrease in LES, and increase in TES activity during this period, relates to a higher IV-RoMmax ($\approx 9^\circ$).

Figure 66: An example of LES and TES activity and L5-S1 IV-RoM during sagittal flexion (An example of a smaller IV-RoMmax)



The highlighted section (between arrow heads) represents the change in muscle activity and IV-RoM between epoch 2 and 3. In this example it is shown that a decrease in TES, and increase in LES activity during this period, relates to a lower IV-RoMmax ($\approx 1^\circ$).

When considering the LES to be locally acting and the TES to be globally acting (Bergmark 1989), then these findings may have important clinical implications, as they raise the possibility of level specific stabilisation/control. Various arguments have been put forward regarding the role of local and global muscles in spinal stability. Whilst Bergmark suggested that inter-segmental (local) muscles were the chief stabilisers (Bergmark 1989), Crisco and

Panjabi concluded that the larger multi-segmental (global) muscles were more efficient (Crisco and Panjabi 1991), however the exact function of each muscle group remains unclear. Indeed, in a study investigating the relative contribution of different trunk muscles to lumbar stability, Cholewicki and Van Vliet concluded that whilst inter-segmental and multi-segmental paraspinals had the greatest effect on stabilisation compared to other muscles (i.e. psoas and rectus abdominis), no distinction could be made between the two (Cholewicki and Van Vliet 2002). The results of the current study however, have for the first time provided evidence of not only a distinctive, but possibly integrated functions of these muscles.

The relationship between changes in TES and LES activity between epochs 2-3 and IV-RoMmax of L5-S1 is of particular interest (Table 27). As un-checked co-contraction is associated with increased spinal loading (Gardner-Morse et al. 1998), it could be suggested that in the presence of increased LES activity during epoch 2-3, the concurrent decrease in TES is a mechanism to avoid excessive spinal loading. The activity changes in these muscles however correlate with the movement of a distal segment (i.e. L5-S1), and such changes occur most frequently before the onset of L5-S1 movement (Figures 65 and 66). It would seem therefore that a strategy to control movement at the very base of the spine, is to preemptively control the motion of motion segments above, or put another way, lower level movement is partially dependent on the kinematics occurring superiorly, and so the correlation may not relate directly to balancing compressive forces acting on L5-S1. No relationships were found between TES and LES changes during these epochs (i.e. epochs 2-3) and the IV-RoMmax at L4-5, which may have been expected given the relationship of this level with the overall ratio of LES/TES during flexion, and so this correlation cannot be explained solely in terms of epochs 2-3. Indeed, although movement onset patterns do vary somewhat, with examples of L2-3, L3-4 and L4-5 all moving first during forward bending, there are no examples of L5-S1 moving first, the segment typically last to begin motion (Appendix M). This suggests that regardless of the movement strategy, L5-S1 is usually last to move or prevented from motion until the latter stages of flexion. It would seem that this motion segment is protected from excessive movement by two mechanisms. Activity of LES during the early-mid stages of flexion (potentially also involved in controlling the movement of segments above), and activity of locally acting LMU during the latter stages.

7.1.7.7 Motor control and segmental interaction

The idea that the control of superior motion segments can effectively influence the movement of a more distal segment is demonstrated in other examples. A decrease in LES

activity in the early stages of flexion (i.e. between epochs 1-2) relates to an increase in L4-L5 IV-RoMmax, and a decrease in TES activity during the same stage relates to an increase in L2-L3 IV-RoMmax (Table 27). These examples highlight the fact that muscle activity occurring sometimes multiple levels above an inter-vertebral level of interest, can relate to its motion. In all of the above examples, anatomically it is possible for the proximal muscles to have a direct influence on the segment itself, however the delay between muscle activity and the onset of vertebral movement suggests that for these muscles (LES and TES), the strategy involves control of superior segments, or at least segments initiating movement before them. The LMU however has been shown to have a more direct influence over the local segment of L5-S1, as the degree of LMU deactivation is associated with L5-S1 IV-RoMmax (Table 27), suggestive of the fact that the locally acting superficial multifidus has a different role to both the locally and globally acting longissimus.

7.1.7.8 A stabilisation strategy of the healthy lumbar spine

“Under dynamic loading conditions, trunk muscles must be recruited in appropriate sequence and appropriate strength to support loads and maintain stability” (Cholewicki and Van Vliet 2002).

The patterns observed can be interpreted in terms of optimisation of spinal control. Returning to the concept of co-contraction, it is a strategy that balances increased stability at the cost of increased spinal loading (Granata and Marras 2000; 2001; Gardner Morse and Stokes 1998), and increased shear forces (Marras 2001). The principles of co-contraction are frequently associated with agonist and antagonist muscle groups, but can equally be applied to other muscle synergies (e.g. different paraspinal muscles). During forward bending the cumulative effect of a linear increase in both local and globally acting muscle activity, whilst increasing stability, would also increase spinal loading, and be costly in terms of energy expenditure. It is likely therefore that the motor control system adapts its strategy in terms of local and global muscle activity, dependent on the biomechanics of the individual. The direct interaction between the two such groups (i.e. LES and TES) has never been shown before during a dynamic task *in vivo*.

If there is a system requirement to restrict movement at L5-S1 (i.e. segmental instability at this level), then an increase in LES activity and a decrease in TES is a possible strategy. Whether or not this is the most efficient strategy in terms of the metabolic demands placed on the muscles (Salmons and Henriksson 1981) is not known, but it may be an important enough requirement to justify a sub-optimal strategy, contradicting the view that specific

inter-vertebral motion is not important due to a likely 'approximate' motor control strategy (Cholewicki and McGill 1996). This relationship shows a possible inter-play between local and global muscles at a specific stage of dynamic activity, and is in agreement with Cholewicki and Van Vliet (2002), who suggested that stability depends on the relative activation of multiple trunk muscles (Cholewicki and Vliet (2002). They also suggested that no particular muscle can be identified as a chief stabiliser, however it can be extrapolated from this study that if restriction of angular rotation at L4-L5 or L5-S1 is the goal of a muscle activity strategy (for example to prevent pain associated with excessive movement at these levels), then an increased activation of LES may be of key importance during forward bending (in association with changes in activation of other muscles). The specific muscles that are key to stabilisation may depend directly on the requirements of specific inter-vertebral levels during a given task. The collection of further normative data during a range of tasks is therefore required.

7.1.7.9 LMU control of L5-S1 IV-RoMmax

Visual inspection of the EMG and motion graphs (Appendices L and M) suggested that the deactivation of LMU in the final stages of flexion would correlate with the IV-RoMmax of L5-S1. This was supported by the correlation found between LMU deactivation between epochs 4-5 and the IV-RoMmax at this level (Table 27). This relationship strongly supports the theory that locally acting muscles stabilise at a segmental level (Bergmark 1989), and that the degree of deactivation correlates with the concurrent sagittal angular rotation achieved. This is the first time that this has been shown *in vivo*, and provides evidence that the degree of LMU muscle activity change relates directly to the sagittal rotation of L5-S1. This therefore suggests that the functional capacity of LMU would be of possible importance, if the movement of L5-S1 were a clinical concern.

It is also clear from the combined motion graphs that LMU deactivation begins well before maximum vertebral rotation of L5-S1 is reached (indeed it is possible that true IV-RoMmax of L5-S1 is not even reached during the 60° of bending). This therefore raises questions over the current theories regarding the mechanism of the FRP (Section 2.3.6), and the methodologies used to examine it. It has been suggested that deactivation occurs when the bending moment is countered by either sufficient tension being reached in the passive structures (Floyd and Silver 1955), the activity of other muscle(s) (e.g. Quadratus lumborum or the deep lateral ES) (Andersson et al. 1996), or the passive resistance of myoelectrically silent stretched muscles (Adams et al. 1980). In the case of LMU, activation begins with the onset forward bending, and deactivation begins prior or during the early stages of L5-S1

rotation, certainly before most of the rotation has been completed. Therefore questions arise over how feedback mechanisms are involved in the initiation and deactivation of LMU muscle activity during normal movements.

7.1.7.10 Could the thoracolumbar fascia (TLF) have a role in sensory feedback?

It appears in the case of LMU, that its activation does not fit the ligamento-muscular synergism model (Solomonow et al. 1998), as the mechanoreceptors in passive tissues are unlikely to be stimulated prior to movement of L5-S1 (Indahl 1997; Indahl et al. 1995; Solomonow et al. 1998; Stubbs et al. 1998), and if they are involved, they must have a low threshold for activation. Indeed, it has been suggested that such mechanisms are actually most likely to be detectors of end ranges of motion (Proske and Gandevia 2009). This would indicate that the dominant neural control strategy is either feedforward governance (Hodges 2001; Hodges et al. 2013; Hodges and Richardson 1999), or that feedback mechanisms are initiated by the passive tissue stress in superior motion segments. An alternative theory was proposed by Willard et al. (2012), who suggested that due to the close proximity of the passive structures (i.e. facet capsules, ligaments and discs) to the axis of the spine, large rotational movements would be required to stimulate the mechanoreceptors within them. The TLF however, as a more superficial structure would require much less inter-vertebral movement to invoke stretch stimulation, which they proposed as a possible alternative feedback mechanism (Willard et al. 2012). This concept could explain the early activation of both LES and LMU muscles observed during flexion in this study, although it is not yet established whether the TLF has the proprioceptive capacity to support such a function (Willard et al. 2012).

7.1.7.11 Is LMU's primary role to restrain forward flexion?

The multifidus has been purported to play a primary role in lumbar spinal proprioception (Bakker and Richmond 1982; Nitz and Peck 1986; Richmond and Bakker 1982). The relationship between LMU deactivation and L5-S1 IV-RoMmax shown in this study however, would suggest that the superficial LMU also plays an important physical role in restraining inter-vertebral movement, and as such is not solely a proprioceptive structure. Indeed, the Richmond and Bakker studies were conducted with feline specimens (Richmond and Bakker 1982, Bakker and Richmond 1982), and the Nitz and Peck (1985) study based their findings on a small sample of 2 adult cadavars and three 36 week old foetuses (Nitz and Peck 1985), and so their relevance could be questioned in relation to a sample of healthy males. The potentially different roles of the superficial and deep multifidus could also be a

consideration, as it is not fully known how functional roles may vary between locations (Macdonald et al. 2006).

The relationship demonstrated between LMU deactivation during epochs 4-5 and L5-S1 is the first time that the degree of muscle deactivation has been directly correlated directly with a concurrent increase in IV-RoMmax *in vivo*. This supports the theory that locally acting muscles stabilise at a segmental level (Bergmark 1989), and suggests that the functional capacity of LMU would be of particular importance if the movement of L5-S1 were a clinical concern. The findings also suggest that the superficial LMU plays an important role in restraining the inter-vertebral movement at L5-S1, and is less likely to have a significant proprioceptive function at this level.

7.1.7.12 Understanding of the roles that TES, LES and LMU have in terms of lumbar stabilisation, can be enhanced by collecting concurrent inter-vertebral and multiple muscle data

In terms of paraspinal muscle deactivation, it has been suggested that sensory feedback mechanisms initiate paraspinal muscle deactivation at approximately 80% of the range of joint motion (Kaigle et al. 1998). In agreement, a number of other FRP based studies have suggested that the onset of muscle deactivation corresponds with the near completion of sagittal bending (Floyd and Silver 1951; Kippers and Parker 1984). As described above however, this study shows that LMU deactivation begins prior to, or concurrently with L5-S1 rotation which would suggest that either the deactivation mechanism is not initiated by changes associated with the completion of L5-S1 rotation, or that high threshold mechanoreceptors do not have a significant role.

This highlights a common limitation of many FRP related studies, in that they use gross measurements, which typically vary between studies, and will usually record EMG from a single muscle, typically longissimus adjacent to L3 (Kippers and Parker 1984; Gracovetsky et al. 1989; Steventon and Ng 1995; Sarti et al. 2001; Solomonow et al. 2003; Olson et al. 2004; Descareaux et al. 2010) (Table 2). This means that no conclusions can be truly made about the potential sensory feedback from inter-vertebral movements, and that no distinction is usually made between different muscles (e.g. TES/LES/LMU). Indeed, even when inter-vertebral information has been collected, the data were pooled from several levels (Kaigle et al. 1998). To date, no other study has investigated the relationship between IV-RoM and the deactivation of specific muscle groups (e.g. local or globally acting paraspinals), which is desirable given the current need for better understanding of neural feedback control mechanisms. It appears likely that muscle deactivation mechanisms may be different

between LES and LMU muscles, and an investigation into the precise timings of LES and LMU deactivation in relation to temporal kinematic parameters (e.g. movement onset, peak laxity, and IV-RoMmax) are warranted. This would be feasible using this study's data set, and is therefore an opportunity for future work.

7.1.8 Conclusions

It is suggested that achieving sufficient spinal stability is a moving target, and that no single muscle can therefore be considered the best stabiliser, as the most important muscle is transient dependent on the task (McGill et al. 2003). The current study's results provide a demonstration of this concept in action during the task of forward bending. Whilst effect sizes are small, inter-vertebral movements have been shown to be influenced by specific muscle activity strategies. Of particular interest was the correlation between decreased LMU and increased IV-RoMmax at L5-S1 in the latter stages of flexion, the apparent co-dependency between LES and TES during early to mid-flexion, and the effect of the LES/TES ratio on the IV-RoMmax at L4-L5.

In a LBP free population sample, it may be assumed that such relationships do not represent adaptations to pain. However, that is not to say that particular activity patterns and thus kinematic behaviours may not be risk factors for future LBP episodes. These relationships, when combined with other influencing factors, may therefore be important when these specific inter-vertebral levels are considered to be sources of pain generation and when considering rehabilitative or surgical planning.

Chapter 8: Contributions to knowledge

8.1 Introduction

The following chapter outlines in detail the areas where this body of work has made a significant original contribution to knowledge.

8.1.1: Contemporaneous QF and sEMG analysis

This study required the development of a protocol that would allow the concurrent investigation of lumbar inter-vertebral motion and muscle activity during movement in the sagittal plane. The resulting protocol combined the use of QF and sEMG for the first time, providing synchronised continuous inter-vertebral kinematic and muscle activity data, an innovation that enabled the investigation of relationships between inter-vertebral level parameters that has not previously been possible.

8.1.2: Observer repeatability of QF weight-bearing IV-RoMmax and initial attainment rate measurements

The QF technique produced IV-RoMmax and initial attainment rate measurements, with good repeatability. Fluoroscopic techniques have now been shown to be capable of producing repeatable IV-RoM measurements during weight-bearing examinations, at all levels of the lumbar spine (other than L1-L2). This adds to the previous body of knowledge that has shown good repeatability during recumbent QF examinations (Mellor F.E. et al. 2014).

This was the first time that the repeatability of the initial attainment rate measurement has ever been investigated during weight-bearing QF imaging. The results suggest that the measurements have acceptable reliability and agreement, which support the use of this parameter within the current study. Further investigations are required to establish whether or not initial attainment rate is representative of the neutral zone during inter-vertebral rotation in the sagittal plane (Breen et al. 2015), however as a parameter of *in vivo* motion segment laxity, it has shown its potential for use in future biomechanical studies.

8.1.3: Intra-subject repeatability of RMS sEMG amplitude measurements (highlighting a benefit of the standardised study protocol)

The repeatability of the sEMG amplitude measurements was also shown to be acceptable, which is notable considering the inherent variability of EMG signal recordings (Lehman and McGill 1999). This was the first time EMG has ever been recorded during motion using the standardised QF motion frame apparatus, and it is suggested that the good agreement and reliability of measurements was in part due to the standardisation of the participant's

movement. The standardisation of the QF examination (Breen et al. 2012) resulted in measurements that were less influenced by variations in how participants moved, and the velocity and range over which they did so. It is likely that this standardisation process had a positive influence on the repeatability of sEMG amplitude measurements, which affords improved comparisons between and within individuals. It is therefore recommended that standardised movement protocols are incorporated into future dynamic EMG studies.

8.1.4: Relationships between lordosis and the IV-RoMmax

This is the first time that lumbar lordosis has been shown to relate to the inter-vertebral range of specific motion segments in healthy controls. This is a potentially important finding as it shows how an individual's spinal curvature in a neutral position can influence subsequent inter-vertebral movement patterns during forward bending. The study showed specifically that in more lordotic spines, greater inter-vertebral rotation will occur at L2-L3, whereas in individuals with a flatter lordosis, more inter-vertebral rotation will occur at L4-L5. This will have consequences in terms of the stress placed on specific motion segments, which warrants further exploration, both in terms of risk factors for the development of LBP, and as aggravating factors in existing CNLBP populations.

8.1.5: Inter-level relationships in terms of IV-RoMmax

The study demonstrates a compensation mechanism between upper and lower regions of the lumbar spine in healthy controls. This supports the findings of Mitchell et al. (2008) who showed an inverse relationship between upper and lower lumbar regions (Mitchell et al. 2008), however this is the first time that relationships between specific inter-vertebral levels (i.e. the direct relationship between L2-3 and L3-4 and the inverse relationship between L2-3 and L4-5 IV-RoMmax) have been found. These findings highlight the existence of segmental interactions in terms of IV-ROM, which may be a mechanism of retaining an optimal sagittal balance (Rothenfluh et al. 2015) in healthy spines. This information is of importance to future research, when the stability of specific segments is of interest. As interactions between levels are evident, the results again suggest that individual segments should not be considered in isolation.

Due to the heterogeneity of IV-RoM in LBP populations, its importance as an indicator of LBP has been questioned (Mellor 2014). However, the demonstration of interactions between levels represents a possible new approach in terms of investigating relationships between IV-RoM and LBP, which may be of particular use when exploring the biomechanical basis of pre-determined CNSLBP sub-groups (O'Sullivan et al. 2005).

8.1.6: Relationships between initial attainment rate and IV-RoMmax

This is the first time initial attainment rate has been investigated during a dynamic weight-bearing activity. The parameter therefore represents motion segment laxity whilst under the influence of trunk muscle contraction. The direct relationship found between the IV-RoMmax at L4-5 and the initial attainment rate at L4-5 is the first time this has been shown *in vivo*, and suggests that increased range relates to an increased segmental laxity, and therefore decreased stiffness. This relationship was only found at this level however and so cannot be considered as a uniform behaviour throughout the lumbar spine. This highlights the benefit of inter-vertebral information, as assumptions regarding uniform behaviour of motion segments informing spinal models are not accurate (Arjmand et al. 2010).

8.1.7: A relationship between the ratio of LES/TES muscle activity and the IV-RoMmax of L4-L5

The results show that the ratio of LES/TES can vary in a population with no history of low back pain, and relate to variations in inter-vertebral biomechanics. In healthy males, it has been shown that an increase in the ratio of LES/TES provides increased restraint at the level of L4-L5 during forward bending, demonstrating for the first time how coordinated changes in the activity of locally and globally acting paraspinal muscle groups can influence the movement of a specific lumbar motion segment. This adds detailed inter-vertebral level information, to the body of work that considers this ratio as a mechanism of altering the stiffness of the lumbar spine (Van Dieen et al. 2003; Reeves et al. 2006; Van Den Hoorn et al. 2012; Lariviere et al. 2002).

8.1.8: Relationships between muscle activity changes during the flexion cycle and the IV-RoMmax of the lower lumbar motion segments (i.e. L4-L5 and L5-S1).

8.1.8.1 TES and LES

Whilst many studies have investigated muscle activation patterns during trunk flexion, the majority have focussed on either responses to perturbation around the neutral position, or the FRP near the end range of movement. To date, less emphasis has been placed on changes in recruitment throughout the entire flexion movement cycle, but the results of the current study have shown that doing so is of value.

This study has shown that interactions between globally and locally acting muscles (i.e. TES and LES) can influence the range of motion at a distal motion segment (i.e. L5-S1), in that more movement at L5-S1 correlates with less activity of LES, more of TES, and vice versa. This

indicates different roles for TES and LES in terms of the control of a specific motion segment, which is in agreement with some of the literature (Bergmark 1989, O'Sullivan et al. 2000).

The relationship shown may therefore represent a mechanism to optimise spinal loading conditions (Granata and Marras 2000), as the increased LES activity during epoch 2-3 coincides with a decrease in TES activity, a strategy that avoids excessive spinal compression. This is the first time these relationships have been demonstrated *in vivo*, and shows that activity changes in superior muscles (i.e. TES at T9 and LES at L2) correlate with the IV-RoMmax of a distal segment (i.e. L5-S1). As these changes occur before the onset of L5-S1 movement (Figures 65 and 66), it is likely that pre-emptive control of motion segments above directly influences the movement of L5-S1.

In agreement with Cholewicki and Van Vliet (2002) who suggested that stability depends on the relative activation of multiple trunk muscles (Cholewicki and Vliet (2002)), the relationships show the inter-play between local and global muscles at a specific stage of dynamic activity. In disagreement with their conclusions however, the current study findings suggest that specific muscles may be of particular importance in terms of individual inter-vertebral level stabilisation. Indeed, a relative increased activation of LES over TES during the flexion movement has been shown to directly affect the restraint of L4-L5, and an increase in LES activation during mid flexion (i.e. epochs 2-3) has been shown to directly affect the restraint of L5-S1. These examples, and the relationship between LMU and L5-S1 IV-RoMmax (discussed below), are evidence of previously undemonstrated muscle specific inter-vertebral control strategies. All of which have potential for further exploration.

8.1.8.2 LMU

The relationship demonstrated between LMU deactivation during epochs 4-5 and L5-S1 is the first time that the degree of muscle deactivation has been correlated directly with a concurrent increase in IV-RoMmax *in vivo*. This supports the theory that locally acting muscles stabilise at a segmental level (Bergmark 1989), and suggests that the functional capacity of LMU would be of particular importance if the movement of L5-S1 were a clinical concern. The findings also suggest that the superficial LMU plays an important role in restraining the inter-vertebral movement at L5-S1, and is less likely to have a significant proprioceptive function at this level.

8.1.9: Relationships between initial attainment rate and muscle activity changes

Further analysis of the study data (Section 9.10) revealed significant relationships between initial attainment rate at L2-L3 and L3-L4 and an increase in TES muscle activity during epoch

2-3, and that there was a trend between an increase in L4-L5 initial attainment rate and increased LES activity during the same period. This is the first time such relationships have been shown, and they suggest that motor control strategies during segmental laxity may be different between the upper and lower lumbar segments.

8.1.10: A relationship between lordosis and the initial attainment rate of L4-L5

Further analysis of the study data also revealed an inverse relationship between lordosis and L4-L5 initial attainment rate. Therefore as lordosis increases, there is either an increase in stiffness at L4-L5, a decrease in bending moment at that level, or both. This again, is the first time correlations have been shown between these parameters. Such relationships further our current understanding of normal biomechanical interactions, and also warrant further exploration in future investigations.

8.2 Summary

The demonstration that weight-bearing inter-vertebral and muscle activity information can be reliably collected concurrently *in vivo*, is an important advancement in spinal biomechanics. It represents a progression from traditional region-based kinematic measurements, which when combined with EMG recordings has provided a level of insight into the interactions between the active, passive and neural control systems that has not previously been achieved.

This study has also demonstrated that a spectrum of lordosis, kinematic and muscle activity measurements exists in a population of healthy controls, and that changes in these parameters relate to the IV-RoMmax of specific inter-vertebral levels. This is an important consideration for future NSLBP research, as any attempts to associate these parameters with LBP, should also now take in to account the normal biomechanical behaviour of an individual's lumbar spine. Indeed, consideration should also be given to the interaction that exists between such parameters, and inter-vertebral levels should not be considered in isolation of the behaviour of the rest of the lumbar spine.

Whilst these findings may potentially be of valuable clinical significance, further normative studies incorporating larger sample sizes are required. The limitations and potential routes forward for this protocol were explored, underlining the large scope for further work.

The following chapter discusses the relevance of these findings to the broader aspects of biomechanical research.

Chapter 9: Discussion

9.1 Introduction

The general purpose of this thesis was to investigate the relationships between kinematic variables, muscle activity variables, lordosis and IV-RoMmax during weight-bearing sagittal flexion. The agreement and reliability of IV-RoMmax, initial attainment rate, and muscle activity amplitude measurements were shown to be acceptable (Chapter 5), and moderate to strong correlations were discovered (Chapters 6 and 7), providing a valuable insight into the normal biomechanics of the lumbar spine. It should be noted however that the results are only generalizable to young, healthy adult males, which should be considered in their interpretation. This chapter considers the implications of these findings to the field of spinal biomechanics and addresses the study's limitations.

9.2 Lumbar biomechanics and back pain: cause or effect?

The ability to directly link CNSLBP with a mechanical cause has eluded the research community despite altered mechanics intuitively being related. As previously described, the problem is partially due to the heterogeneity in both healthy and NSLBP populations in terms of their inter-vertebral movement and muscle activity behaviours. Indeed, it has been suggested that the measurements of one small group of LBP participants within a heterogeneous LBP group can be counterbalanced by opposing results from other sub-groups (Van Dillen et al. 2003), perpetuating an inability to uncover a clear cause and effect.

Referring back to the examples in the background section (Table 1), the contrasting results found between studies that attempt to identify parameters associated with LBP, may be partially explained by the normal biomechanical relationships shown in the current study. The relationships show that specific kinematic and muscle activity variables can influence the degree of inter-vertebral rotation in healthy participants, and so corresponding behaviours found in the biomechanics of NSLBP groups may in some way be related to the pain production mechanism, or can be considered as normal biomechanical variations that are not necessarily involved. If such mechanical behaviours are associated with LBP, there are two ways they could be involved. 1. They somehow predispose to pain, 2. They are a consequence of pain (i.e. an adaptation to avoid pain or a mechanism to stabilise the spine).

Using muscle activation as an example, Sanchez-Zuriaga et al. 2015 when comparing muscle activity between healthy controls and patients with recurrent LBP (during a pain free period), suggested that an increase in ES muscle activity associated with the LBP group may be a mechanism to increase stability. Their study recorded ES activity adjacent to L3 and not L2

(as in this study), however their recordings also represent activity of the lumbar longissimus, and the authors proposed that this increase in activity may be a compensation mechanism for damaged spinal structures in the LBP group (Sanchez-Zuriaga et al. 2015). The findings of the current study however show that an increase in LES activity can also relate (although the relationship did not quite reach significance) to a decrease in L4-L5 IV-RoMmax (Table 26), suggestive that either a normal motor control strategy is more common in the Sanchez-Zuriaga et al. study LBP group, or that the activity relates to a physiological requirement for increased restraint at L4-L5. The same can be said regarding changes in the LES/TES ratio, as normal changes in this ratio also relate to the IV-RoMmax of L4-L5. Therefore the motor control strategies observed by Van Dieen et al. 2003 (i.e. the LES/TES ratio increases in order to stabilise the lumbar spines of the LBP group), could also be explained by normal biomechanical behaviours, possibly unrelated to the LBP (i.e. the LBP group had a high proportion of participants who did not move much at L4-L5). In both cases however, it is possible that in the LBP groups, the stabilisation of L4-L5 is in some way related to the LBP. If inter-vertebral measurement techniques can be applied in such studies, there is an opportunity to investigate these kinds of level-specific mechanical relationships. Similar arguments could also be made for the kinematic variables studied and lordosis, underlining the potential for future inter-vertebral level based research.

9.3 Muscle activity patterns and pain predisposition

The co-dependent nature of lumbar biomechanics has been shown by this study to accommodate certain parameter excesses through adaptations by others. These interactions enable the healthy lumbar spine to retain function whilst performing tasks such as forward flexion without pain, but whether these adaptations are sustainable, or may themselves eventually lead to injury or pain cannot be extrapolated. In the current study, no participants had LBP prior to or during the investigation, and so it may be assumed that the muscle activity behaviours found were not influenced by pain. In a study of healthy controls, Gregory et al. (2008) found that participants who developed pain whilst standing for extended periods had different pre-existing muscle activation patterns to those that did not, with the so called pain developer group demonstrating higher levels of muscle co-activation (in this case of the gluteus medius and the trunk flexors and extensors) (Gregory et al. 2008). Differences have also been shown in terms of muscle activation onset times. Nelson-Wong et al. 2012 also using a method of exposing healthy controls to prolonged standing, showed that during sagittal flexion and extension, pain developers demonstrated activation of the lumbar extensors prior to gluteus maximus, which was reversed in those that did not develop

pain (Nelson-Wong et al. 2012). This suggests that certain muscle activity patterns that are found in normal participants may predispose to pain development. Future studies therefore could feasibly consider the relationships found in this study in terms of pain predisposition.

9.4 A focus on forward flexion

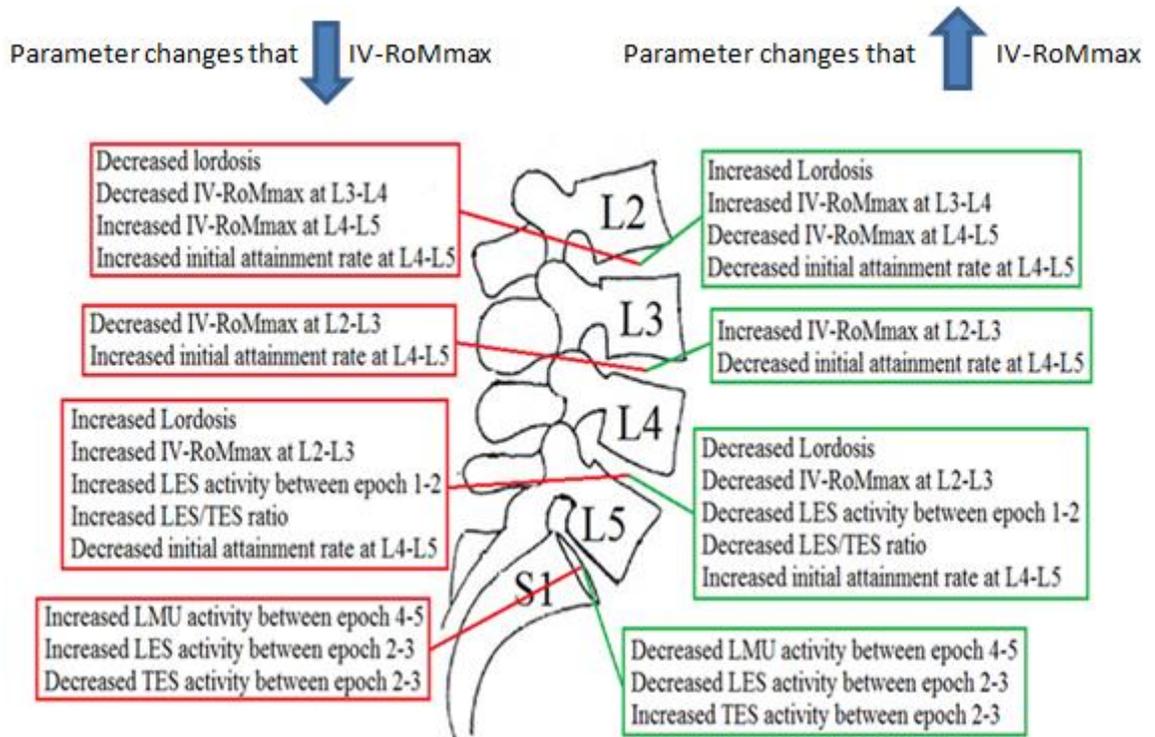
Clinically, the present study has most relevance to LBP that is related to tasks involving forward flexion. It has been shown that people with LBP are likely to have a more flexed sitting posture than non-LBP groups (Womersley and May 2006), and that even in pain free populations it is common for some individuals to naturally assume more flexed postures than would be considered ideal (O'sullivan et al. 2010). Studies have also shown that tasks involving extended periods of sitting (slouching) or repeated forward flexion can increase the risk of developing LBP (Lotters et al. 2003; O'sullivan et al. 2006b), and it has been suggested that a larger range of lumbar flexion during such flexion activities, may be a reflection of increased passive tissue laxity, and therefore a diminished ability to stabilise the spine (Hoffman et al. 2012). Such tissue characteristics are believed to predispose individuals to low back injury and therefore pain (McGill and Cholewicki 2001).

Considering the example of the LES/TES ratio, an increase or decrease in the ratio will have certain biomechanical consequences. A relative increase in LES activity will place greater physiological requirements on that muscle specifically, which could feasibly predispose the muscle to injury, particularly when exposed to repetitive flexion movements (Dickey et al. 2003). Likewise if the LES/TES ratio is smaller, there would be an increased IV-RoMmax at L4-L5, correlating with a relative decrease in local muscle control, possibly leaving the passive structures vulnerable to unexpected perturbations. It has been suggested that at less than 3 degrees of sagittal flexion rotation, the disc is protected from injury (Bogduk 2012). If an individual has a large normal (i.e. pain free) inter-vertebral IV-RoMmax, and a further strain is imposed in addition to the pre-existing strain due to flexion, then they may be more likely to sustain an injury to that level.

This raises the possibility of NSLBP mechanical phenotypes. For example, individuals with a large L4-L5 IV-RoMmax, a shallow lordosis, a stiff upper lumbar spine and a small LES/TES ratio could feasibly be at greater risk of certain kinds of mechanical injury. Figure 67 shows all the mechanical parameters that have been found in this study to influence changes in IV-RoMmax at different inter-vertebral levels. If such phenotypes can be established, then it may lead to improved methods of subgrouping. Particularly relevant are the Movement

System Impairment (MSI) (Sahrmann 2002) and the O’Sullivan (OSC) (O’Sullivan 2005) classification schemes, as both identify movement directions that elicit symptoms.

Figure 67: The influence of the investigated parameters on the IV-RoMmax at different lumbar levels



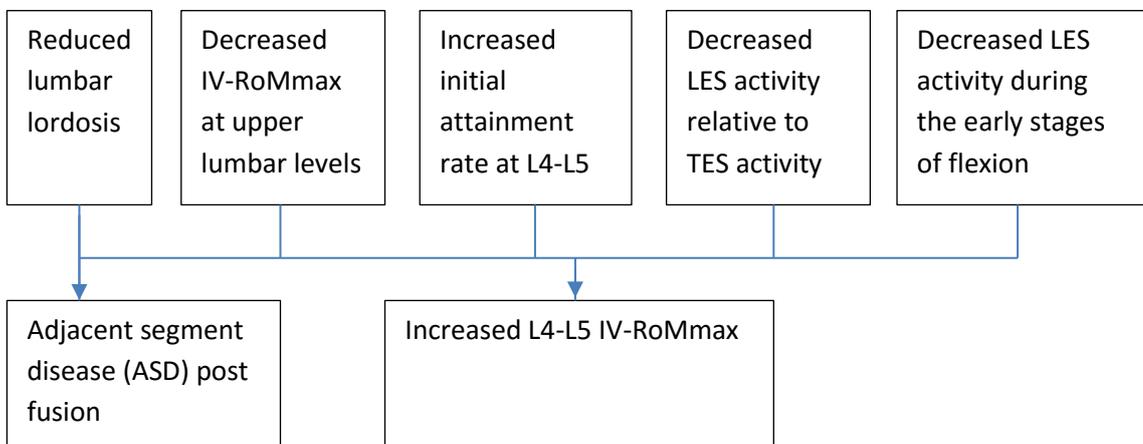
9.5 Sub-grouping populations

The O’Sullivan et al. (2005) sub-grouping system has been used in many studies that attempt to classify NSLBP (Dankaerts et al. 2006; 2009; Hemming et al. 2015), yet it remains unclear as to the underlying mechanical reasons for the symptoms that determine group allocation. An exploration of these sub-groups using inter-vertebral level information may help to reveal the mechanical reasons (if any) why such sub-groups exist. Using the lordosis versus IV-RoMmax results as an example, sub-grouped flexion and extension pattern patients tend to occupy opposing ends of the lumbar posture spectrum (Dankaerts et al. 2009). Flexion aggravated patients more commonly have a kyphotic lumbar spine, which from this study is associated with more movement at L4-5, and less at L2-L4, which may contribute to pain generation at these lumbar levels. Inter-vertebral information may therefore help provide more insight into the mechanism of pain in such groups.

9.6 Spinal surgery

The concepts raised also have possible consequences for spinal surgery, where the effects of spinal fusions on other areas of the spine is of obvious interest (Radcliffe et al. 2013), and maintaining sagittal balance is believed to be an important outcome predictor (Le Huec et al. 2015). Of the spinal parameters considered in the current study, a decreased lordosis has been linked to an anterior sagittal balance and LBP (Glassman et al. 2005; Le Huec et al. 2015; Jackson et al. 1994), and the development of ASD (Rothenfluh et al. 2015). Using the example of movement at L4-5, a commonly surgically fused motion segment (Le Huec et al. 2015), it has been shown that increased movement correlates with a decrease in lordosis, which indicates a possible association between L4-L5 movement and pain. All the biomechanical parameters that have been shown to influence L4-L5 IV-RoMmax in this study are shown in (Figure 68). If this segment is surgically fused therefore, it is likely that the movement will have to be taken by other segments, and biomechanical parameters that are associated with an increased L4-5 range may therefore be considered as associated risk factors for pseudarthrosis (Lee et al. 2011).

Figure 68: Diagram of the possible influences of lordosis, segmental kinematics and sEMG patterns on a surgically stabilised L4-L5 motion segment



In the case of L4-5 inter-vertebral motion, surgical planning could also feasibly consider interactions between these sEMG activity variables, the patient's standing lordosis and the kinematics of upper lumbar levels as possible risk factors for adjacent segment disease (ASD). However, *in vitro* investigations which have now been validated using *in vivo* kinematics shed little light on the actual effect of different stabilisations (e.g. fusion) on adjacent segments (Volkheimer et al. 2015), and so any proposed effects remain intuitive.

It must also be remembered that the present study's results represent normal biomechanical behaviour in healthy individuals where the movement interactions would be considered pain free compensations. It is possible however that these parameters (e.g. Figure 68) could predispose to LBP. Much depends on an individual's ability to adapt, be that through changes in muscle recruitment strategy, spinal kinematics, lordosis, pelvic incidence or thoracic curvature, and whether these changes are adaptive or maladaptive (O'Sullivan et al. 2005).

9.7 Segmental biomechanics and individualised care

If further work confirms the normal relationships found within this study, it could provide a foundation for more individualised diagnosis and management plans for people with NSLBP. Much depends on the ability to identify whether the nociceptive source is a consequence of segmental instability or restriction (i.e. associated with inter-vertebral movement). Instability is believed to be associated with size of the neutral zone (Panjabi 1992b; Youssef 2008), and so the initial attainment rate parameter, as a surrogate indicator of the neutral zone (Breen et al. 2015), may be of use in future instability studies. In terms of IV-RoM, excessive movement has been described as lumbar segmental instability (LSI) (Abbott et al. 2006; Ahmadi et al. 2009; Teyhen 2004, O'Sullivan et al. 2000, Hasegawa et al. 2011) and limited movement as lumbar segmental rigidity (LSR) (Abbott et al. 2006; Mayer et al. 2004; Teyhen et al. 2007). However in a systematic review of tests purported to clinically diagnose LSI, Alquarni et al. (2011) concluded that the majority of tests were inadequate for doing so (Alquarni et al. 2011). The natural heterogeneity in terms of inter-vertebral movement makes it difficult to assess instability in a clinical situation even with the use of spinal imaging, as the thresholds at which motion segments are determined as either hypo or hypermobile are not standardised and remain largely arbitrary (Abbott et al. 2006; Ahmadi et al. 2009). A lumbar instability questionnaire has recently been developed (Macedo et al. 2014); however this is yet to be validated. Nevertheless, aberrant spinal movement patterns are widely believed to relate to spinal pain and dysfunction (Iguchi 2004; Kanemura 2009; Smit et al. 2011; Spinelli et al. 2015), and so the relationships demonstrated in this study may be of importance, if the control of a specific motion segment, and a knowledge of interactions with other levels is desirable.

The use of this information in terms of spinal rehabilitation is dependent on having baseline inter-vertebral kinematic data. Intuitively, stabilisation will not be helpful for an already hypomobile segment, but will be of benefit if hypermobility is assumed to be the problem, and vice versa, in terms of mobilisation. This is consistent with an RCT conducted by Fritz et al. 2005, who concluded that patients categorised as being either hypo or hypermobile

respond better to respective mobilisation or stabilisation treatment programs (Fritz et al. 2005). The current study findings showed that the motor control strategies varied between participants, and so rehabilitation program design should account for the biomechanical characteristics of the individual. The results suggest that the recruitment patterns of local and global stabilisers in healthy controls influence the inter-vertebral ROM of lower lumbar segments. Specifically, if the overall ratio of LES over TES increases, then less movement will occur at L4-L5. Likewise, if there is a relative increase in LES activity and a decrease in TES activity during mid-cycle, then L5-S1 range will be reduced. The movement of L5-S1 has also been shown to relate directly to the activity of LMU.

This could have consequences for the treatment of LBP, as the benefit or detriment of targeting specific muscles during rehabilitation exercises becomes dependent on the biomechanics of the individual, and whether or not these biomechanics are a pain determinant. Reeves et al. (2006) argued that preferential recruitment of the thoracic erector spinae (i.e. a globally acting muscle) could be more appropriate than recruiting locally acting muscles, due to the greater proportion of fatigue resistant fibres in this region (Reeves et al. 2006), however the results of the present study would suggest that such a strategy would place particular stress on L4-L5 during forward bending, which would be counterproductive if this segment was unstable. Previous studies have shown that it is possible to target specific lumbar muscles during rehabilitation. Daneels et al. (2001) for example, showed that an increase in the size of multifidus could be achieved through a combination of stabilisation training and dynamic static strengthening (Daneels et al. 2001). Stevens et al. (2007) demonstrated that specific stabilisation training (bridging and 4 point-kneeling exercises) could alter local vs global muscle activity ratios in healthy participants (Stevens et al. 2007), and most recently, it has been shown that selective activation of LMU over LES can be achieved (Kim et al. 2015).

A systematic review of motor control RCTs concluded that a motor control intervention (focussed on TrA and Multifidus) is no more effective than manual therapy or other forms of exercise in reducing LBP and disability, but is superior to minimal intervention (Macedo 2009). This conclusion can perhaps be explained by the fact that some individuals will benefit from an increased function of multifidus, as it may relate to a specific biomechanical problem, but some will not. Therefore, whilst Cholewicki and Van Vliet (2002) suggest that no single muscle can be the most important in terms of lumbar spine stability (Cholewicki and Van Vliet 2002), the present study has shown that under certain biomechanical

conditions, preferential isolation of specific muscles may be beneficial. Future focus for NSLBP rehabilitation should therefore be placed on improved understanding of lumbar kinematics on an individual basis.

9.8 Spinal modelling

A primary aim of spinal models is to predict joint reaction and muscle forces, and in order to do so, an approach known as inverse dynamics is commonly used (Hansen et al. 2006). In inverse dynamics, the spinal kinematics and external loads are presumed known, and the goal of the model is to calculate internal forces (De Zee et al. 2007). One way of estimating individual forces at a given moment is through the use of EMG recordings (Dolan et al. 2001; Gagnon et al. 2001; McGill and Norman 1985; McGill 1992; Sparto et al. 1998). Therefore the potential value of the present work in relation to inverse dynamics may be found in the approach's capacity to provide continuous kinematic data, which combined with continuous EMG data, could lead to more sophisticated models (Hansen et al. 2006). There is a recognised need for more sophisticated models (Galibarov et al. 2011), which may eventually lead to more individual specific data.

Currently however, any study conclusions based on both reductionist and systems approaches can to some degree be questioned. Due to unknown interactions we cannot be certain that reductionist conclusions are correct, and generally speaking current systems models are not well enough informed to incorporate all possible influences. The combination of QF and sEMG technologies enables the investigation of numerous lumbar biomechanical variables, from both the passive and active elements of the spine. These variables can be measured with good agreement and reliability, and therefore represent an opportunity for both systems and reductionist approaches. There would seem two logical ways forward, systems models with more detailed inputs encompassing as many influencing variables as possible (as we are now in a position to measure more of them), or a reductionist approach that focusses on the variables that have been shown to have the greatest influence on IV-RoM (if this is the outcome variable of interest), which would require further, more expansive studies. It should be noted that the current study has shown no parameter to have a particularly dominant influence over IV-RoMmax, reaffirming the fact that spinal control is multifactorial, and that reductionist approaches may potentially miss important influences.

9.9 An emphasis on kinematic parameters

With this in mind, it should be noted that the current study's results show an obvious difference in the strength of relationships found between the sEMG variables, and the kinematic variables, in terms of their influence on IV-RoMmax, sEMG variables being the

weaker. In terms of the relative importance of relationships, it has been suggested that the strength of the correlation can be divided into small, medium and large (Cohen 1992). This was adapted by Dolphens et al, who used the following criteria for the strength of r values (Dolphens et al. 2012).

Weak correlation ($0.1 \leq r < 0.3$), Moderate ($0.3 \leq r < 0.5$), Strong ($r \geq 0.5$)

The r values from Chapter 6 all represent strong correlations, but the r values for sEMG variables and IV-RoMmax (Chapter 7) are in the moderate category. Although an argument can be made regarding the appropriateness of the sEMG variables selected (i.e. alternative analytical approaches or sEMG parameters could feasibly alter the strength of the correlations), it appears from the results, that the lordosis and kinematic parameters are more influential over IV-RoMmax than the muscle activity parameters. The two cannot be easily separated, however this does highlight the importance of the passive elements. Therefore, although this study has shown that lordosis and kinematic parameters have stronger relationships than paraspinal muscle activity with IV-RoMmax, it should also be considered how other parameters other than IV-RoMmax at different levels interact with each other.

9.10 Further data analysis

Although the primary outcome variable for this study was IV-RoMmax, the relationships demonstrated suggested that intuitively, correlations would also be found between other parameters. The following section explores additional correlations that were found between the initial attainment rate, lordosis and muscle activity parameters.

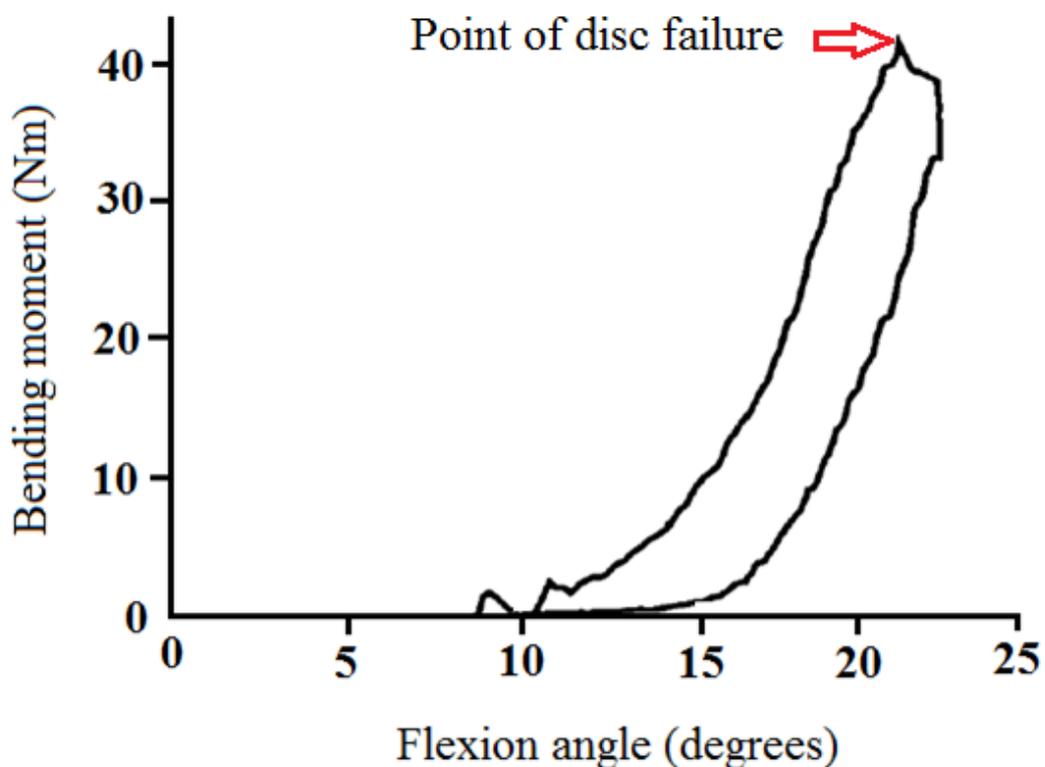
9.10.1 Lordosis versus initial attainment rate

Further analysis of the study data revealed that there was an inverse relationship between lordosis and L4-L5 initial attainment rate during flexion ($r = -.492$ $p = .038$). Therefore as lordosis increases, there is either an increase in stiffness at L4-L5, or a decrease in bending moment at that level. This may help explain why when lordosis increases, IV-RoMmax at that level decreases, and the positive relationship between L4-L5 IV-RoM max and initial attainment rate (i.e. if the segment is lax, it moves further) (Table 21).

This finding may relate to the inter-vertebral mechanics of the disc and other passive tissues. However, in an individual with a large lordosis, the L4-L5 disc, in a neutral standing position, will be loaded posteriorly, and the posterior ligaments will be in a shortened state. When the disc is compressed there is a resulting build-up of hydrostatic pressure in the nucleus

pulposus that creates a subsequent tensile stress within the annulus fibrosus (Adams 2004). This disc compression would be expected to be maximal at the neutral position, and so restraint provided by the disc due to compression would be minimal at the start of the bend, gradually increasing as the individual bends forward. It is also likely that tensile restraint produced by stretching the posterior annulus would be minimal in the early stages of flexion at this level. Indeed, Adams et al. 1994 suggests that its role in resisting movement is minimal during the early stages of sagittal bending (Adams et al. 1994) (Figure 69).

Figure 69: A loading and unloading curve for an L4-5 lumbar disc during sagittal bending (Image adapted from Adams et al. 1994)



Note: The graph shows that minimal resistance is provided by the L4-L5 disc during the initial stages of intervertebral rotation. Adams et al. (1994) also suggest that resistance provided by the posterior ligaments is also low in the early stages (Adams et al. 1994).

If this is the case, then the increase in stiffness at L4-L5 associated with an increase in lumbar curvature is not primarily due to restraint from the passive tissues, and so either the restraint provided by the trunk muscles is of more influence, the bending moment at this level is reduced, or both.

In a flatter spine, there is an increased initial attainment rate (i.e. decreased stiffness), which may also relate to the starting position of the L4-L5 vertebral segment. If the motion segment begins movement from a horizontal position, the disc compression would be most uniform initially, and resistance provided by tensioning of the posterior elements may be expected to come about sooner. This again does not seem to fit with the idea that passive structures are primarily responsible for the relationship between lordosis and initial attainment rate, implying again that the active system or changes in bending moment are more influential in the early stages of flexion. It also provides possible mechanical reasoning behind why L4-L5 is a segment commonly associated with LBP, as the morphological conditions mean that it is in immediate demand for restraint from the moment of movement onset, and this may be provided by the muscles, leading to an increased likelihood of compromised function and injury over time. To the author's knowledge there are no previous studies that have investigated the influence of lordosis on laxity measurements, and so no comparisons can be made.

9.10.2 Muscle activity changes versus initial attainment rate

Given the relationships found between TES and LES muscle activity and the IV-RoMmax of lower lumbar segments, the absence of such relationships with upper lumbar segment movement was of interest. Considering that the initiation of upper lumbar motion segment movement typically occurred during epochs 2-3 (Appendix M), it was of additional interest to investigate the correlations between initial attainment rate at these levels and the associated muscle activity during this period.

Further analysis of the kinematic and muscle activity data (Chapters 6 and 7) revealed strong positive relationships between initial attainment rate at L2-L3 and L3-L4 and TES activity during epoch 2-3 ($r = -.577$ $p = 0.012$ and $r = -.676$ $p = 0.002$ respectively). This suggests that these globally acting muscles may react to stabilise upper lumbar flexion when these segments are lax, a strategy capable of controlling the combined effects of laxity increases in two adjacent motion segments. This strategy may also partially explain the correlation found between L3-L4 initial attainment rate and L5-S1 IV-RoMmax (Table 21), as the increased stabilisation of upper lumbar segments by the globally acting TES, may provide a stable enough system to allow localised movement at L5-S1.

Cholewicki and McGill (1996) suggest that the lumbar spine is vulnerable in the neutral zone when there is minimal muscle activity (Cholewicki and McGill 1996). The results presented here would suggest however that increased global muscle activation is a compensatory

recruitment strategy when there is upper lumbar laxity (i.e. less local stabilisation, allowing local level movement, but maintaining regional stability through the action of the global muscles). Cholewicki and McGill (1996) also suggest however, that under such conditions (i.e. a relatively larger neutral zone) the activation of locally acting muscles is increased (Cholewicki and McGill 1996). This is in disagreement with this study's findings in the upper lumbar segments; as the correlations found between L2-L3 and L3-L4 initial attainment rate and an increase in TES activity, demonstrate a mechanism of globally acting control.

Exploring this idea further, a trend was also found between L4-L5 initial attainment rate and LES activity during epoch 2-3 ($r = -.412$ $p = 0.90$) (although not reaching significance). This would suggest that laxity at L4-L5 relates to increases in the locally acting LES, which is more in agreement with Cholewicki and McGill (1996). These contrasting stabilisation mechanisms appear to show that motor control strategies relating to inter-vertebral laxity can vary between lumbar levels, and therefore future investigations should not presume uniform control within the same spinal region. This further highlights the need for more research in the field of motor control that considers inter-vertebral information.

9.11 Limitations of this work

A study investigating a structure as complex as the lumbar spine, in a single plane, and using two technologies such as QF and sEMG will inherently have limitations associated with it. The following examples highlight some of these.

9.11.1 The size of the QF image field

It was not possible using the current equipment to measure the inter-vertebral movements at levels above L2 (including the thoracic region), and so theories about the kinematics that accompany muscle activity adjacent to T9 can only be speculative. Nevertheless, the information from the globally acting muscles was thought worthy of investigation. Likewise, the current QF equipment cannot image below S1, and so it is not possible to measure variables such as pelvic incidence. This would be beneficial if sagittal alignment is of interest, as the ability to adapt pelvic incidence is believed to help avoid cumulative detrimental effects associated with changes elsewhere in the kinematic chain (Rothenfluh et al. 2015).

Although it is not currently feasible to image over a wider area, future technological advances may make this possible. This would allow QF imaging beyond 60° and to full flexion, meaning the FRP could be assessed in relation to inter-vertebral kinematics. Therefore, whilst the

current protocol is also designed to standardise the bending movement in order to make comparisons between individuals more plausible, there would be benefits to alternative methodologies.

9.11.2 Securing the Pelvis

Securing the pelvis was a necessary part of the study protocol, primarily to standardise the bending movement, but also to keep the lumbar spine in the image field. Restraint does however create an arguably unnatural flexion movement pattern, as in an unconstrained forward bending movement, there is simultaneous motion of the lumbar spine, pelvis and hips. The majority of movement occurs in the lumbar spine during the initial stages of flexion, but is joined by movement of the pelvis in the mid stages, eventually shifting to a pelvic dominance when approaching full flexion (Spinelli et al. 2015). It may be feasible in future studies to use a skin surface tracking device such as the Fastrak (Abdoli-E and Stevenson 2008), Optotrak (O'Shaughnessy et al. 2013), or Flock of birds (Hsu et al. 2008; Butler et al. 2009; Bull and McGregor 2000), to provide pelvic, and thoracic kinematic measurements that would add important additional insight into the overall biomechanical picture. It may be hypothesised for example that a flatter lordosis would be associated with an increased pelvic tilt and a decreased thoracic kyphosis (Le Huec et al. 2015), which could also relate to the activity of the globally acting TES.

9.11.3 Pelvic restraint versus free bending

The weight-bearing sagittal flexion motion graphs (Appendix M) show much variation in terms of the IV-RoMmax reached by each inter-vertebral level, the steepness of the motion graph curves (attainment rate) and in the presence of phase lag (i.e. the tendency for different spinal levels to commence or end movement at different points in the trunk motion sequence). The most frequent movement pattern is demonstrated in Figure 26, where a cascade in movement can be seen from L2-L3 to L5-S1, however this was not always evident, and examples of movement initiating from lower lumbar segments, or even at the same time were seen (Appendix M). These findings reaffirm the lack of consensus in the literature regarding normal lumbar spinal movement patterns during sagittal flexion as discussed previously (section 6.1.5.1). Whilst differences in the methods used to interpret the initiation of movement may partially explain such disagreement in findings, this study has shown that there are likely many variations in normal sagittal bending movements that relate to the unique biomechanical requirements of each individual. It is also possible however that restraining the pelvis directly affects the kinematic pattern, and so could be considered a limitation.

The mean IV-RoM at each level (L2-L3 9° SD 2.7°, L3-L4 10.3° SD 1.6°, L4-L5 9.8° SD 3.9° and L5-S1 6.4° SD 3.2°) shows that using the current protocol, the least movement (as a proportion of total lumbar movement L2-S1) typically occurs at L5-S1. This finding is in agreement with Pearcy et al. (1984) who demonstrated using bi-planar radiography, similar but slightly increased ranges relative to those shown in this study (L2-L3 10°, L3-L4 12°, L4-L5 13°, L5-S1 9°), also showing the smallest movement at L5-S1. The Pearcy et al. study used eleven male volunteers of a similar age range to this study (mean 29.5 years), and whilst also restraining the pelvis (in order to keep the spine in the image field during flexion), allowed participants to fully bend (i.e. beyond 60° of trunk flexion) (Pearcy et al. 1984). This suggests that although the current protocol is designed to fully stress all lumbar motion segments, 60° of standardised flexion either does not do so, or that following the motion frame somehow minimises the movement at the base of the spine. This is difficult to interpret however, as the sample sizes in both studies are relatively small (i.e. 11 and 18), and considering L5-S1 in particular, a review of inter-vertebral motion studies showed this level to have the greatest inter-subject variability (Deitz 2011). In agreement, Li et al. (2009) reported a greater contribution to angular range from cephalic segments; however the protocol used was markedly different (Li et al. 2009), in that there was no restraint of the pelvis, and that the flexion movement was also limited to 45°. In this instance, if the cascade of motion was from cephalic to caudal (Kanayama et al. 1998), then it may be that at 45° of flexion the lower lumbar segments have not been afforded adequate opportunity to reach their maximum angular range.

These findings are in disagreement with other studies that have measured IV-RoM without restraining the pelvis. Boden (1990) suggested that in measurements of the overall angular rotation using flexion and extension radiographs, mean ranges at L5-S1 were actually greater than at other lumbar levels (Boden 1990), and when comparing the results of Boden to those of Pearcy et al., there is a clear difference in terms of IV-RoM reached by upper and lower lumbar levels, which may be in part due to the use of pelvic restraint. It seems that in free bending the upper levels rotate less, and the lower levels comparatively more, than in the current study or Pearcy et al's where the pelvis is restrained. This either suggests that the upper lumbar segments are stressed more, L5-S1 in particular is stressed less, or that 60 degrees is not enough to stress the lower lumbar segments (e.g. L5-S1) fully. This is a possible design consideration for future studies, as 60 degrees of flexion was used as it is believed to fully stress the entire lumbar spine (Dvorak et al. 1991).

Methodological differences (i.e. pelvic restraint and limited range of movement) may therefore influence the measurements. When the pelvis is restricted, the influence of the gluteal and hamstring musculature on lumbar stiffness will be different to normal free bending, as it is likely their activity will be diminished and the associated stiffening through the TLF may be smaller. The pelvic restraint also appears to relate to relatively smaller ranges at L5-S1 however, which may be explained by a locally acting effect of pressure produced by the support placed over the sacrum, producing a localised stiffening. Indeed, in a recent study investigating normative values of IV-RoM using seated flexion-extension radiographs during free bending, IV-RoM at L5-S1 was similar to other lumbar levels (mean 12.8°) (Staub et al. 2015). In a seated position with knees flexed, there would be less tension through the hamstrings, indicative perhaps of the important kinematic influence the use of a sacral support may have, possibly as a result of a direct pressure on the sacro-tuberous ligament.

9.11.4 The use of sEMG technology in isolation

The use of sEMG only provided information regarding superficial paraspinal muscles. The longissimus thoracis pars lumborum caudal inter-muscular aponeurosis attaches directly into the ilium. The LES signal adjacent to L2, predominantly from longissimus thoracis pars lumborum, is inevitably contaminated by cross talk from the multifidus at that level. The multifidus fascicle from L1 originates from the spinous process and inserts into the mammillary process of L4, and its common tendon into the mammillary process of L5, S1 and the posterior superior iliac spine. The fascicle of L2 extends from the spinous to mammillary of L5, and the common tendon into the mammillary process of S1, the posterior superior iliac spine and the iliac crest (Bogduk 2012). It is therefore feasible that LMU activity deep to the LES is also influencing the patterns. Intra-muscular needles would enable EMG recording directly from the multifidus and the longissimus lateral to the L2 spinous process, and the relative influence of each muscle on IV-RoMmax could then be assessed.

9.11.5 The determination of lordosis

The number of vertebrae evaluated when determining lumbar lordosis varies between researchers, with some using T12-S1 (Kim et al. 2006), but most commonly L1-S1 (Jackson 1994; Mao et al. 2014; Rothenfluh et al. 2015). Therefore the lordosis measurement used in this study (i.e. the angle between L2-S1) is not directly comparable with other studies, and again highlights the problem with unstandardised methodologies. Any future advancement in the QF protocol should therefore consider ways of expanding the image field to include L1-L2, and if possible the femoral heads (i.e. for pelvic incidence measurements).

9.11.6 Pre-examination activity

In the current study, there was no requirement for the participants to avoid exercise prior to their examination. Future participant information sheets should stipulate that they should not attend the gym or perform any prolonged physical activity immediately prior to the study protocol. This would reduce the possibility of muscles behaving differently due to fatigue (Descarreaux et al. 2008), or as a result of changes in proprioceptive responses for example from prolonged lengthening (Ge et al. 2011; Hendershot et al. 2011).

Chapter 10

10.1 Conclusion

This thesis was embarked upon on the basis that, furthering understanding of relationships between normal lumbar inter-vertebral kinematic and myoelectric behaviour, would assist in revealing what is normal in terms of inter-vertebral spinal mechanics.

The present study validated a methodology for measuring inter-vertebral kinematics and local myoelectric activity contemporaneously. The repeatability of the kinematic and muscle activity parameters, and the agreement and reliability of intra-subject sEMG amplitude measurements were shown to be good, and therefore supported the methodological approach. This was partly attributed to the standardisation of the data recording protocols, and therefore it is recommended that standardisation of participant movement also be incorporated into future kinematic and EMG based studies.

By combining QF and sEMG technologies, it was shown that the concurrent recording of continuous standardised lumbar inter-vertebral and muscle activity data is achievable *in vivo*. In doing so, previously unknown relationships between lordosis, inter-vertebral kinematic measures (L2-S1) and muscle activity parameters were discovered. Using L4-L5 as an example, these included a direct relationship between L4-L5 IV-RoMmax and L4-L5 initial attainment rate, and inverse relationships between L4-L5 IV-RoMmax and lordosis, L2-L3 IV-RoMmax and the LES/TES ratio. These relationships and others provide a fresh insight into mechanisms of spinal control at an inter-segmental level in pain free individuals.

The demonstration of such relationships and interactions has consequences for future LBP biomechanical studies. The IV-RoMmax at levels throughout the lumbar spine was shown to be influenced by a number of different mechanical variables, and therefore these should no longer be viewed in isolation in CNSLP investigations. Although the findings of this study were derived from a limited number of variables, they show in detail, the interactions between lumbar inter-vertebral kinematics and myoelectric activity during forward bending. These intrinsic interactions should therefore be a consideration in the design of future biomechanical studies of the lumbar spine.

10.2 Future Work

There is much scope for future investigation that builds on the work of the current study. The following section considers some of the possibilities for future work.

10.2.1 Sample size

A next step would be to confirm the relationships discovered within this study by repeating the investigations using a larger population. The sample would have to be sufficiently large for multivariate analysis, in order for the relative importance (in terms of the outcome measures variance) to be determined. The data could then also be explored using more complex analysis techniques such as structural equation modelling (Fife-Schaw 2000), or principal component analysis (Jolliffe 2002).

The study's results are also only representative of one small, young and healthy male population. Future work may therefore consider a sample including female participants and a wider age range. However, consideration would need to be given to the influence of factors such as the increased variation in soft tissue thickness (STT) associated with females, and musculoskeletal degeneration in elderly populations.

10.2.2 Exploring the influence of other mechanical variables on IV-RoMmax

Primarily to limit the complexity of the current study, it was decided to limit the number of variables selected for analysis. There are however many other variables that could also potentially influence inter-vertebral movement, and so exploring the potential for correlations between IV-RoMmax and other such parameters could be a relatively straightforward progression from the current research. The following sections review two parameters that could feasibly be measured using the current QF protocol.

10.2.2.1 Translation

During sagittal flexion there is an associated anterior translation, predominantly controlled by the action of longissimus (Bogduk 2012). In terms of spinal control therefore, the link between this movement and muscle activity is also of interest, however the parameter was omitted from the current study due to concerns over the limited size of measurements expected in a healthy population. Indeed in a group of asymptomatic volunteers, Pearcy et al. 1984, showed that translation during sagittal flexion ranged between only 1-2mm at levels L2-S1 (Pearcy et al. 1984), which considering a measurement error of almost 1mm when using the QF protocol (Breen et al. 2012) makes its use almost untenable. Translation has also previously been investigated as a parameter of instability (Teyhen 2004), however there is no general consensus regarding any link to LBP. Weiler et al. (1990) found that when

translation and rotation were combined as an 'instability factor' there was a significant difference between participants with degenerative disc disease (DDD) and healthy controls, however no such difference was found with a NSLBP group (Weiler 1990). McGregor et al. (2002) compared translation measurements during flexed and extended positions between patients with spondylolisthesis and healthy controls, and also found no significant difference (McGregor et al. 2002). The literature would suggest therefore that whilst translation should not be discounted completely as a parameter for future investigations involving specific LBP groups (e.g. patients with DDD), its use may be limited in studies that focus on NSLBP.

10.2.2.2 Disc degeneration

Disc degeneration is also believed to influence the IV-RoM of inter-vertebral motion segments (Iatridis et al. 2013), however there is a lack of consensus regarding the effects that increasing disc degeneration has on IV-RoM during flexion (Muriuki et al. 2016), and so correlations between disc degeneration and IV-RoMmax would be of interest. A loss of disc stiffness would potentially allow a vertebra greater forward rotation, and an increase would reduce it (Bogduk 1995). This relationship will itself be multifactorial however, as under loading, other active and passive system parameters will inevitably contribute. In the absence of muscle activation, it has been demonstrated in porcine spines that damage to the disc alone does not appear to have a significant effect on inter-vertebral rotation ranges, and that other aspects of the control system such as facet joint capsules and longitudinal ligaments may have a greater influence (Kaigle 1995), whilst other studies contend that the ligamentum flavum, supraspinous and interspinous ligaments provide the most resistance to flexion (Von Forell and Bowden 2014).

These conclusions all conflict at least in part, to those of Adams (1980), who conducted a cadaveric study investigating the resistance to flexion of various lumbar joint elements (Adams et al. 1980). Their study also highlighted the importance of passive structures in the control of forward bending, but suggests that lumbar inter-vertebral rotation is resisted primarily by the ligaments of the facet joint capsules and by the disc (Adams et al. 1980). To assess the direct role of ligaments *in vivo* would be a challenge, however it is relatively simple to assess the state of the disc using radiographs (Kellgren and Lawrence 1958) or MRI (Tan 2000). A future study could therefore investigate the relationship between disc degeneration and IV-RoMmax *in vivo*, as this has not been investigated under the influence of muscle activity and loading.

10.2.2.3 Motion sharing variability (MSV)

A kinematic variable known as motion share variability (MSV) is under development at the IMRCI and is a representation of the evenness of segmental motion sharing during bending. To calculate this variable the average distance between the levels that received most and least motion during the movement is taken. The square root of the variance of these distances is taken for every data point over the sequence is calculated providing the MSV. The MSV has been shown to relate to CNSLBP, i.e. more uneven inter-vertebral motion sharing is associated with CNSLBP (Breen and Breen 2017 accepted).

It was theorised that increased muscle activity may be a possible mechanism to stabilise motion segments during forward bending, something that has never been demonstrated before *in vivo*. In order to investigate this theory, the kinematic data from this thesis was used to calculate the MSV for each participant, and correlations between MSV and the mean normalised EMG during flexion of TES, LES and LMU were explored. EMG activity in all three muscle groups was found to be moderately and significantly correlated with MSV. All correlations were negative (LMU $r=-0.54$, $p<0.05$; LES $r=-0.069$, $p<0.01$; TES $r=-0.54$, $p<0.05$). Further nonparametric regression analysis (tau) showed a weak dependence of MSV on sEMG activity for all three muscles groups (LMU -0.38 , $p<0.05$, LES -0.47 $P<0.01$, TES -0.34 $p=0.05$). These results indicate that increased muscle activity may be a stabilising mechanism (i.e. provides increased damping) that limits irregular rotational displacements during forward bending. Replication using a larger sample is recommended to reaffirm these findings.

10.2.3 Other future research possibilities

10.2.3.1 Reference ranges

The determination of reference ranges for kinematic measurements such as IV-RoMmax is already underway in the form of the creation of a normative database (Section 4.2.1). Determining whether mechanical parameters found in LBP patient groups are within normal reference ranges, would make associations between LBP and mechanical parameters much clearer, enabling physicians to establish whether or not the LBP problem is likely to have predominantly mechanical components, and therefore stratify more dependably, candidates likely to respond to CBT based treatment programmes (Hill et al. 2011). In terms of mechanical LBP research, one of the biggest obstacles is the heterogeneity within LBP groups. This is especially problematic in terms of EMG measurements, although this study

has demonstrated that clinically useful reference ranges for EMG variables may be more likely if standardisation of movements can sufficiently minimise participant variability. The determination of reference ranges for sEMG amplitude parameters, is an area that therefore warrants further exploration.

If such parameters are ever to be used as outcome measures in longitudinal studies involving LBP populations, then normal intra-subject variation will also have to be relatively small, so that any changes post treatment can be clearly designated as a result of the treatment itself, and not simply as a result of the normal variation between baseline and follow up measurements. This threshold is known as the minimum detectable change (MDC) and estimates the intra-subject parameter change that would be found 95% of the time. Unfortunately due to the inherently large heterogeneity in parameters such as IV-RoM and EMG amplitude, currently MDC's are typically so large, that changes resulting directly from the treatment intervention would be difficult to interpret. Although it may be expected that MDCs for both kinematic and muscle activity parameters would be too large to be of clinical value, this needs to be confirmed by further studies.

10.2.3.2 Pain developer groups and sub-grouping

As discussed previously, it is possible that kinematic, morphological and muscle activity parameters that are found in healthy participants, may feasibly be pre-disposing factors for LBP development. To test this theory would require an investigation into how repeated flexion affects individuals who demonstrated particular kinematic behaviours (e.g. large or small IV-RoMmax at L4-L5 during sagittal flexion), a possible hypothesis being that individuals demonstrating smaller IV-RoMmax's at L4-L5 will have larger LES/TES ratios and develop lumbar pain (i.e. possibly due to repetitive strain of the LES) quicker during repeated flexion than those with larger L4-L5 ranges. It would also be of interest to investigate the inter-vertebral biomechanics of LBP patients allocated to the O'Sullivan et al. (2005) sub-groups, in particular those in the flexion pain provocation group. This would allow the exploration of relationships between biomechanical parameters and LBP at a sub-grouped and inter-vertebral level.

Collaboration with Dr Rebecca Hemming (A member of the Arthritis Research UK Biomechanics and Bioengineering Centre at the University of Cardiff). A collaboration is planned to use the protocol developed in this thesis to explore the kinematics and muscle activity of CNSLBP patients that have been sub-grouped in accordance with the O'Sullivan (2005) sub-grouping system. This approach has been used in many studies that have

attempted to classify NSLBP (Dankaerts et al. 2006; 2009; Hemming et al. 2015), yet it remains unclear as to the underlying mechanical reasons for the symptoms that determine group allocation. An exploration of these sub-groups using inter-vertebral level information is planned to investigate the possible mechanical reasons why such sub-groups exist. Using the lordosis versus IV-RoMmax results as an example, sub-grouped flexion and extension pattern patients tend to occupy opposing ends of the lumbar posture spectrum (Dankaerts et al. 2009). Flexion aggravated patients more commonly have a kyphotic lumbar spine, which from this study is associated with more movement at L4-5, and less at L2-L4, which may contribute to pain generation at these lumbar levels. Inter-vertebral information may therefore help provide more insight into the mechanism of pain in such groups.

10.2.3.3 Plane of investigation

The scope of this study was limited to sagittal plane flexion and provides no information about the return phase or sagittal extension from neutral. Future studies may possibly include these, and also consider the coronal plane, although a preliminary study concluded that relationships would most likely be found between kinematic and EMG parameters in the sagittal plane (Chapter 3). Further consideration of the appropriate muscle groups to investigate would be necessary in each case.

10.2.3.4 Removing the contamination of gravity and loading

The methodology of this study was designed to minimise possible confounding factors such as variations in load and disc stiffness. Participants were all aged between 20 and 40, had a BMI < 30, with no history of low back pain, and the likelihood of disc degeneration or disc injury having an influence on the kinematic patterns was therefore reduced. However, a limitation of this study was an inability to account for the axial loads acting on each motion segment during bending, and so it was not possible to know how such forces may have influenced the findings. A future study therefore, may investigate what effect muscle activity 'alone' has on the lumbar spinal kinematics. In order to do so, a protocol would be required that removes the confounding influence of loading, and in an ideal testing environment therefore, the effects of gravity and changes in both passive and active spinal properties due to loading would be removed. One possible solution would be to conduct both a passive and an active recumbent sequence, as any difference in the kinematic behaviour of the spine between the two examinations, would theoretical be a result of the muscle activity alone. This in itself is not a perfect solution, as muscles will behave differently under loading, and there would be issues regarding friction of the motion table during the active bend. In terms

of the muscles under investigation, it would also not be feasible to study the paraspinals during recumbent flexion, as an individual's weight-bearing eccentric lowering function would be replaced by the agonist activity of the abdominal muscles. A solution would be to measure the activity of the extensors during the return phase from full flexion to neutral, as in theory, passive recumbent, active recumbent and weight-bearing sequences could then all be compared.

10.2.3.5 Maximising sEMG data collection

Traditional sEMG techniques such as those used in this study, can be limited in terms of the number of recording sites achievable within a region of muscle, primarily due to the size of individual electrodes. A possible solution to this problem is the use of EMG arrays (Finneran et al. 2003; Hu et al. 2010). Such arrays are capable of producing continuous topographical representations of the distribution of sEMG signals, and are therefore a novel way of showing muscle contraction patterns. Their expansive nature (i.e. covering large parts of the back) would not cause an issue for the concurrent use with QF based modalities; however it may be problematic if required for use in combination with devices that also require attachment to the skin. If electrode arrays were incorporated into methodologies such as those used in this study, they could help provide a more detailed inter-vertebral level specific insight into relationships with the lumbar paraspinal muscle activity.

10.2.3.6 Investigating other muscles

The scope of this study limited focus to the lumbar paraspinal muscles. The longissimus and multifidus are not the only muscles involved in the control of spinal movement during bending however, and so not all relevant muscles were investigated. The roles of other regional muscles such as the TrA, diaphragm, psoas and QL, are all of potential importance, and so the investigation of relationships between IV-RoMmax and such muscles would also be of interest. It would also be beneficial to measure muscle activity from several different groups concurrently, in order to investigate co-contraction strategies.

It has also been suggested that the lumbar spine should not be viewed in isolation from the lower limbs due to the inter-dependence between them (McGregor and Hukins 2009). The measurement of muscle activity in the gluteals, hamstrings, and quadriceps would also therefore be of interest; however their inclusion would depend on an adaptation of the current protocol to incorporate the movement of the hips and pelvis.

10.2.3.7 Further exploration of the Flexion relaxation Phenomenon (FRP)

As described previously, the FRP literature is typically limited due to the measurement of regional kinematics, and the recording of EMG from single muscle sites. This level of detail is inadequate if a better insight into the relationships between inter-segmental movement and muscle deactivation (i.e. the mechanism of reflex muscle inhibition) is to be gained. The combined use of EMG and fluoroscopy provides the possibility to conduct research that considers the activation patterns of numerous muscles in relation to specific lumbar motion segments. As the FRP has been shown to occur near the end range of forward flexion, the QF protocols would have to be adapted to allow the participants to reach full flexion (currently limited to 60°), which is only problematic in terms of keeping the lumbar spine within the image field. If these problems can be addressed, the opportunity to investigate the FRP of paraspinal musculature (including the use of needle EMG to record from the deep multifidus) should be taken.

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Appendices

Appendix A: Research Dissemination

Publications

Du Rose, A., and Breen, A., 2015. Influence of paraspinal muscle activity on lumbar inter-vertebral flexion rotation range. *The Spine Journal*, 15 (3 Suppl 2), S57-S58.

Du Rose, A., and Breen, A., 2016. Relationships between paraspinal muscle activity and lumbar inter-vertebral range of motion. *Healthcare*, 4 (4)¹⁴.

Du Rose, A., and Breen, A., 2016. Relationships between lumbar inter-vertebral motion and lordosis in healthy adult males: A cross sectional cohort study. *BMC Musculoskeletal Disorders*, 17 (121).

Breen A, Breen Ax, Mellor F, du Rose A, Bentman S. Control of intervertebral Motion: A Critical Element of Spine Control and a Target for Treatment. *Journal of Orthopaedic and Sports Physical Therapy*, (submitted).

Oral presentations

Du Rose, A. and Breen, A., 2015. Influence of paraspinal muscle activity on lumbar inter-vertebral flexion rotation range. British Association of Spine Surgeons Annual Conference, The Assembly Rooms, Bath March 2015.

Du Rose, A. and Breen, A., 2015. The influence of motor control on lumbar inter-vertebral kinematics during forward bending. Research Seminar, Department of Rehabilitation Sciences and Physiotherapy, Ghent University May 2015.

Du Rose, A. and Breen, A., 2015. Paraspinal muscle activity and morphology and their effects on lumbar inter-vertebral kinematics. Research Seminar, Department of Surgical Specialities, College of Osteopathic Medicine, Michigan State University June 2015.

Du Rose, A., 2016. The Society for Back Pain Research (SBPR) Travel Fellowship Review: The normal biomechanics of the lumbar spine: a quantitative fluoroscopy and electromyography study. BritSpine Conference, Nottingham Conference Centre, Nottingham April 2016.

Du Rose, A., 2016. Normal biomechanics of the lumbar spine: an inter-vertebral level insight. Research Seminar, College of Biomedical and Life Sciences, Cardiff University, Cardiff October 2016.

¹⁴ **Note: In response to the open access publication in Healthcare (see above) and in recognition of the novel combination of sEMG and QF, the author was invited to submit an entry into the 14th Annual Delsys Prize (A prize designed to promote innovation in the use of electromyography).**

Poster presentations

Du Rose, A. and Breen, A., 2015. The effect of paraspinal myoelectrical activity on the range of lumbar inter-vertebral motion. WFC Congress and ECU Convention, Hilton Athens Hotel, Greece May 2015.

Appendix B: Anatomical planes of movement

Movements of the human body can be divided into 3 planes of motion (Figure 70). The main study protocol required participant movement in the sagittal plane (a plane passing through the body anterior to posterior) referred to in the study as sagittal flexion and extension (Figure 71). Movements in the coronal plane (a plane dividing the anterior and posterior body) are referred to as lateral bending (Figure 71) and those in the transverse plane (a plane dividing the superior and inferior body) as axial rotation.

Figure 70: Anatomical planes of movement

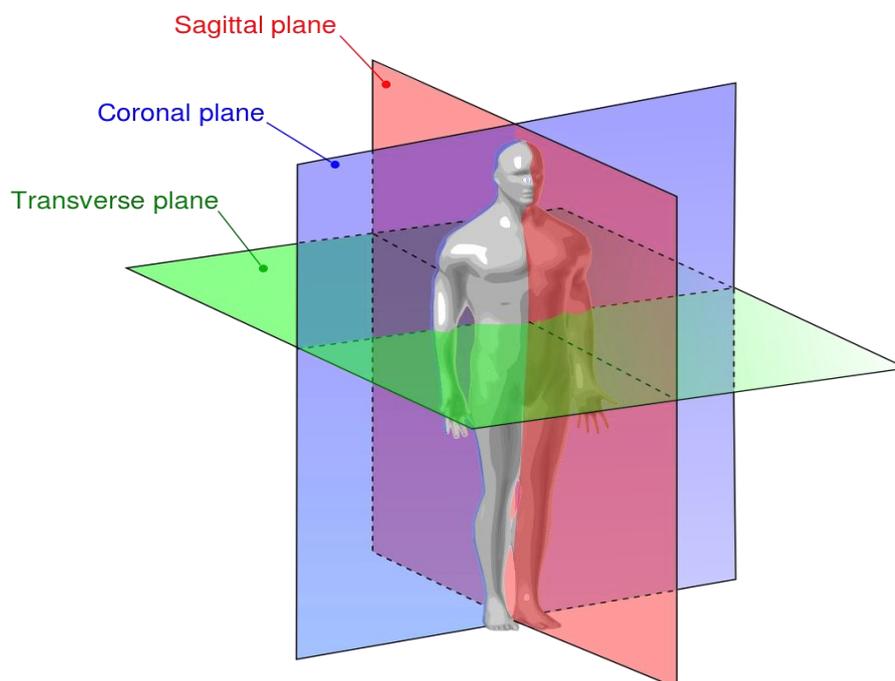
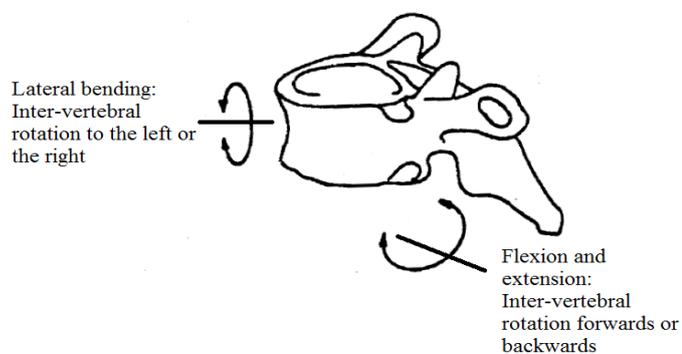


Figure 71: Sagittal and coronal plane vertebral rotations



Appendix C: Participant information sheets

Information sheet for the sEMG reliability and agreement study



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PARTICIPANT INFORMATION SHEET

Study Title: Surface Electromyography of the Lumbar Paraspinal Muscles during the Weight Bearing Sagittal Plane OSMIA Acquisition Procedure: An intra-subject repeatability study.

You are being invited to take part in this research study. Before you decide whether to do so, it is important that you understand why the research is being done and what it will involve. Please take time to read this Information Sheet carefully and discuss it with the Principal Investigator if you wish.

Please ask if there is anything that is not clear or if you would like more information. Take time to decide whether or not you wish to take part.

What is the purpose of the study?

This study is being conducted to find out if the electrical activity generated by the back muscles, is the same each time we bend. If it is, then it should be possible to use it together with motion x-ray images to find out how function of the back muscles and the movement of bones in the back, normally relate to each other. This will further our understanding of spinal function, and will hopefully become a useful tool in the investigation of disorders affecting the lower back in the future. This study is part of the development process for a larger study, and its findings will help inform decisions regarding the larger study's design.

Who is eligible for the study?

Males between the ages of 20 and 40 who are;

- able to understand written information

- willing to participate and able to freely give informed consent
- have a body-mass index under 30
- without any history of back pain that has prevented normal activity for at least 1 day in the previous year

Do I have to take part in this study?

Participation in the study is completely voluntary. There is no penalty for declining and you are free to withdraw your consent at any time without explanation.

How is the study being done?

The testing process will take about 30 minutes per session, and participants will be asked to attend 2 separate sessions over a 1 week period with a minimum of a 2 day break between each session. All testing will take place in the x-ray department of the AECC clinic. **HOWEVER, YOU WILL NOT RECEIVE ANY X-RAYS.**

Firstly, we will measure your height, weight and skin fold thickness using calipers. Next, the skin over your lower back region will be prepared for sEMG electrodes (which are self-adhesive pads). This should not be painful, but it does involve abrading the skin lightly with a cloth, cleaning with an alcohol swab and if necessary shaving the area. Fifteen electrodes will then be placed on the skin of your mid to lower back.

The actual test involves standing next to a motorised motion frame with your pelvis held still by a strap. The frame will guide you as you bend your low back 60° forwards, and back to your original position. You will then also be asked to bend as far forward as you can. This process will be repeated 4 times at each session. In order to ensure the accuracy of positioning the electrodes over subsequent sessions, an outline will be traced around each electrode with an indelible marker.

In order to get an idea of the largest signal your back muscles can produce, you will be asked to lie face down on a cushioned bench with your hands behind your head. Your pelvis and legs will be supported and you will be asked to raise your upper body off the bench and hold this position for 5 seconds. You will be asked to repeat this process 3 times, but only on your initial session.

(Participants with long hair will be provided with bands to tie their hair back.)

Will my taking part in the study be kept confidential?

All participants will be allocated numbers. Data will be stored analysed and published anonymously using these numbers. The data will be stored on a password protected computer.

What if I agree and then change my mind?

Participants will be free to withdraw from the study at any time without prejudice. If a participant decides to withdraw following data collection, already collected data will still be used in the study, but no further data will be collected.

What will happen to the results of the research study?

These will eventually be published as part of my PhD thesis at Bournemouth University. They may also be presented at scientific conferences, and published in a scientific journal.

Who is organising and funding the research?

The research is organised by the Institute for Musculoskeletal Research and Clinical Implementation (IMRCI) and funded by the European Chiropractors' Union Research Fund (ECURF).

Who has reviewed this study?

The protocols for this study have been reviewed by Professor Alan Breen, DC PhD IPEM Director of Research at the Institute for Musculoskeletal Research and Clinical Implementation (IMRCI). This study has also been reviewed and approved by the Ethics Sub-Committee of the Anglo-European College of Chiropractic.

What are the risks of taking part in the study?

There are no significant risks in taking part in this study. However you should be prepared for possible temporary minor red skin marks due to the electrode preparation and removal process. There is a very small chance that a participant may have an allergy to, or become irritated by the gel electrodes used in the study. There is also the possibility that the forward or return bending may cause physical discomfort. If this occurs you can cease the testing at any time. You will be given an emergency stop button for this purpose.
(PARTICIPANTS WILL NOT BE SUBJECT TO ANY RADIATION)

What if there is a problem?

If you have any complaints about the way in which this study has been, or is being conducted or wish to comment in any other way, please contact the Principal Investigator using the details below.

Principal Investigator:

Alister du Rose

Telephone: 01202 436353

Anglo European College of Chiropractic

Email: adurose@aecc.ac.uk

13-15 Parkwood Road

Bournemouth

Information for volunteers:

Low back inter-vertebral motion patterns in healthy adults: Reference ranges and reliability.

I would like to invite you take part in this research study. Before you decide it is important for you to understand why the research is being done and what it would involve for you. **My contact details are at the end of this information and I would be happy to answer any questions you may have.**

This information leaflet will:

1. Outline the purpose of the research.
2. Explain why you have received this leaflet.
3. Describe what happens next.
4. Describe what will happen if you decide to participate.
5. Clarify the risks and benefits to you of taking part.
6. Inform you about confidentiality and data protection.
7. Describe what to do if you have a problem
8. Explain what will happen to the results of this research
9. Tell you who is funding the research
10. State who has reviewed the study
11. Give contact details for the clinical investigator so you can ask any further questions.

1. Purpose.

This study is being conducted to establish a database of the normal mechanics of the low back in people without back pain. This is so there will be a reference for patients being investigated for mechanical pain to help with treatment. A lot of treatment for back pain is based on improving the functional mechanics of the spine, which is reflected in the patterns of inter-vertebral motion. However, until now it has been impossible to measure these in living people without penetrating the skin. Quantitative Fluoroscopy is an X-ray video method doing this which was invented and developed at the Anglo-European College of Chiropractic (AECC), where it has been called 'OSMIA' (Objective Spinal Motion Imaging Assessment). This research is to determine the limits of normal inter-vertebral motion so that clinicians who use it in the future will be able to interpret its results and researchers will be able to test the ability of treatments to improve spinal mechanics in living people.

A small number of volunteers will also be asked if they would like to participate in a sub-group study that will investigate the activity of the muscles in the lower back during the OSMIA procedure. This sub-group study is being conducted to explore the relationships between the normal mechanics of the low back and the concurrent activity of the low back muscles. This study uses a technique called surface electromyography (sEMG) which measures the electrical activity produced by your muscles as they contract.

2. Why Have I Received this Leaflet?

You have received this leaflet because you are aged between 21 and 70 years and you replied to an email or advertisement in the College asking for volunteers who fit the inclusion criteria and who would like to take part in this research study. This leaflet will explain the research in further detail.

3. What Happens Next?

After at least a week, I will contact you to ask if you are still interested in taking part. **I am happy to answer any questions you may have but it is entirely your decision whether or not you decide to join the study. You are free to refuse to participate or withdraw at any time prior to the taking of the x-ray video without giving a reason (see Confidentiality and Data Protection p6).**

4. What Will Happen if I Decide to Participate?

If you take part in this research your name, gender, age, height and weight, address and telephone number and email will be stored on a password protected database. You will be invited to attend the x-ray department at a time convenient to you. I will meet and go through this Information Leaflet with you and explain the examination. If you are happy to proceed you will be asked to sign two consent forms, one of which will be for you to keep.

You will then be allocated to have either a forward-backward bending examination or a side-bending one. You may also be asked to agree to have an additional one in 6 weeks time. ***If you are allocated to the forward-backward bending examination, you may also be asked if you are willing to participate in the sEMG sub-group study.*** You will then be shown to a changing room and asked to change into a gown. We will then show you how the equipment works. OSMIA uses specially designed motion tables and low dose video x-rays. You can view this in advance on the College website if you wish. (<http://www.aecc.ac.uk/imrci/osmia.aspx>). The tables rotate so that the upper half of the body moves slowly from side to side.

One table is for lying examinations and the other is for standing.

First you will be asked to lie on one motion table. The upper half of the table will swing slowly from side to side and video x-rays will be taken showing the movement of your vertebrae as you bend. Then you will be asked to move to an upright motion table and stand against it. Again the table will slowly swing while you bend, following a moving arm rest, while the x-rays are taken simultaneously. Before we take the x-rays we will find the range of bending that you are comfortable with.

sEMG sub-group study only: If you have agreed to participate in this study there will be some additions to the procedure which will add approximately 15 minutes to your visit. At the point between the lying and standing examinations, the skin over your lower back region will be prepared for sEMG electrodes (which are self-adhesive pads). This should not be painful, but it does involve abrading the skin lightly with a cloth, cleaning with an alcohol swab and if necessary shaving the area. We will also measure the thickness of a fold of your skin at 3 different levels, both on the left and right sides of your lower back. Fifteen electrodes and three small wireless transmitters will then be placed on the skin of your mid to lower back. This will enable the

measurement of your back muscle activity during the standing phase of the examination. In order to get an indication of the maximum activity your low back muscles can produce, at the end of the examination procedure you will then be asked to lie prone on a padded bench with your hands behind your head. You will then be asked to raise your upper body off the couch and hold this position for 5 seconds whilst your legs and pelvis are supported. Finally, when testing is completed you will also be asked a simple question about your experience of the examination procedure.

During examinations, your lower abdomen will be covered with a lead apron to protect the reproductive organs. You will also be provided with a button that will stop the table should you begin to feel pain or discomfort. **The whole procedure, including filling in a form, will take no more than 30 minutes. (If you are participating in the sEMG sub-group study the whole procedure will take no more than 45 minutes).** We may then make an appointment for you to have the same examination 6 weeks later. **(If you have volunteered to be in the sEMG sub-group, this second examination will not include sEMG.)** Before doing it we will check to make sure you have had no disabling back pain since the first examination. If you have, we will not proceed with the second examination.

5. Risks and Benefits of Participating.

This examination uses x-rays. Therefore it is important you understand the risks and benefits of taking part. **Females please note, x-rays may harm an unborn child. It is therefore vital that you inform us beforehand if you are pregnant or suspect you might be.**

The radiation dose from the examination is roughly the same amount of naturally occurring background radiation you would receive in the UK over a 17 month period. Experts agree that it is very difficult to determine the risk of inducing cancer from such low doses, however it is estimated that there is a **1 in 8,000 – 1 in 13,000 extra chance of getting cancer from this examination. (This is in addition to the quoted 1 in 3 natural lifetime risk of you contracting cancer throughout your lifespan.)** You may wish to consider this risk in relation to some more familiar events as in the table on page 5. There is no direct benefit to you from the radiation dose; however, the risk is seen as minimal.

<u>Some familiar risks (Sedgwick and Hall 2003)</u>	<u>Chance they will happen</u>
Getting three balls in the UK national lottery	1 in 11
Needing emergency treatment in the next year after being injured by a can, bottle, or jar	1 in 100
Death by an accident at home	1 in 7100
Getting five balls in the UK national lottery	1 in 11 098
Death by an accident at work	1 in 40 000
Death playing soccer	1 in 50 000
Death by murder	1 in 100 000
Being hit in your home by a crashing aeroplane	1 in 250 000

"Teaching medical students and doctors how to communicate risk." *BMJ* 327(7417): 694-695.

There is also a chance that an 'incidental' finding will be seen on your video x-ray. An incidental finding is one that is discovered unintentionally. To date, 60 patients have undergone this examination and there have been no significant incidental findings. I will be reviewing all video x-rays and in the event of an incidental finding you will be referred to your GP if that is what you would like.

Such detection has the benefit of starting treatment early but in a small number of cases may have implications for future employment and insurance. There may be no overall benefit to you from this study but the information I receive might help improve the diagnosis of patients with NSLBP. If you are a student or faculty member you will probably find the experience educational and you will be able to watch the movement of your lumbar vertebrae and see a report on it.

SEMG sub-group study only: If you agree to participate in the sEMG sub-study, there are no significant additional risks. You may however experience minor discomfort during calliper measurements of skin fold thickness, as a result of skin preparation prior to electrode attachment, or as a result of electrode removal, any of which could possibly result in transient minor red marks on the skin surface. There is also a very slight risk of allergy or irritation caused by the adhesive on the electrodes.

6. Confidentiality and Data Protection

Ethical and legal practice will be followed with respect to any information obtained from you in this study. Your details will be kept on a password protected database until all the volunteers have been recruited. After this, all identifying details will be destroyed. If you enter the study your GP will be informed and you will be asked to provide your GP's details (name and address) on the consent form. Following review of your video x-rays all of your data will be anonymised so you cannot be identified.

Consequently, you will not be able to withdraw from the study once your data have been collected. This does not affect your right to withdraw from the study prior to, or during data collection. Your anonymised data will also be retained indefinitely for use in further studies.

7. What if there is a problem?

If you have a concern about any aspect of the study you should speak to me in the first instance and I will do my best to answer your questions. If you remain unhappy and wish to complain formally you can do this by contacting Professor Thiel, the Chief Executive of the AECC.

In the event that something does go wrong and you are harmed during the research

due to someone's negligence, you may have grounds for legal action for compensation against the AECC but you may have to pay your own legal costs.

8. What will happen to the results of this study?

The results from this study will be anonymised, collated and analysed and published in scientific journals as a reference database. It will also be presented at international conferences such as that of the Society for Back Pain Research. Some data will be referred to on the AECC website (www.aecc.ac.uk). You are welcome to keep up to date with the study's progress by periodically checking the website, or by contacting me at any time; my details are at the end of this leaflet.

sEMG sub-group study only: The results from this study may also be published as part of a PhD thesis at Bournemouth University.

9. Who is funding the research?

This research is being funded by the Anglo-European College of Chiropractic. ***The sEMG sub-group study is funded by the European Chiropractors' Union Research Fund (ECURF).***

10. Who has reviewed the study?

This research has been extensively reviewed by a spinal surgeon, a radiologist, a statistician a medical physics expert, a bioengineer an ergonomist, a chief superintendent radiographer and the South West 3 Research Ethics Committee (REC Reference10/H0106/65). ***The sEMG sub-group study has been reviewed by the Research Ethics Subcommittee of the AECC, and the AECC Patient and Public Involvement (PPI) group.***

11. Further information and contact details

Professor Alan Breen

Director

Institute for Musculoskeletal Research and Clinical Implementation,
Anglo-European College of Chiropractic.

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Tel: 01202 436275

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Contact for sEMG sub-group study:

Mr Alister du Rose

Doctoral Research Fellow

Institute for Musculoskeletal Research and Clinical Implementation

Anglo-European College of Chiropractic.

13-15 Parkwood Road

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Appendix D: Participant consent form for preliminary sEMG studies

INFORMED CONSENT FORM

Title of Study:

Surface Electromyography of the Lumbar Paraspinal Muscles during the Weight Bearing Sagittal Plane OSMIA Acquisition Procedure: An intra-subject repeatability study.

- I confirm I have read and understood the Information Sheet for this study.
- I understand that my participation is voluntary and that I am free to withdraw at any time, and without giving a reason.
- I understand that the information collected about me will be kept confidential and treated at all times in an anonymous manner.
- I agree to take part in the study.

Participant name (please print) Signature Date

Name of person taking consent Signature Date

One copy for participant, one copy for file

Contact Information (Researcher):

Alister du Rose

Anglo European College of Chiropractic

13-15 Parkwood Road

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BH52DF

Email adurose@aecc.ac.uk

Telephone 01202 436353

Appendix E: Copies of ethical approval letters



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RESEARCH & GRADUATE STUDIES

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Alister du Rose
AFCC
Bournemouth

13 June 2013

Dear Alister

Re: Surface Electromyography (sEMG) of the lumbar and thoracic paravertebral muscles during the weight bearing sagittal plane Objective Spinal Motion Imaging Assessment (OSMIA) acquisition procedure: An intra-subject repeatability study (Ref: E51/05/13)

Thank you for submitting an application for ethics approval for conducting the above study.

Following receipt of the revised Information Sheet, the application (as attached) has now been granted approval by the AECC Research Ethics Sub-Committee.

May I take this opportunity to wish you every success in the study.

Yours sincerely

Professor J E Bolton, PhD, MA Ed, FHEA, FCC(Hon), FBCA, FFFAC
Chair, AECC Research Ethics Sub-Committee

c.c. Professor Alan Breen

Company Number: 11414614
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NRES Committee South West - Central Bristol

Bristol Research Ethics Committee Centre
Whitefriars
Level 3, Block B
Lewin's Mead
Bristol BS1 2NT
Email: nrescommittee.southwest-bristol@nhs.net

22 November 2013

Professor Alan C Breen
Professor of Musculoskeletal Health Care
Anglo-European College of Chiropractic
13-15 Parkwood Road
Bournemouth
BH5 2DF

Dear Professor Breen

Study title: **Characteristics of lumbar spine inter-vertebral kinematics in healthy adults and their reproducibility over time: A standardised reference and reliability study for future explanatory trials of mechanical interventions for non-specific back pain**

REC reference: 10/H0106/65
Amendment number: 1 – 14/10/2013
Amendment date: 14 October 2013
IRAS project ID: 61592

The above amendment was reviewed on 15 November 2013 by the Sub-Committee in correspondence.

Ethical opinion

The members of the Committee taking part in the review gave a favourable ethical opinion of the amendment on the basis described in the notice of amendment form and supporting documentation.

Approved documents

The documents reviewed and approved at the meeting were:

Document	Version	Date
Notice of Substantial Amendment (non-CTIMPs)		14 October 2013
Participant Information Sheet: Volunteers	3.0	06 November 2013
Covering Letter		06 November 2013
Protocol	4.0	06 November 2013
Pre-Study Form	3.0	04 November 2013

Research Ethics Committee established by the Health Research Authority

Membership of the Committee

The members of the Committee who took part in the review are listed on the attached sheet.

R&D approval

All investigators and research collaborators in the NHS should notify the R&D office for the relevant NHS care organisation of this amendment and check whether it affects R&D approval of the research.

Statement of compliance

The Committee is constituted in accordance with the Governance Arrangements for Research Ethics Committees and complies fully with the Standard Operating Procedures for Research Ethics Committees in the UK.

We are pleased to welcome researchers and R & D staff at our NRES committee members' training days – see details at <http://www.hra.nhs.uk/hra-training/>

10/H0106/65:	Please quote this number on all correspondence
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Yours sincerely



Dr Pamela Cairns (Chair)

Chair

E-mail: nrescommittee.southwest-bristol@nhs.net

Enclosures: List of names and professions of members who took part in the review

Copy to:

Prof Haymo Thiel, Anglo-European College of Chiropractic

NRES Committee South West - Central Bristol

Attendance at Sub-Committee of the REC meeting on 15 November 2013

Name	Profession	Capacity
Dr Pamela Cairns (Chair)	Consultant Neonatologist	Expert
Mrs Angela Clarke	(Ex-social worker)	Lay Plus

Also in attendance:

Name	Position (or reason for attending)
Miss Lidia Gonzalez	REC Admin Assistant

Appendix F: Summary of outcomes from PPI group meetings

Key points that were raised:

By participant

- Participants need to be re-assured that the bending will not cause the electrodes to fall off. The patient felt that because they feel like they were moving slightly it can alter how they were holding themselves, which could affect the muscle activity.
- When carrying out the sub-MVC, it would be beneficial to have a bench with a head piece for comfort and avoiding a disproportionate measurement from one side due to head positioning e.g. looking one way or the other.
- Practice repetitions would be helpful for the participant to get more used to the protocol.

By observers

- A concern was raised over the weight of the transmitters and whether this would affect the findings.
- A concern was raised over the protocol if used with LBP patients, particularly the requirement to fully flex the spine. It was explained that this was not a requirement for this study, and that all current study participants were healthy.
- The question of age was raised. The author explained the limitations, pros and cons.
- Are obese participants going to fit in the equipment/ affect the signal? The author explained the inclusion criteria of BMI<30 does account for this to some degree, although the criteria may need to be tightened in future studies.
- There was concern that some of the information given to the participant may be repeated and that some of the headings in the information sheet may be excluded. It was explained that the headings are there usually at the request of the ethical institutions themselves, and thus not under our control.
- It was suggested in order to improve participant comfort that they be given something to rest on during the electrode application stage (whilst in flexion).
- If the participant decided at a late stage that they did not want the electrodes applied, it was explained that they could still continue with the QF part of the study, without the sEMG.
- Change to the wording of the information sheet: In the event of something going wrong to IN THE UNLIKELY EVENT.
- Point made about pre-warning the participant about a question afterwards that relates to pain, as this may instil the idea of pain in them and affect the activity. The problem could

be solved by exchanging the specific question to ...you will be asked to complete a short questionnaire

- Dr Val (name change) needs to be updated on the information sheets.
- The issue of chaperones was raised as female patients may be asked to remove their bras for the testing procedure. It was decided to err on the side of caution that chaperones should be offered in the event of a female radiographer not being present, and the participant should be asked to make it clear that this is a requirement with good notice before the session.
- There should be a mention of how many people will be in the room.
- Females will be gowned but will have opening at the back.

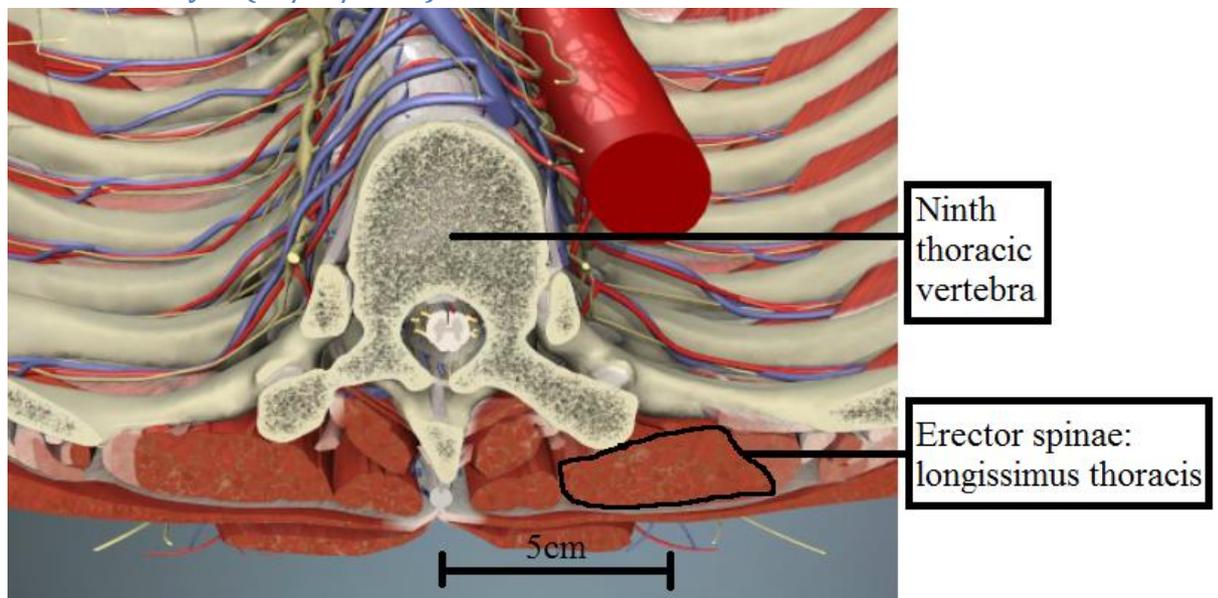
Timings

- Additional time allocated due to electrode application, skin preparation and sMVC contraction was deemed appropriate.

Appendix H: Cross sections of the spine at the levels of T9, L2 and L5

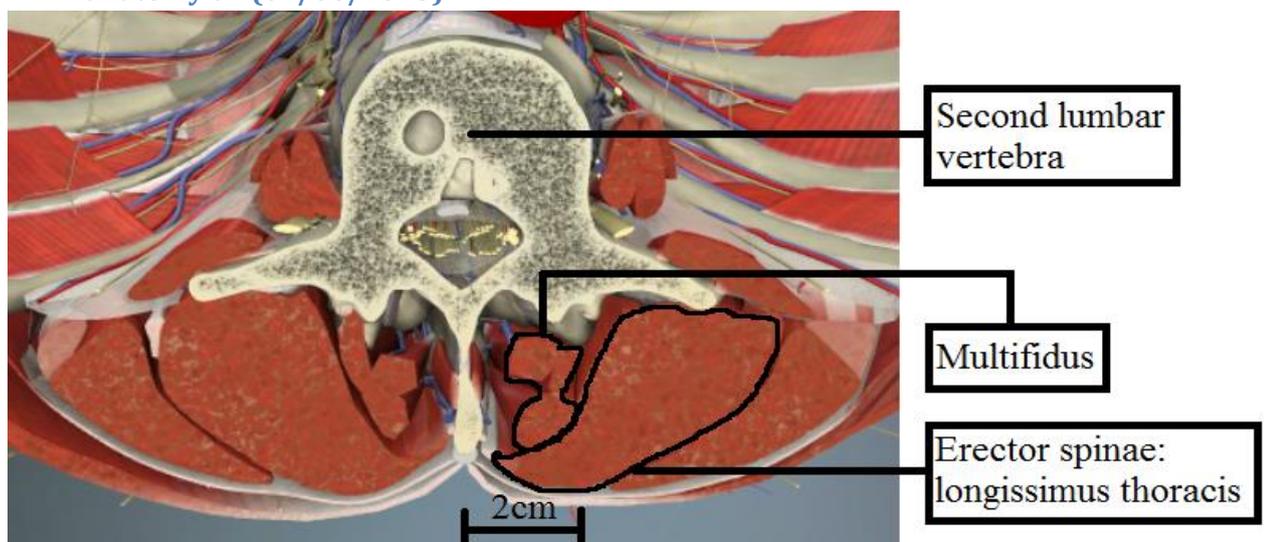
The following cross sections show the muscle layers that are typically found at each of the vertebral levels to be used as guidelines for electrode placement in the main study.

Figure 72: Cross section of the spine at the level of T9; image taken from www.anatomy.tv (04/06/2015)



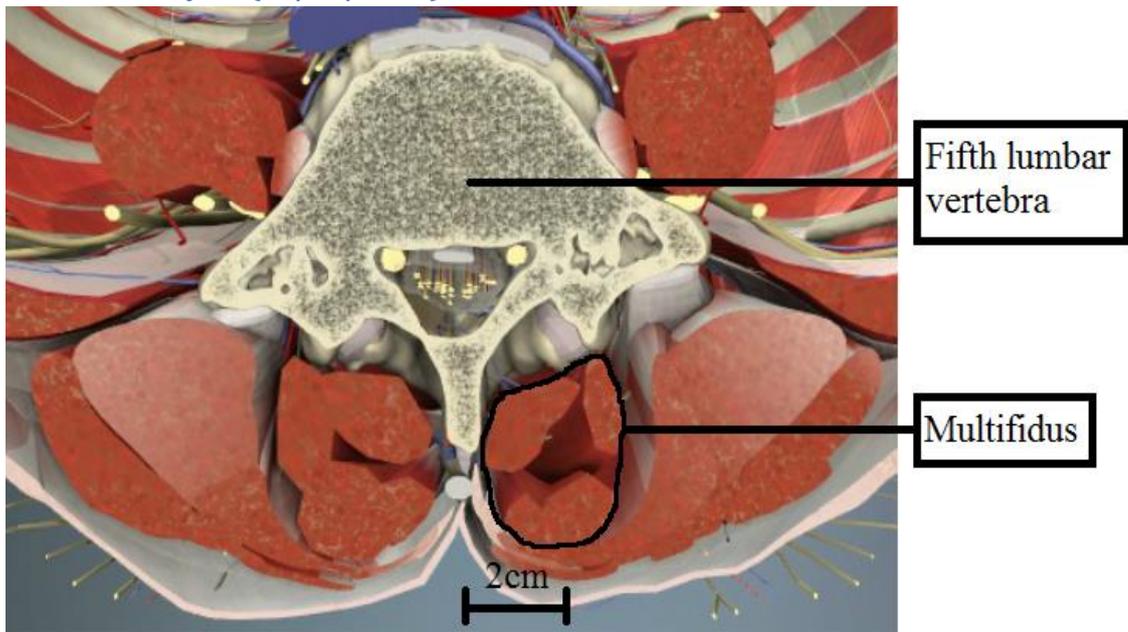
Note: The electrode placement used in the main study was 5cm lateral to the spinous process of T9. At this location sEMG signals were predominantly recorded from longissimus thoracis.

Figure 73: Cross section of the spine at the level of L2; image taken from www.anatomy.tv. (04/06/2015)



Note: The electrode placement used in the main study was 2cm lateral to the spinous process of L2. At this location sEMG signals were predominantly recorded from longissimus thoracis.

Figure 74: Cross section of the spine at the level of L5; image taken from www.anatomy.tv. (04/06/2015)



Note: The electrode placement used in the main study was 2cm lateral to the spinous process of L5. At this location sEMG signals were predominantly recorded from multifidus.

Appendix I

Tables of raw data for main study

Table 31: LMU sEMG amplitude changes throughout the flexion cycle

	Normalised EMG at L5 (epoch 1-epoch2)	Normalised EMG at L5 (epoch 2-epoch3)	Normalised EMG at L5 (epoch 3-epoch4)	Normalised EMG at L5 (epoch 4-epoch5)
Participant				
RS082	-11.67483828	-8.478589139	-4.197349043	0.039149158
RS083	-7.138046797	-1.540323357	0.214186976	0.320520935
RS084	-5.440492648	-2.682543672	-1.357295463	1.17255247
RS085	-5.740485158	-6.245760135	-1.298275983	9.523029778
RS086	-7.987617247	-4.699143588	-2.150298447	0.717384051
RS087	-7.127847821	-6.320504313	-3.102189781	-4.927007299
RS088	-7.240250642	-1.917378629	1.840468852	-0.860316344
RS089	-6.406495426	-1.704702947	0.985747944	0.349949182
RS091	-8.483887619	-7.632180518	-7.709608436	-5.729665954
RS092	-20.90073159	-21.07691791	-7.636418552	1.902812269
RS093	-7.934198948	-1.020059982	0.871001922	2.336440206
RS094	-11.65020996	-8.160254228	-3.109749177	3.024628306
RS095	-6.768691023	-4.426264215	-0.588253085	-0.281272683
RS096	-4.926907797	-5.444914704	-5.583179706	-6.60166697
RS097	-5.825448383	-2.895071318	-2.072447394	-2.354893377
RS098	-6.282675048	-5.718322202	5.416880723	13.78471488
RS099	-8.306332842	-6.889638463	-5.287662122	-5.458691624
RS100	-7.213062597	-0.41587601	5.631370232	9.258354449
Shapiro-Wilk Statistic	0.683	0.777	0.965	0.91
df	18	18	18	18
Sig.	0	0.001	0.707	0.085
mean	-8.169345546		-1.618503919	0.90089008
sd	3.667082139		3.801355762	5.471893874

Table 32: LES sEMG amplitude changes throughout the flexion cycle

	Normalised EMG at L2 (epoch 1- epoch 2)	Normalised EMG at L2 (epoch 2- epoch 3)	Normalised EMG at L2 (epoch 3- epoch 4)	Normalised EMG at L2 (epoch 4- epoch 5)
Participant				
RS082	-7.376342204	-4.160677008	2.351279391	9.363867048
RS083	-3.884732128	4.10457549	3.952184069	1.199145609
RS084	-0.107118695	-0.120273622	- 0.236788694	-0.210478839
RS085	-3.583946723	2.468162168	6.747283561	1.454609183
RS086	-3.695282404	0.965469595	3.408002729	2.68914765
RS087	-3.312377833	-2.205610439	4.543293359	0.612816314
RS088	-0.064352365	2.659897751	0.714004811	-0.177735103
RS089	-4.269183305	-0.668904297	4.154718398	1.89940204
RS091	-6.078212291	-4.931098696	-2.44320298	-2.003724395
RS092	-15.37917225	-2.673853149	7.93122657	3.789464175
RS093	-2.494378404	4.07796871	2.115032477	0.270194504
RS094	-1.878702582	2.16018067	2.186364678	2.169999673
RS095	-2.551885575	-1.263501309	1.783278652	2.507649206
RS096	-5.17243915	-2.559011692	- 1.517758659	0.383851754
RS097	-0.521930665	0.252186682	0.142054523	0.178765243
RS098	-1.036708046	6.089790049	6.121100024	0.584452858
RS099	-8.594453053	-4.891977293	- 0.397853755	-0.353196701
RS100	-0.497889656	-0.164278313	0.174387747	0.232516996
Shapiro-Wilk Statistic	0.842	0.97	0.974	0.789
df	18	18	18	18
Sig.	0.006	0.793	0.875	0.001
mean		-0.047830817	2.318255939	
sd		3.213548543	2.883091056	

Table 33: TES sEMG amplitude changes throughout the flexion cycle

	Normalised EMG at T9 (epoch 1-epoch 2)	Normalised EMG at T9 (epoch 2-epoch 3)	Normalised EMG at T9 (epoch 3-epoch 4)	Normalised EMG at T9 (epoch 4-epoch 5)
Participant				
RS082	1.136810636	1.093541027	0.314688065	-3.292423885
RS083	-0.338603087	0.186815496	0.154122784	-0.52308339
RS084	-1.38611413	-1.839041564	-1.050323101	-1.807805189
RS085	1.017690918	1.872134488	-3.695642106	-4.022136804
RS086	-1.317099034	2.625826675	-0.976662419	-1.601726367
RS087	0.418951703	3.415832501	0.076451041	-1.889869727
RS088	-0.047527938	0.053011931	0.120647842	-0.285167627
RS089	0.007318858	0.764820688	-1.599170529	-4.24493779
RS091	-2.679246869	2.187526267	2.843153736	-2.458603009
RS092	-1.034300284	-0.43296291	-0.168374465	-1.895415404
RS093	-0.98007382	-1.729542036	-1.938279868	-2.192741225
RS094	-0.082115086	-0.074650078	-0.011197512	0.011197512
RS095	-2.928949536	2.389181921	2.937472183	0.292610865
RS096	-2.867195741	-0.140174014	-0.006371546	-0.915378788
RS097	-0.997549877	-0.553664047	-0.715944888	-0.548891081
RS098	-0.500691542	2.746849429	0.067222475	-2.505775724
RS099	-0.698249365	2.702792997	-0.434317787	0.65147668
RS100	-0.681569167	-1.537155993	-0.771800546	-2.160719273
Shapiro-Wilk Statistic	0.941	0.935	0.893	0.964
df	18	18	18	18
Sig.	0.297	0.239	0.044	0.684
mean	-0.775472964	0.762841266		-1.632743901
sd	1.179855488	1.680524585		1.411183281

Table 34: Mean sEMG amplitudes over the entire cycle

Participant	Normalised Average EMG amplitude over entire cycle		
	T9	L2	L5
RS082	16.20250177	24.1521769	34.66657967
RS083	8.763981972	13.05819604	19.92090779
RS084	6.653347824	9.349394873	25.8809853
RS085	17.29379892	19.18740507	27.79714145
RS086	9.615939131	12.41205613	23.39376413
RS087	11.06399462	12.49669819	24.7235125
RS088	3.970410812	8.266214498	21.36124343
RS089	18.45084167	14.48696471	21.56172041
RS091	27.24636463	38.77839851	42.59088563
RS092	9.0922211	29.37173484	46.52325659
RS093	7.180581418	14.71396959	24.80110626
RS094	7.774805599	14.71541256	35.64294632
RS095	13.37771423	15.22145833	21.60204452
RS096	15.26834828	15.55261416	23.7270532
RS097	5.756196901	8.432292664	16.93969778
RS098	13.7272931	16.70213198	35.91006618
RS099	18.25805158	29.26863925	39.52776854
RS100	6.980041678	12.16417722	21.65623189
Shapiro-Wilk Statistic			
df			
Sig.			
mean	12.03757973	17.12944086	28.23482842
sd	5.932283588	8.203623936	8.652125932

Table 35: Mean sEMG amplitudes during the flexion phase of the cycle

Participant	Normalised Av EMG Amp during Flexion phase		
	T9	L2	L5
RS082	11.0494847	17.49209365	30.09451725
RS083	3.376690096	7.667037236	19.61405836
RS084	5.563979254	3.061339547	23.31154958
RS085	11.67044876	8.166841921	21.59348757
RS086	6.030192821	7.729003878	21.27497868
RS087	6.709343337	6.262877067	21.75956647
RS088	2.57930463	3.54397667	17.57835556
RS089	14.52061478	8.314802339	18.174836
RS091	21.1565941	30.79702048	35.95973748
RS092	6.350122673	15.74050376	39.50600711
RS093	6.242255164	6.821069207	18.50295422
RS094	6.07651633	8.892743757	32.17568948
RS095	8.491396862	13.06631769	18.9118074
RS096	11.58347079	9.731597912	17.14875497
RS097	5.269354377	2.09570325	13.06312668
RS098	11.39305059	9.438217883	27.19772748
RS099	9.324468796	24.24269079	34.7275405
RS100	4.303714525	4.134758764	13.83299056
Shapiro-Wilk Statistic			
df			
Sig.			
mean	8.427277921	10.39992199	23.57931585
sd	4.565336485	7.515351467	7.854274933

Table 36: Mean sEMG amplitudes during the return phase of the cycle

Participant	Normalised Av EMG during return phase		
	T9	L2	L5
RS082	21.35158524	30.80101001	39.28338398
RS083	14.14660346	18.4456075	20.21712382
RS084	7.768095948	15.80658498	28.51263039
RS085	22.92756912	29.45145461	34.03588388
RS086	13.17099034	17.09510837	25.516257
RS087	15.40335569	18.71467061	27.68745853
RS088	5.350549009	12.95934054	25.11730232
RS089	22.38106855	20.66270411	24.94513998
RS091	33.32983105	46.74487896	49.22203377
RS092	11.82469813	43.01651586	53.54553997
RS093	8.110955754	22.60329124	31.09207478
RS094	9.473094868	20.53153536	39.06480536
RS095	18.2469863	17.3724518	24.28774498
RS096	18.95747347	21.37215972	30.30729883
RS097	6.243039425	14.76568984	20.82333004
RS098	16.06153561	23.96430663	44.61787193
RS099	27.19497528	34.29661757	44.3222956
RS100	9.653146282	20.19359567	29.48629086
Shapiro-Wilk Statistic			
df			
Sig.			
mean	15.64419742	23.82208463	32.89358144
sd	7.689721144	9.491738737	9.904758693

Table 37: sEMG ratio data for LMU/TES, LMU/LES and LES/TES

Participant	Ratio over entire cycle			Ratio during Flexion phase			Ratio during Return phase		
	L5/T9	L5/L2	L2/T9	L5/T9F	L5/L2F	L2/T9F	L5/T9R	L5/L2R	L2/T9R
RS082	2.139582	1.43534	1.490645	2.723613	1.720464	1.583069	1.839835	1.275393	1.442563
RS083	2.273043	1.525548	1.489984	5.808664	2.558232	2.270578	1.429115	1.09604	1.303889
RS084	3.889919	2.768199	1.405217	4.189726	7.61482	0.550207	3.670479	1.803845	2.034808
RS085	1.607347	1.448718	1.109496	1.850271	2.644044	0.699788	1.484496	1.155661	1.284543
RS086	2.432811	1.884761	1.290779	3.528076	2.752616	1.281718	1.937307	1.492606	1.297936
RS087	2.234592	1.978404	1.129492	3.243174	3.474372	0.933456	1.797495	1.479452	1.214974
RS088	5.380109	2.584163	2.081954	6.815153	4.960065	1.374005	4.694341	1.938162	2.422058
RS089	1.168604	1.488353	0.785166	1.251657	2.185841	0.572621	1.114564	1.207254	0.923222
RS091	1.563177	1.098315	1.42325	1.699694	1.167637	1.45567	1.476816	1.052993	1.402494
RS092	5.11682	1.583946	3.230425	6.221298	2.509831	2.478772	4.52828	1.244767	3.637853
RS093	3.453913	1.685548	2.049133	2.964146	2.712618	1.092725	3.833343	1.375555	2.786761
RS094	4.584416	2.422151	1.892705	5.295088	3.618196	1.463461	4.123764	1.902673	2.167352
RS095	1.614778	1.419184	1.137822	2.227173	1.447371	1.538771	1.331055	1.398061	0.952072
RS096	1.554003	1.525599	1.018618	1.48045	1.762173	0.840128	1.598699	1.418074	1.127374
RS097	2.942863	2.008908	1.464907	2.479075	6.23329	0.397715	3.335447	1.410251	2.365144
RS098	2.615961	2.150029	1.21671	2.387221	2.881659	0.828419	2.777933	1.861847	1.492031
RS099	2.16495	1.350516	1.603054	3.724345	1.432495	2.5999	1.629797	1.292323	1.261138
RS100	3.102593	1.780329	1.742708	3.214198	3.345538	0.960742	3.054578	1.46018	2.091919
mean	2.76886	1.785445	1.531226	3.394612	3.056737	1.27343	2.536519	1.436952	1.733785
sd	1.262471	0.456615	0.550121	1.668205	1.69842	0.650538	1.215068	0.273811	0.72645

Table 38: IV-RoMmax data

Participant	Angular range			
	L2/L3	L3/L4	L4/L5	L5/S1
RS082	14.8	11.8	2.8	1
RS083	11.9	11.6	4.4	6.4
RS084	8	11.7	11.6	6.1
RS085	8.4	11.2	15.3	7.9
RS086	11.3	10.7	8.7	2
RS087	9.1	9.9	13	5.2
RS088	10	11	11.2	5.4
RS089	9.9	11.9	8.8	7.6
RS091	7.4	10.6	7.4	4.6
RS092	4.6	6.4	9.7	9.5
RS093	7.7	10.3	9.7	9.2
RS094	10.4	13.4	6.2	11.4
RS095	4.3	9.3	16.6	0.8
RS096	12	10.19	9.1	5
RS097	5.1	7.7	11	10.5
RS098	7.7	9.7	13.3	9.8
RS099	9.8	8.9	4.1	4.3
RS100	8.3	9.3	14	8.7
Shapiro-Wilk Statistic	0.969	0.968	0.978	0.958
df	18	18	18	18
Sig.	0.78	0.767	0.931	0.556
mean	8.927777778	10.31055556	9.827777778	6.411111111
sd	2.712889832	1.646388105	3.875534962	3.179139523

Table 39: Initial attainment rate data

	laxity data for 18			
Participant	L2/L3	L3/L4	L4/L5	L5/S1
RS082	0.2168	0.0076	0.0425	0.0202
RS083	0.2954	0.2883	0.008	0.1006
RS084	0.3595	0.1779	0.0526	0.0505
RS085	0.1765	0.1304	0.0748	0.0697
RS086	0.1916	0.0429	0.0179	0.1373
RS087	0.2069	0.074	0.132	0.0211
RS088	0.1074	0.0476	0.0444	0.2587
RS089	0.2149	0.0873	0.0163	0.1441
RS091	0.1155	0.0875	0.0797	0.1161
RS092	0.1926	0.2004	0.1565	0.0843
RS093	0.3544	0.1143	0.039	0.0727
RS094	0.1238	0.1768	0.026	0.094
RS095	0.07	0.0344	0.1507	0.0077
RS096	0.2658	0.1519	0.1193	0.1015
RS097	0.1945	0.2239	0.0791	0.13
RS098	0.1675	0.1118	0.0699	0.641
RS099	0.1087	0.0428	0.0404	0.0453
RS100	0.237	0.4892	0.1776	0.3293
Shapiro-Wilk Statistic	0.949	0.847	0.909	0.698
df	18	18	18	18
Sig.	0.417	0.008	0.084	0
mean	0.199933333		0.07370555 6	
sd	0.081584456		0.05242889 3	

Table 40: Lordotic angle data

	lordosis data	
Participant	Absolute angle between L2 and S1	Absolute angle between L5 and S1
RS082	56.5	9.1
RS083	53.6	9.1
RS084	51.3	16.8
RS085	52.1	17.1
RS086	46	12.3
RS087	60.8	18.7
RS088	66.6	21.5
RS089	61.4	17.6
RS091	50.3	19.4
RS092	50.2	23.2
RS093	52.9	21.1
RS094	58	23.8
RS095	33.9	21
RS096	54	17.8
RS097	41.8	25.1
RS098	44.1	14.7
RS099	63.9	21.2
RS100	37.6	8.5
Shapiro-Wilk Statistic	0.977	0.924
df	18	18
Sig.	0.908	0.155
mean	51.94444444	17.66666667
sd	8.839032568	5.128581962

Table 41: Normality test data (Shapiro-Wilk test)

Tests of Normality						
	Kolmogorov-Smirnova			Shapiro-Wilk		
	Statistic	df	Sig.	Statistic	df	Sig.
L5epoch1min2	0.299	18	0	0.683	18	0
L5epoch2min3	0.2	18	0.056	0.777	18	0.001
L5epoch3min4	0.08	18	.200*	0.965	18	0.707
L5epoch4min5	0.182	18	0.117	0.91	18	0.085
L2epoch1min2	0.185	18	0.105	0.842	18	0.006
L2epoch2min3	0.087	18	.200*	0.97	18	0.793
L2epoch3min4	0.107	18	.200*	0.974	18	0.875
L2epoch4min5	0.183	18	0.116	0.789	18	0.001
T9epoch1min2	0.136	18	.200*	0.941	18	0.297
T9epoch2min3	0.135	18	.200*	0.935	18	0.239
T9epoch3min4	0.239	18	0.008	0.893	18	0.044
T9epoch4min5	0.112	18	.200*	0.964	18	0.684
L2L3angrange	0.12	18	.200*	0.969	18	0.78
L3L4angrange	0.112	18	.200*	0.968	18	0.767
L4L5angrange	0.108	18	.200*	0.978	18	0.931
L5S1angrange	0.098	18	.200*	0.958	18	0.556
BMI	0.142	18	.200*	0.951	18	0.437
L2L3laxity	0.14	18	.200*	0.949	18	0.417
L3L4laxity	0.143	18	.200*	0.847	18	0.008
L4L5laxity	0.177	18	0.142	0.909	18	0.084
L5S1laxity	0.308	18	0	0.698	18	0
LordosisL2S1	0.144	18	.200*	0.977	18	0.908
L5S1angle	0.155	18	.200*	0.924	18	0.155
* This is a lower bound of the true significance.						
a Lilliefors Significance Correction						

Appendix J: Scatter plots for sEMG variables vs IV-RoMmax

Figure 75: LES epoch 1-2 (%MVC) vs L4-L5 IV-RoMmax (°)

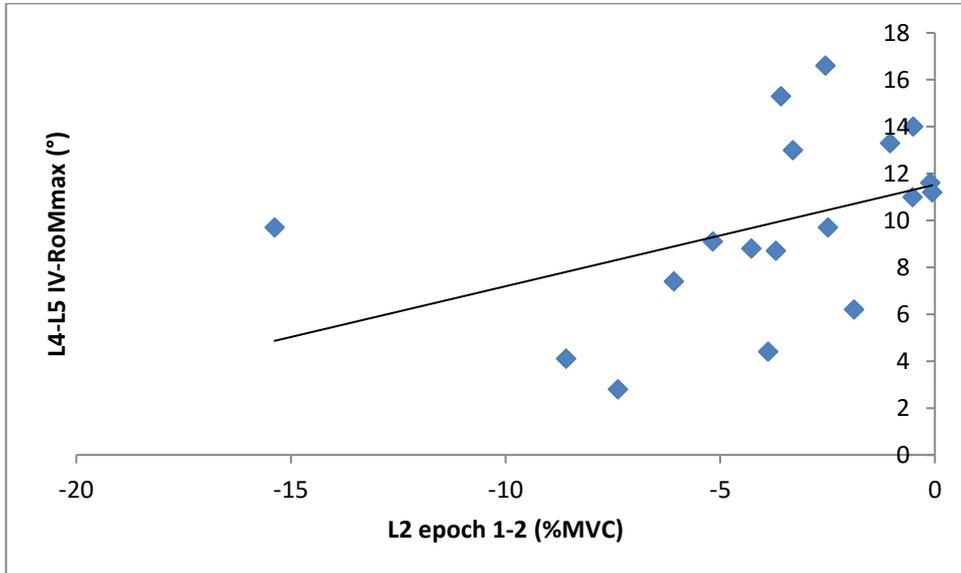


Figure 76: LMU epoch 4-5 (%MVC) vs L5-S1 IV-RoMmax (°)

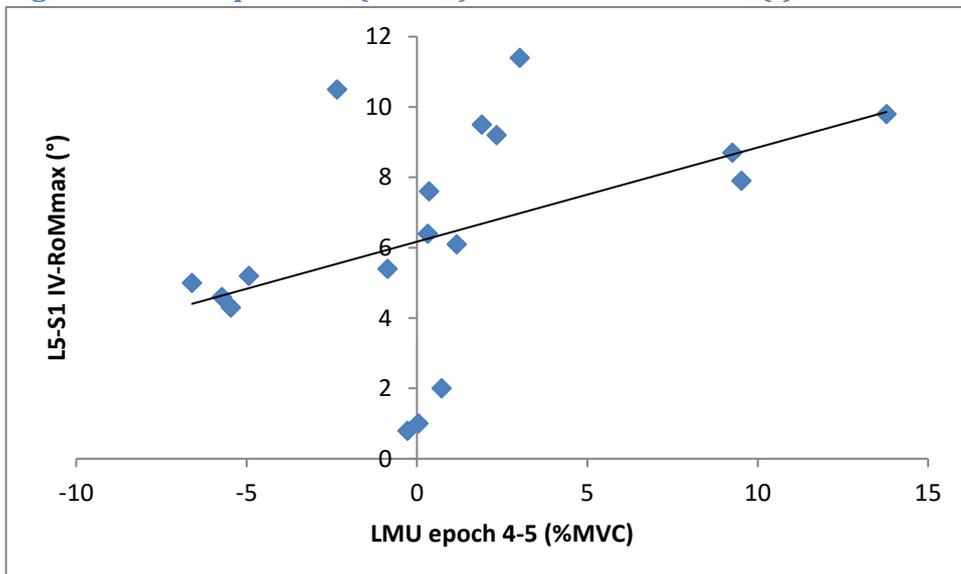


Figure 77: LES epoch 2-3 (%MVC) vs L5-S1 IV-RoMmax (°)

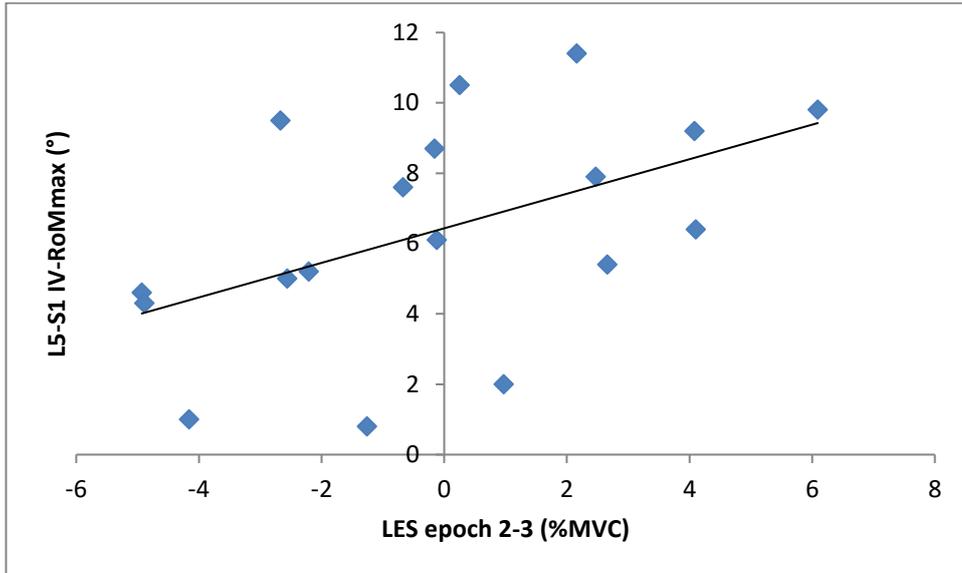


Figure 78: TES epoch 2-3 (%MVC) vs L5-S1 IV-RoMmax (°)

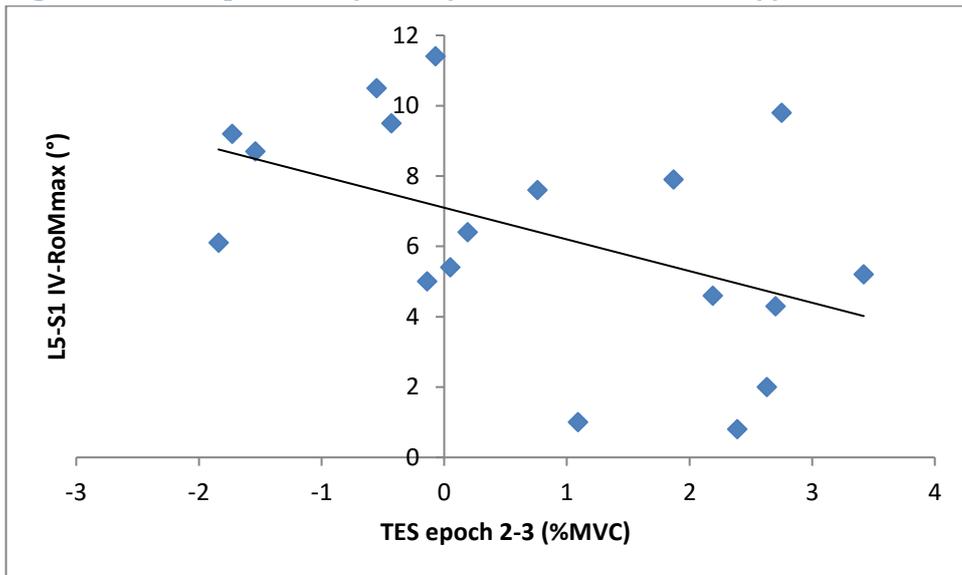


Figure 79: LES/TES ratio during flexion phase vs L4-L5 IV-RoMmax (°)

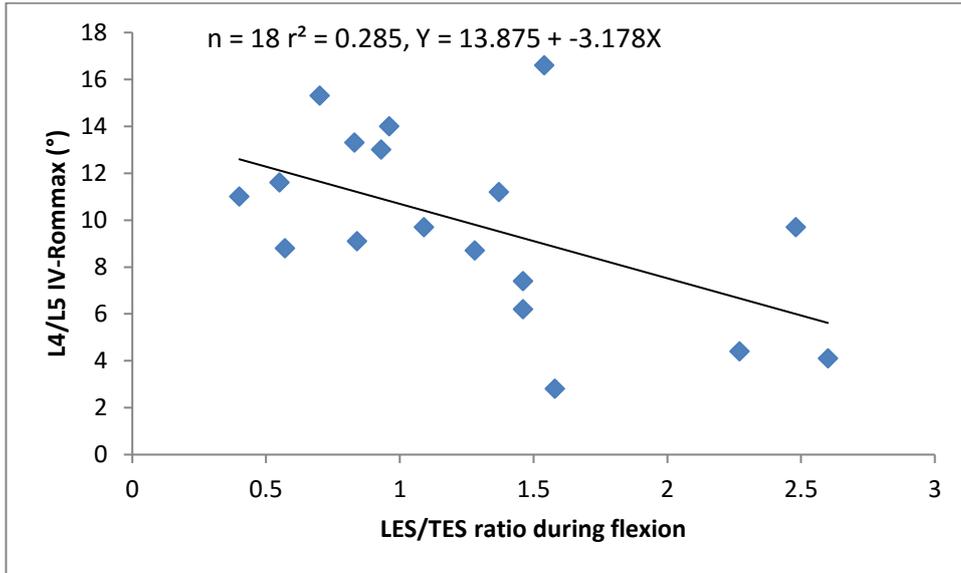
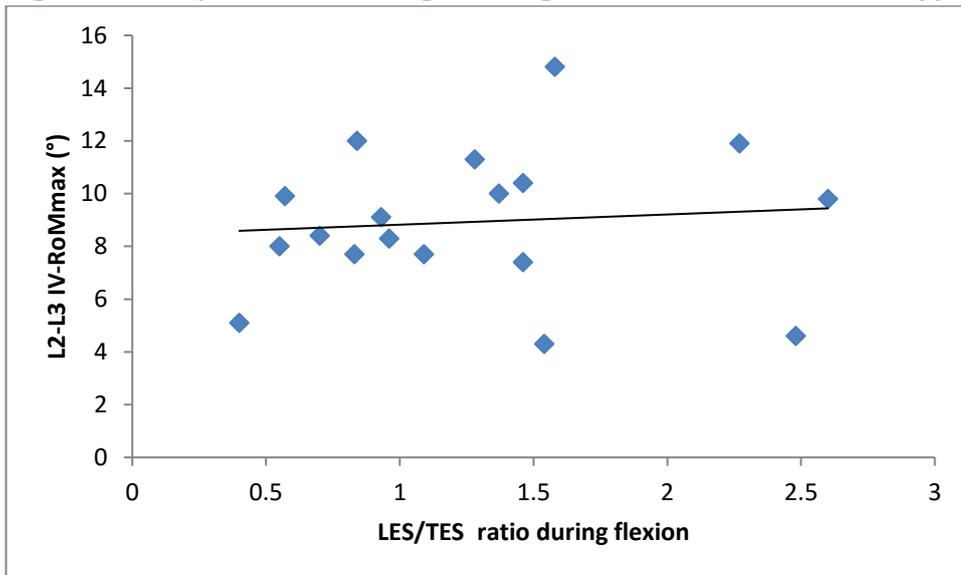


Figure 80: LES/TES ratio during flexion phase vs L2-L3 IV-RoMmax (°)



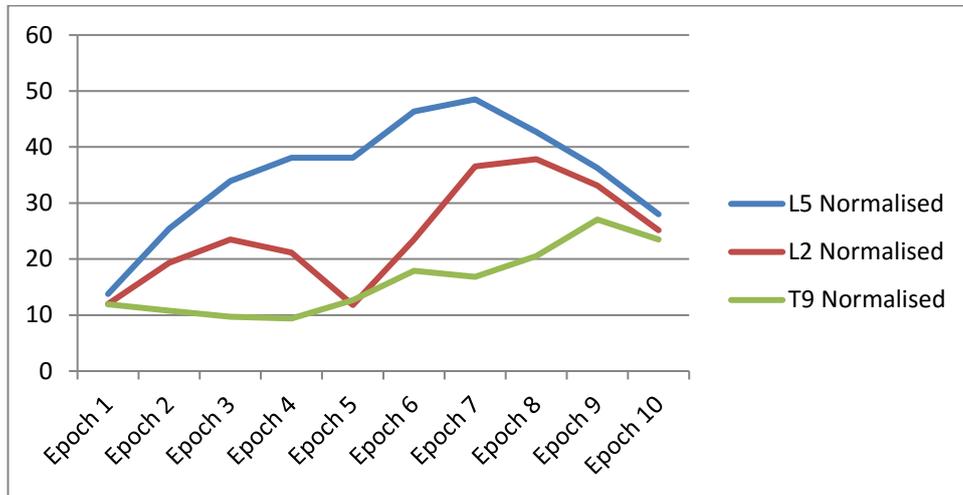
Appendix K: Preliminary study raw data

Table 42: Raw data for Preliminary Study 1

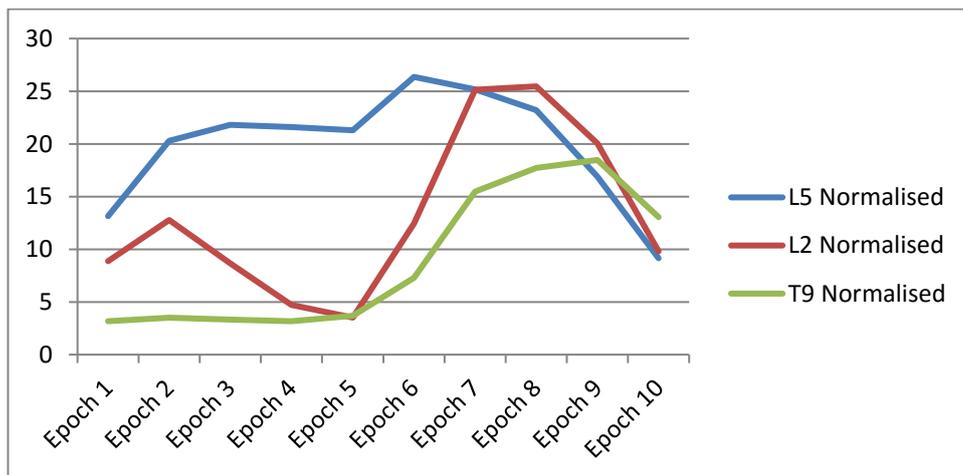
IV-RoMmax									
Recumbent	L2/L3	L3/L4	L4/L5	L5/S1	Weight-bearing	L2/L3	L3/L4	L4/L5	L5/S1
	4.729	7.364	9.329	7.989		3.973	10.18	13.31	11
	3.634	6.56	6.695	5.068		12.69	12.93	13.11	3.67
	3.936	6.754	5.451	6.338		7.343	9.932	9.283	7.097
	1.441	4.646	4.858	8.509		7.577	10.53	12.56	5.11
	5.74	8.522	4.566	1.813		14.49	8.486	1.952	2.158
	3.025	5.068	6.086	5.126		0	7.046	10.16	7.941
	4.501	5.397	6.23	4.7		11.45	13.578	14.394	4.293
	3.208	7.442	4.86	2.913		7.038	8.042	12.99	6.864
	3.894	5.131	6.8671	7.1724		15.953	12.957	3.5645	1.8339
	4.2	5.112	8.796	5.7		9.776	9.713	11.43	6.188
Motionshare									
Recumbent	L2/L3	L3/L4	L4/L5	L5/S1	Weight-bearing	L2/L3	L3/L4	L4/L5	L5/S1
	16.0790	25.0382	31.7194	27.1633		10.3294	26.466	34.6046	28.5989
	16.5505	29.8765	30.4914	23.0814		29.9292	30.4952	30.9198	8.65566
	17.5096	30.0458	24.2493	28.1952		21.8184	29.5112	27.5828	21.0875
	7.40721	23.8819	24.9717	43.7390		21.1784	29.4323	35.1063	14.2829
	27.8087	41.2867	22.1210	8.78348		53.4962	31.3298	7.20667	7.96721
	15.6695	26.2522	31.5255	26.5527		26.1923	28.0192	40.4024	31.5783
	21.6103	25.9122	29.9116	22.5657		0	31.0602	32.9269	9.82042
	17.4130	40.3951	26.3800	15.8117		20.1465	23.0205	37.1844	19.6484
	16.8830	22.2463	29.7734	31.0971		46.4988	37.7662	10.3895	5.34533
	17.6411	21.4717	36.9455	23.9415		26.3454	26.1756	30.8028	16.6761
Initial attainment rate									
Recumbent	L2/L3	L3/L4	L4/L5	L5/S1	Weight-bearing	L2/L3	L3/L4	L4/L5	L5/S1
	0.1407	0.1762	0.2109	0.1595		0.225	0.2581	0.3544	0.2268
	0.1917	0.0822	0.2247	0.3636		0.4578	0.5452	0.7297	0.4529
	0.0788	0.0929	0.2064	0.2084		0.1673	0.322	0.261	0.0743
	0.2254	0.2168	0.2305	0.2148		0.4395	0.005	0.0766	0.0337
	0.1919	0.2594	0.3138	0.3174		0.1725	0.1143	0.2456	0.0384
	0.1291	0.1728	0.2114	0.1832		0.3164	0.2548	0.0394	0.2318
	0.126	0.178	0.1583	0.1343		0.3405	0.4961	0.2367	0.3244
	0.0846	0.2049	0.2921	0.1958		1.0872	0.2975	0.1184	0.1498
	0.1484	0.1683	0.2157	0.2173		0.1756	1.0681	0.1952	0.0211

Appendix L: sEMG graphs for each participant during the entire flexion and return cycle

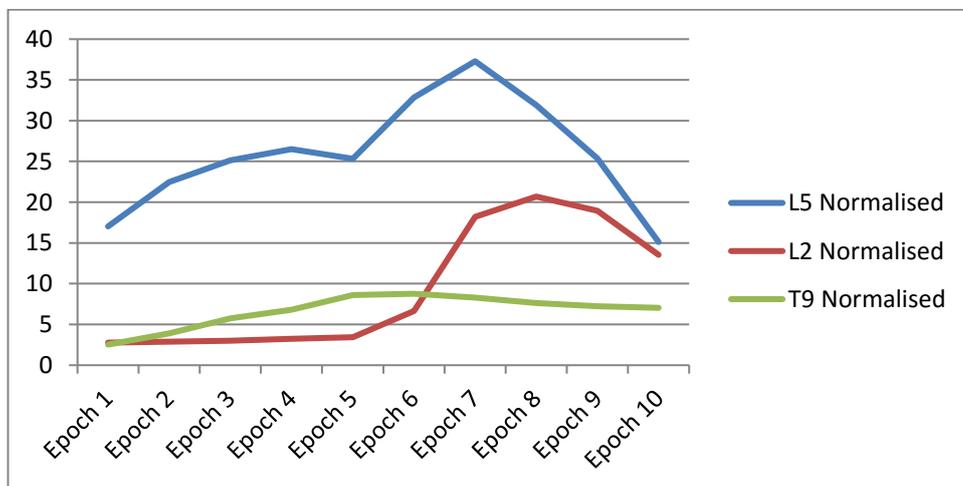
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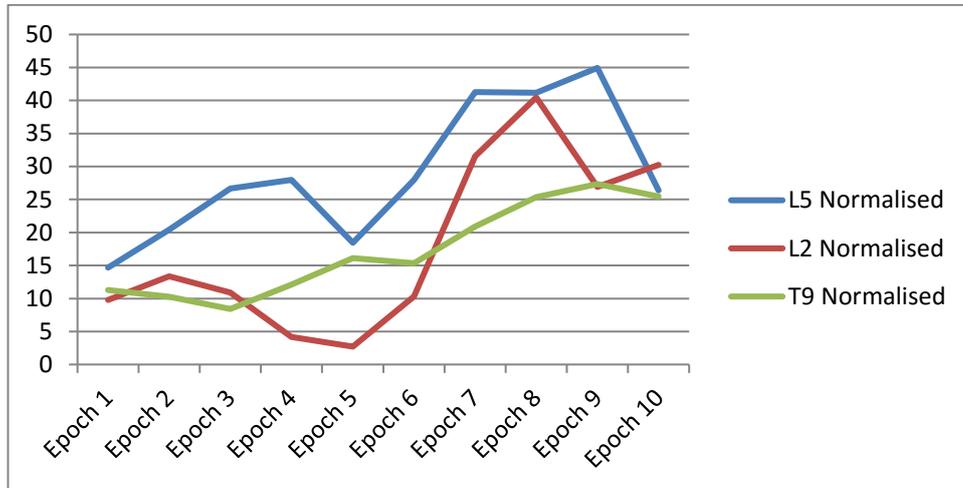
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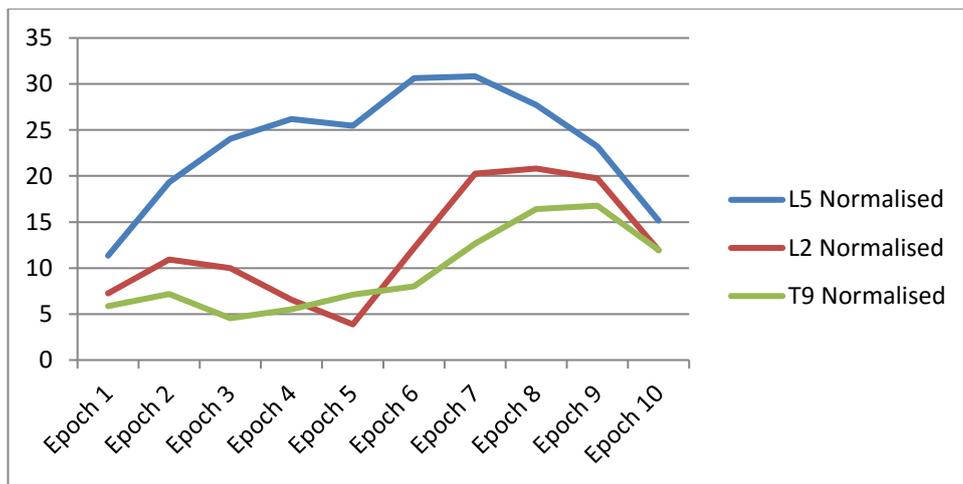
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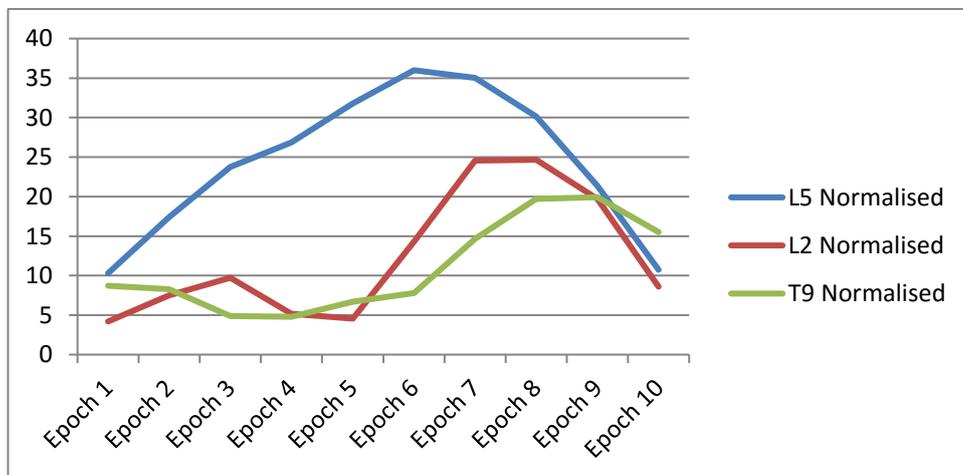
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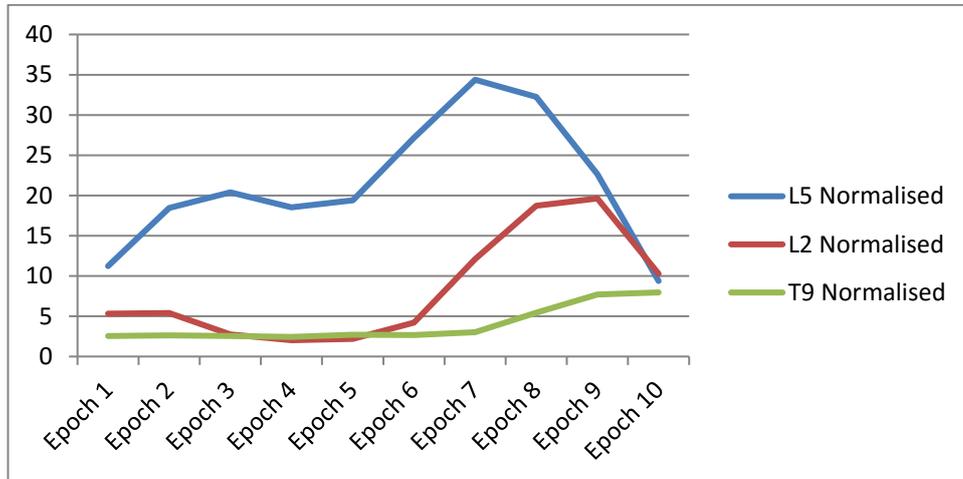
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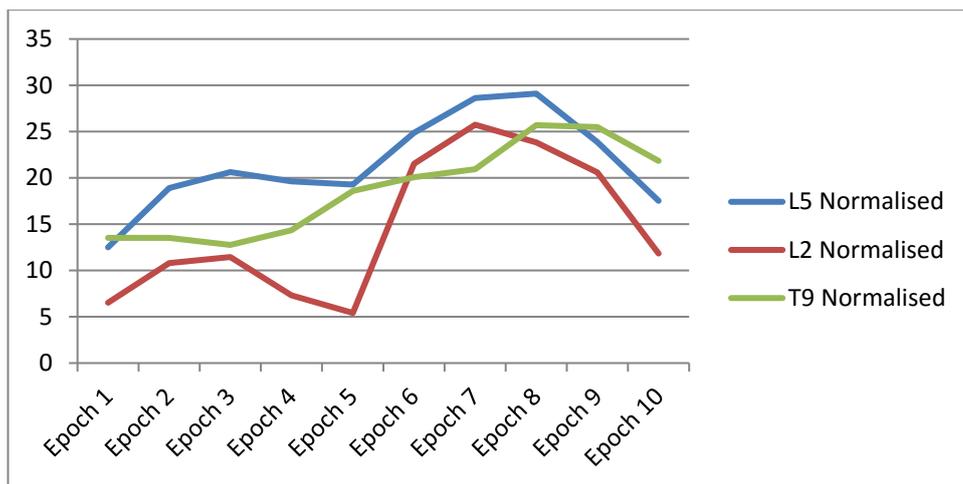
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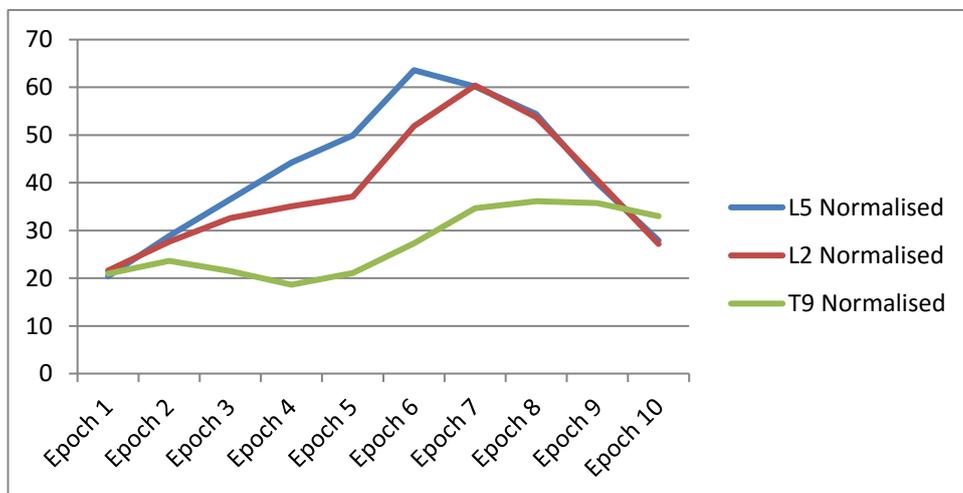
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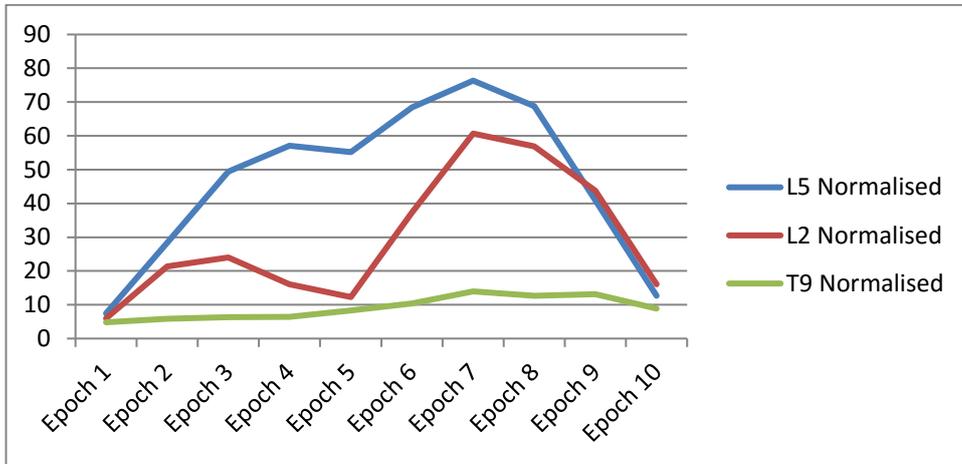
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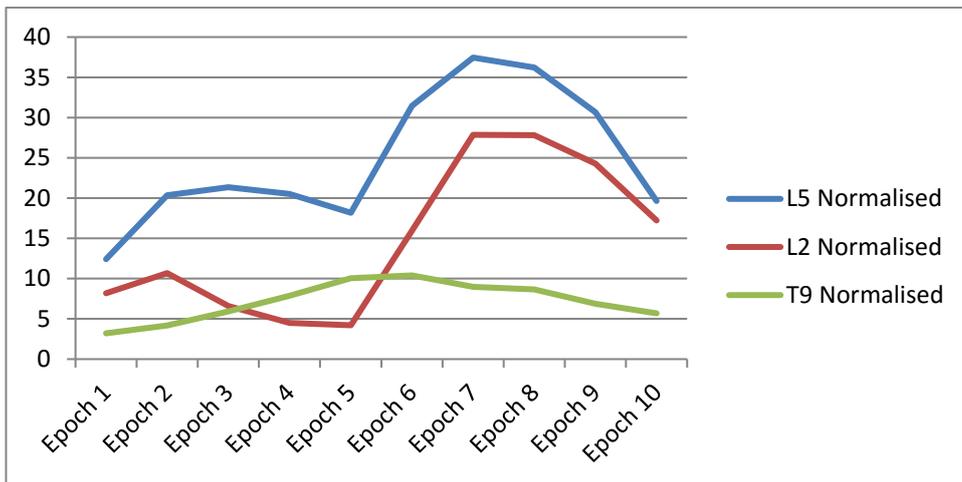
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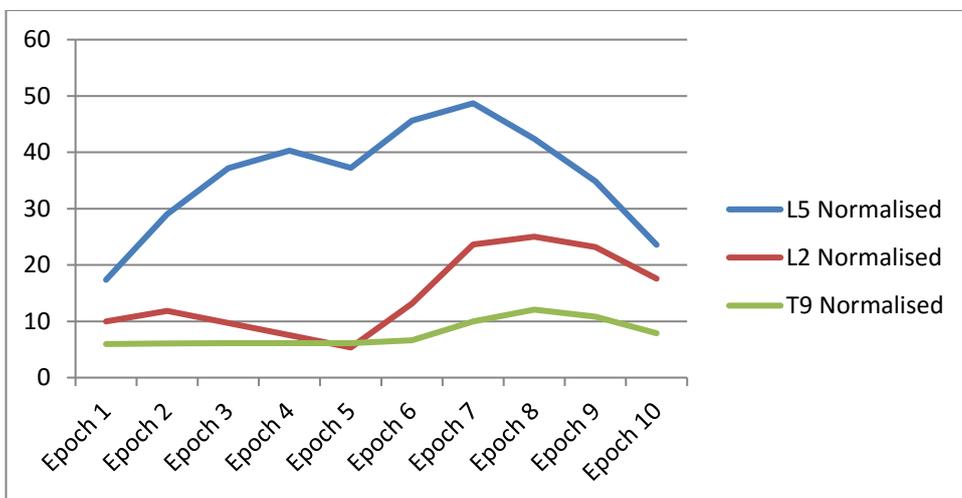
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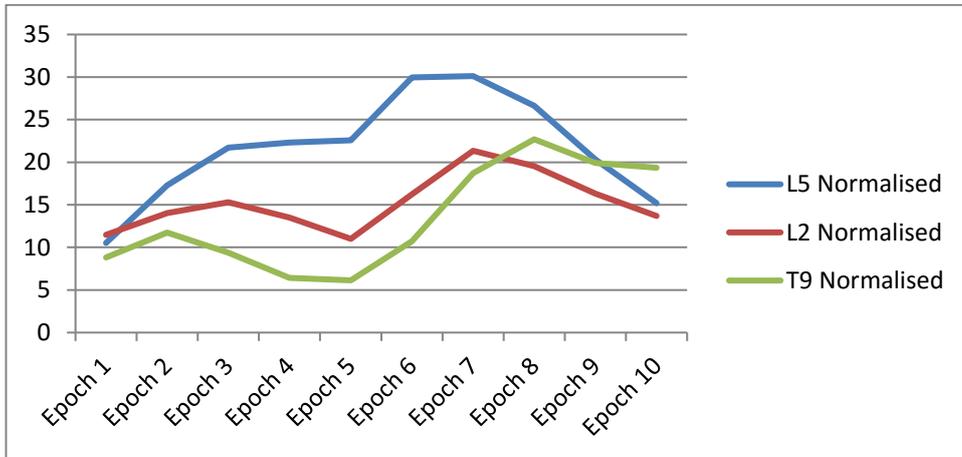
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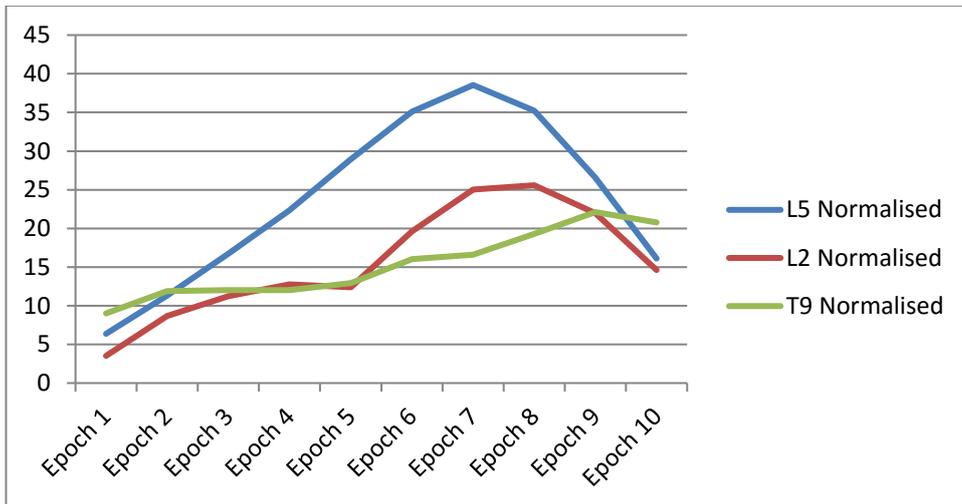
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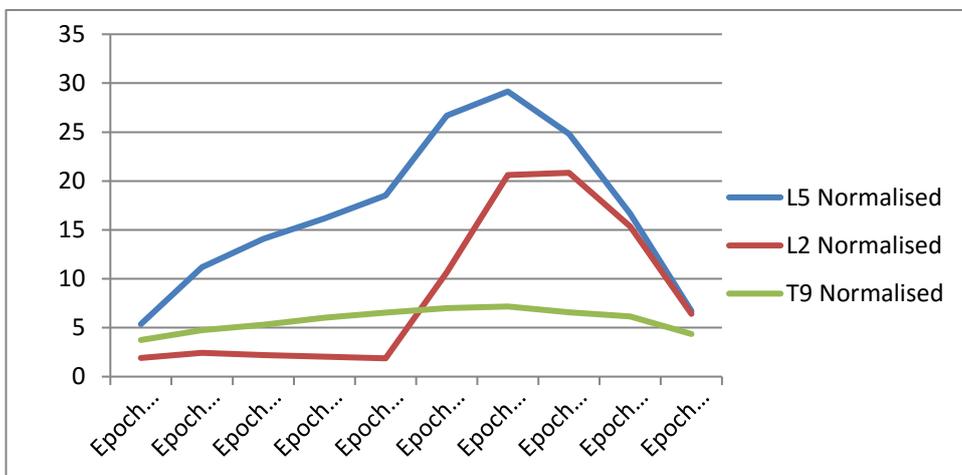
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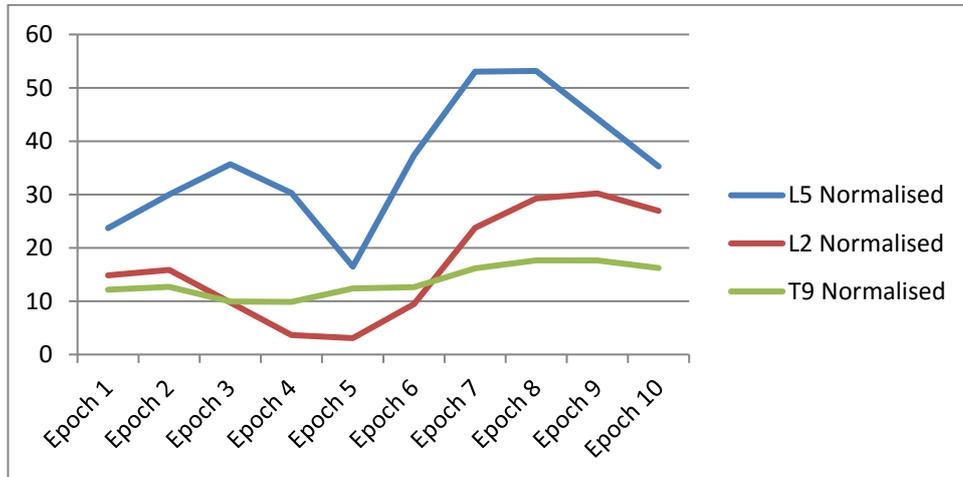
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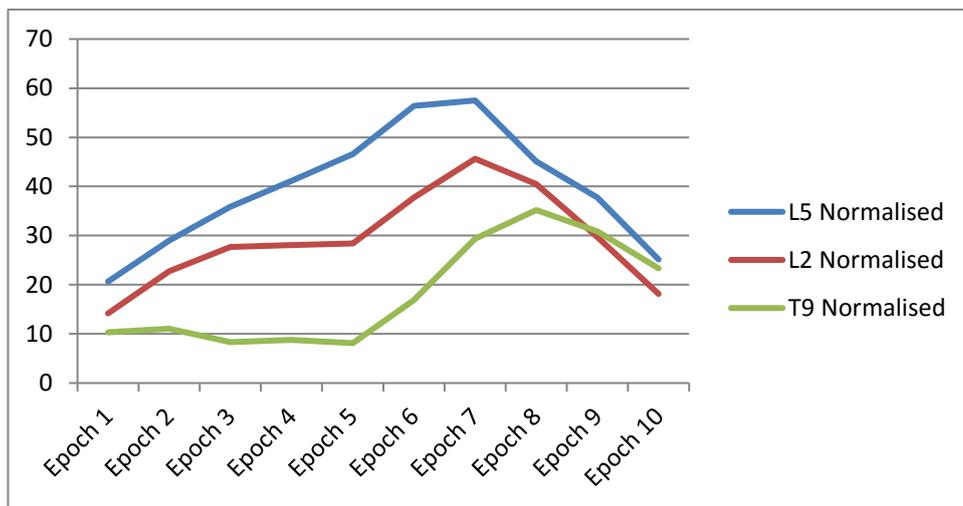
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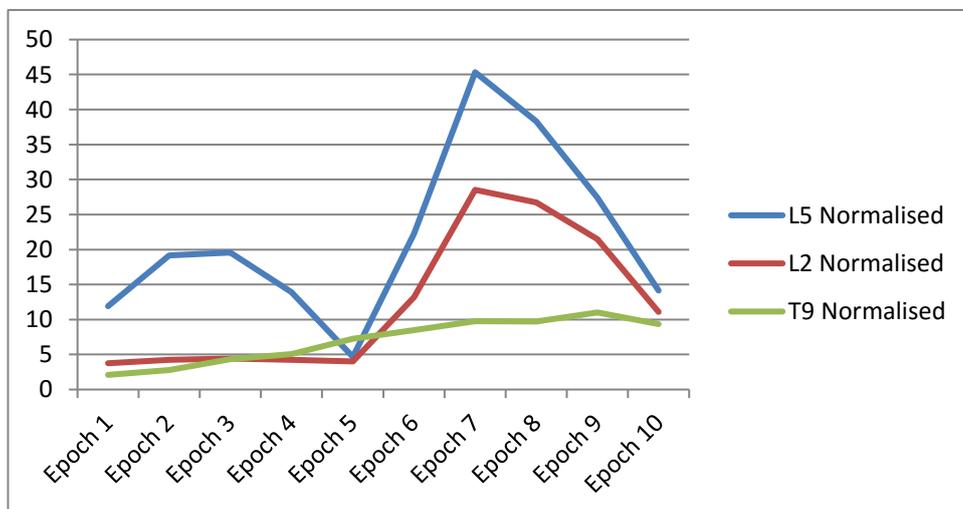
RS098



RS099

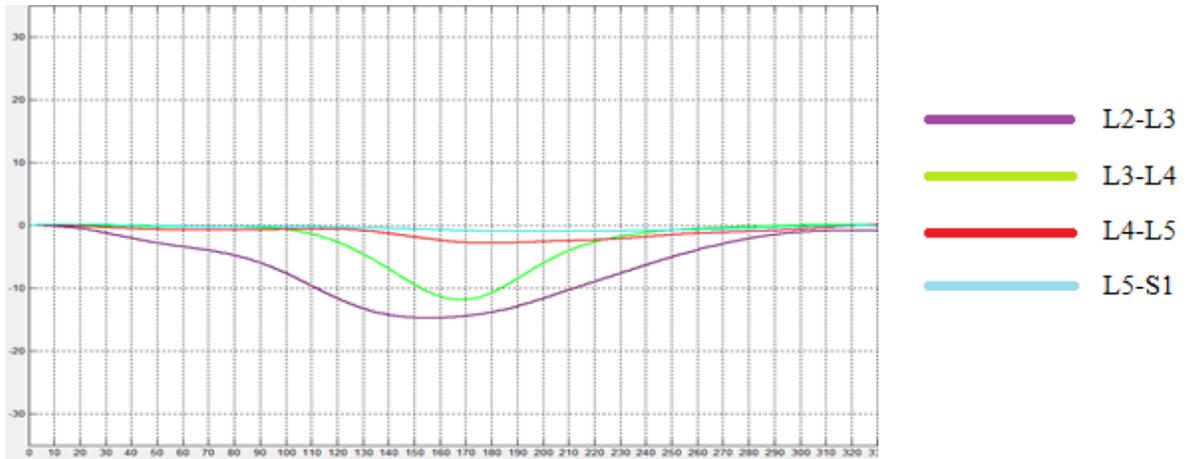


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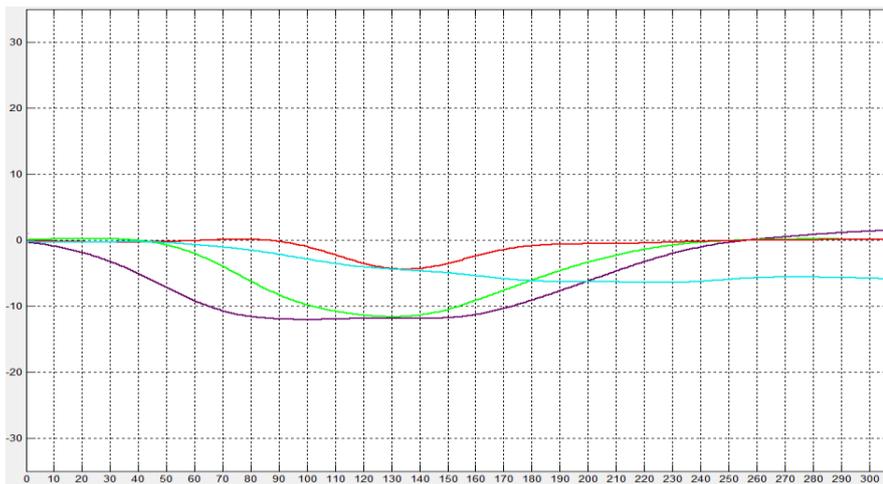


Appendix M: IV-RoM graphs for each participant during the entire flexion and return cycle

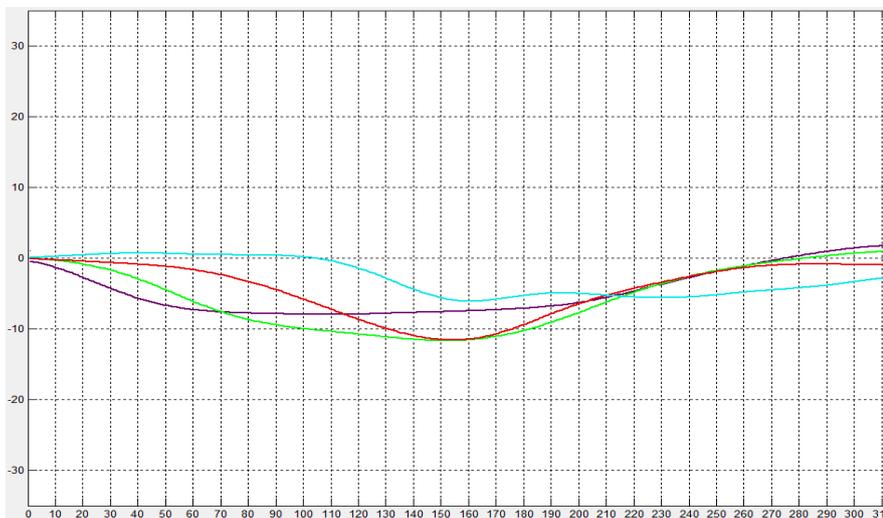
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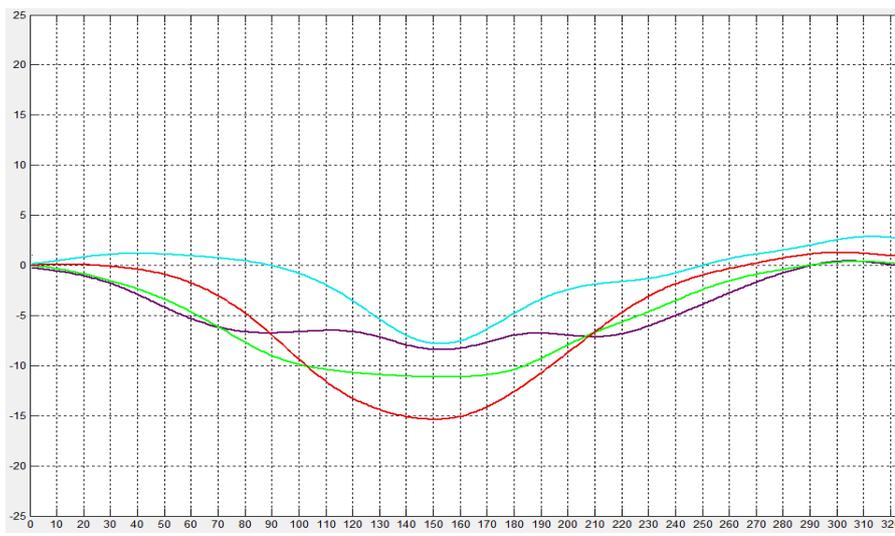
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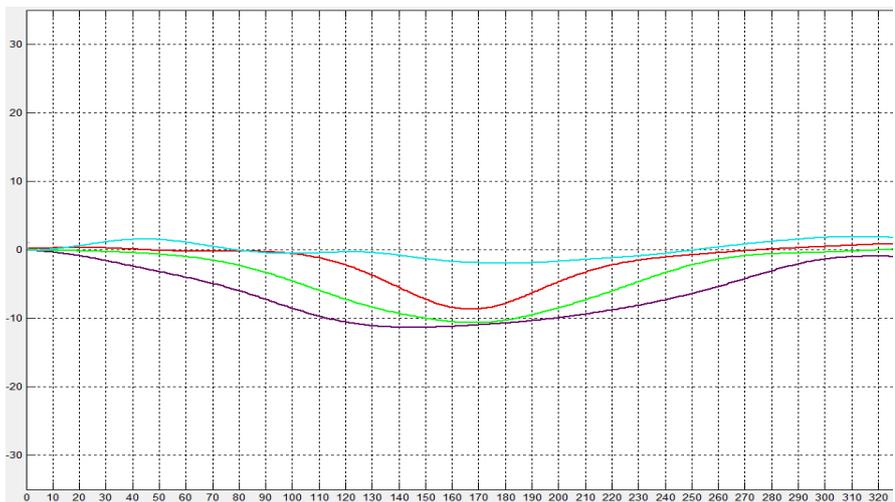
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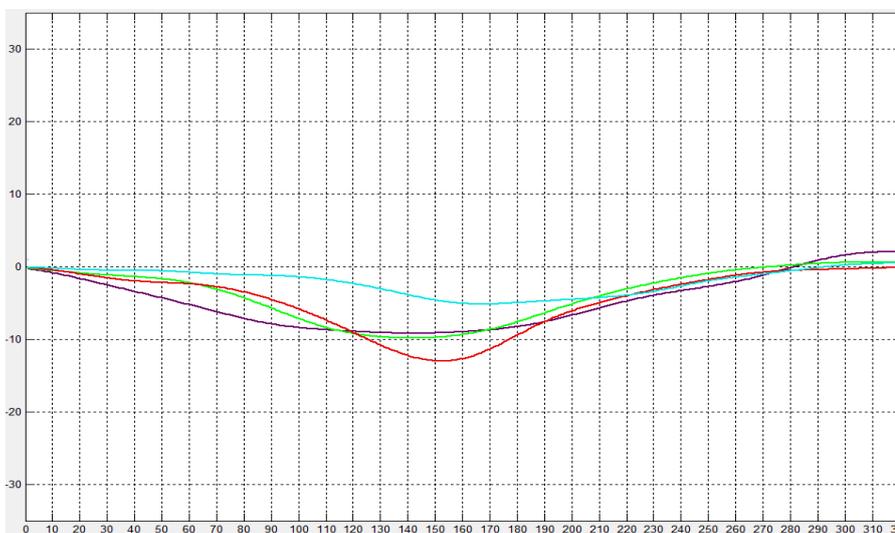
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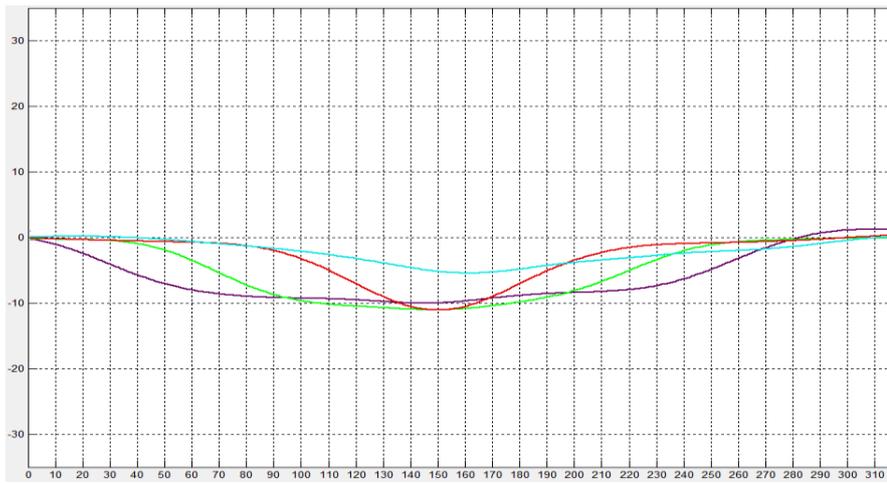
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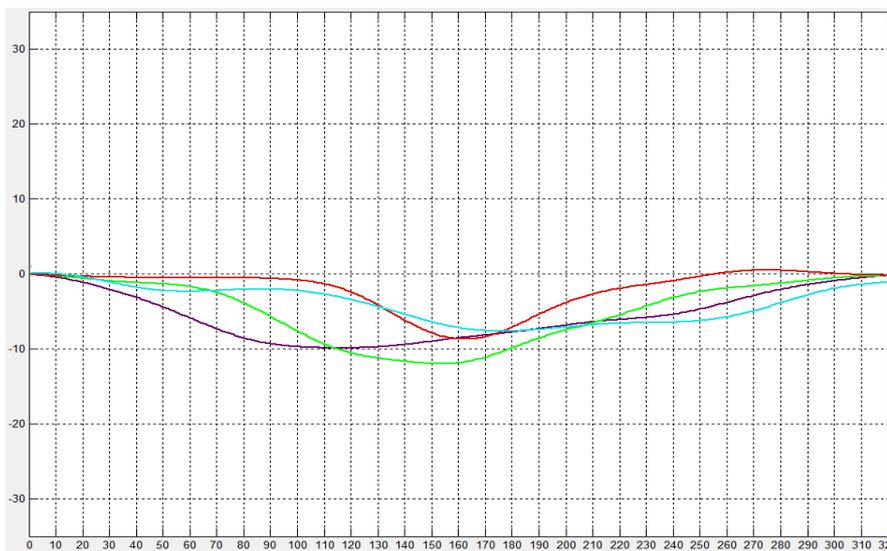
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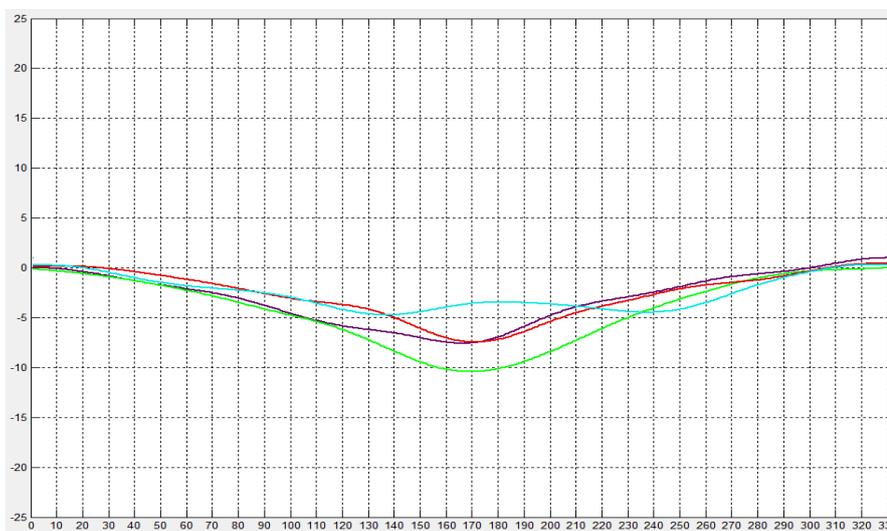
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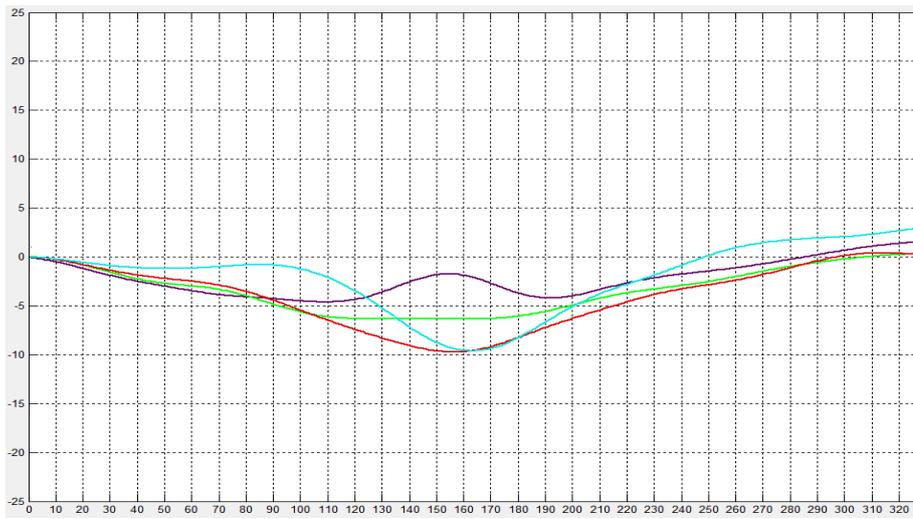
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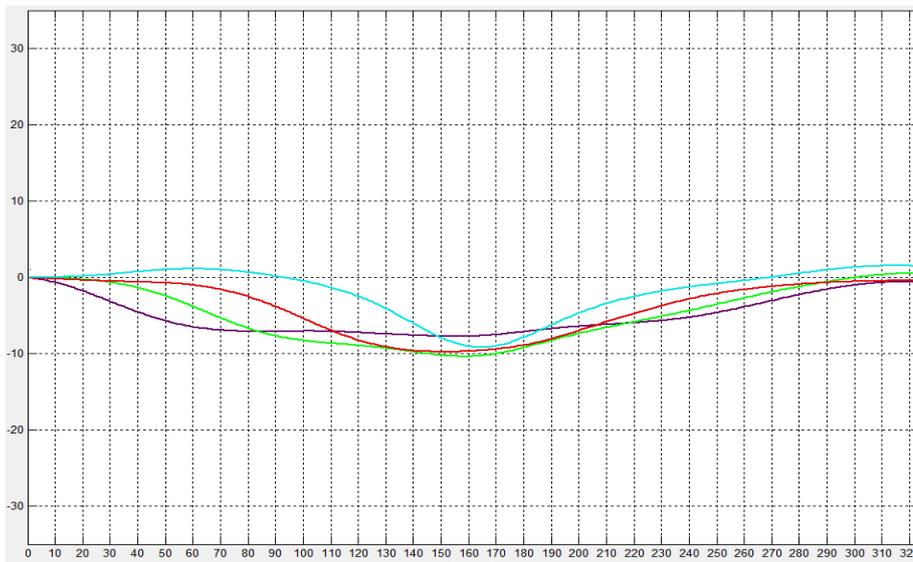
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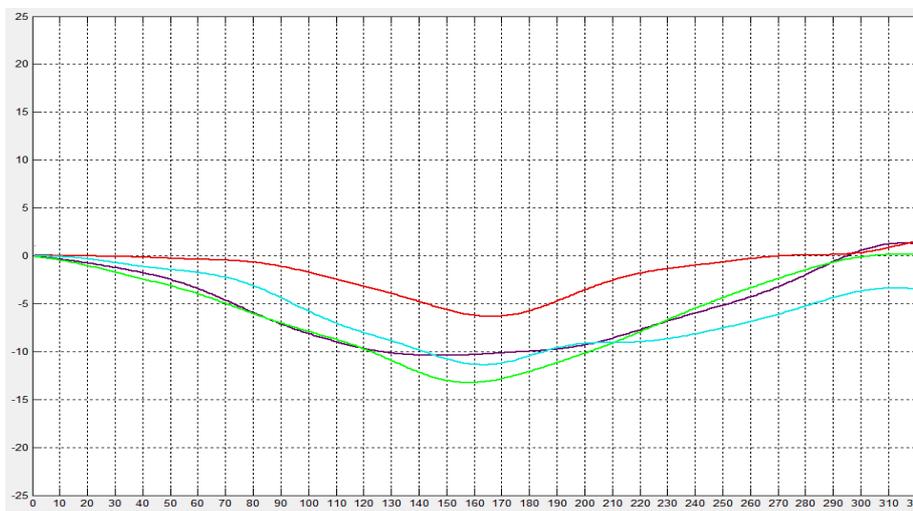
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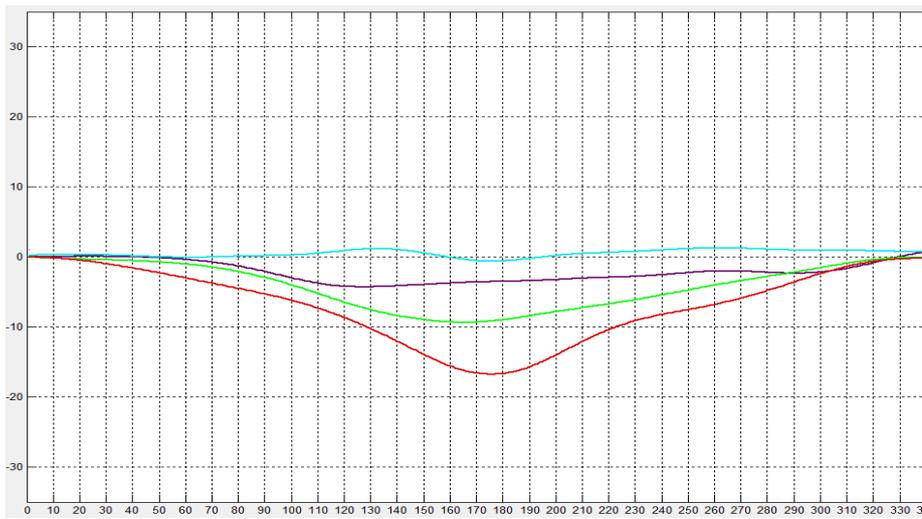
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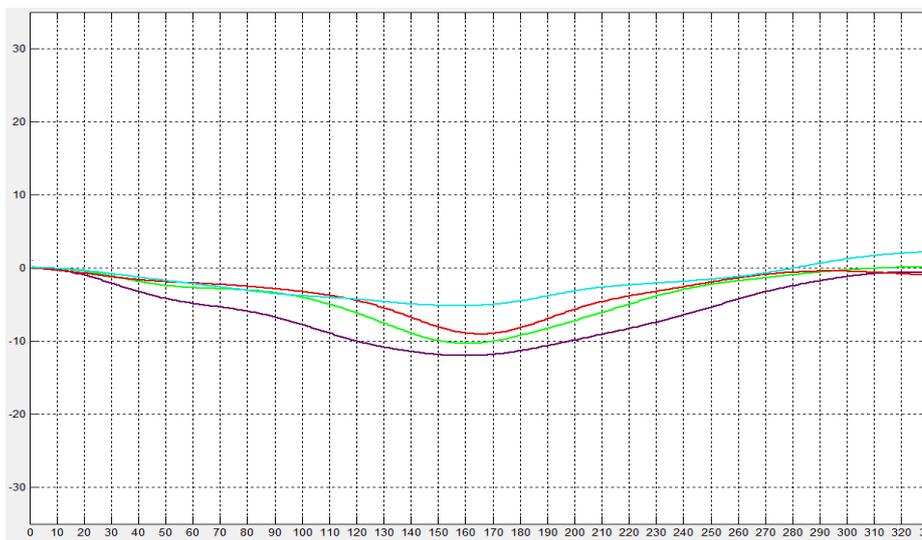
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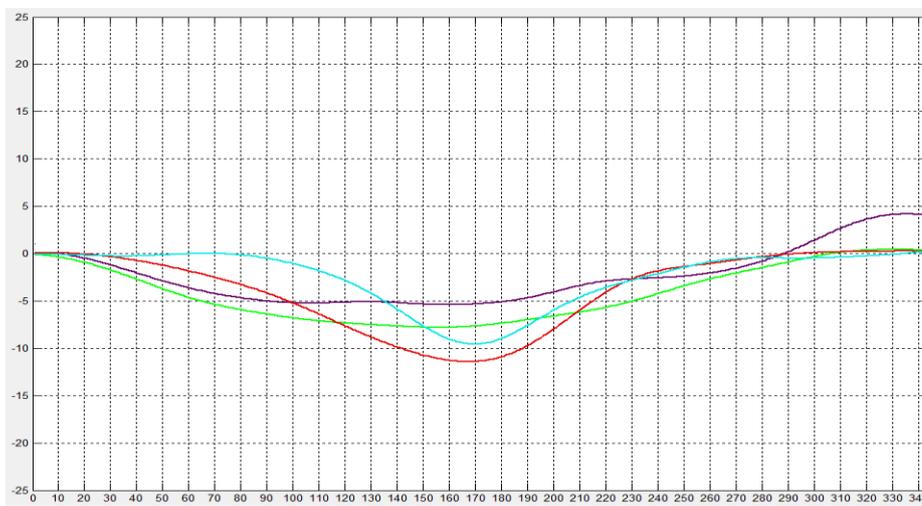
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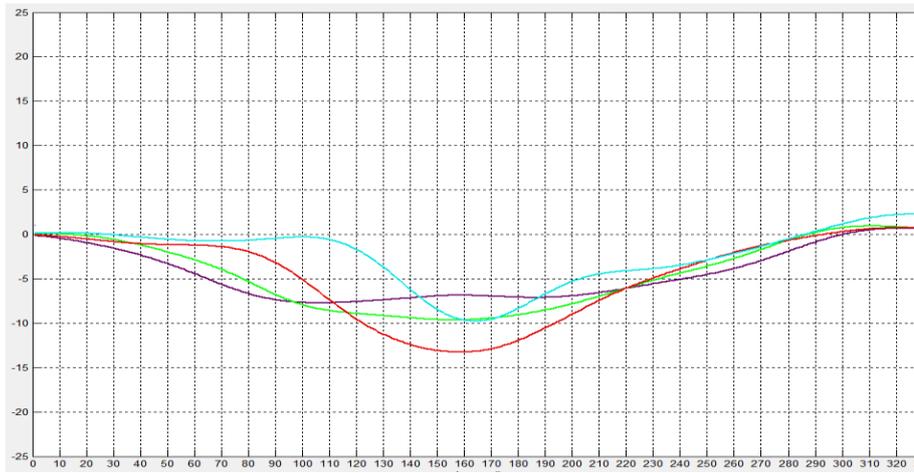
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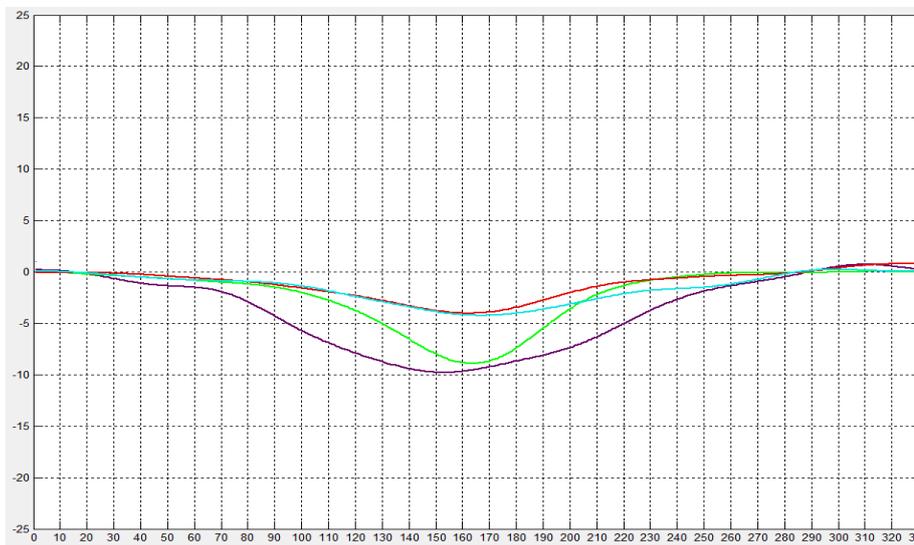
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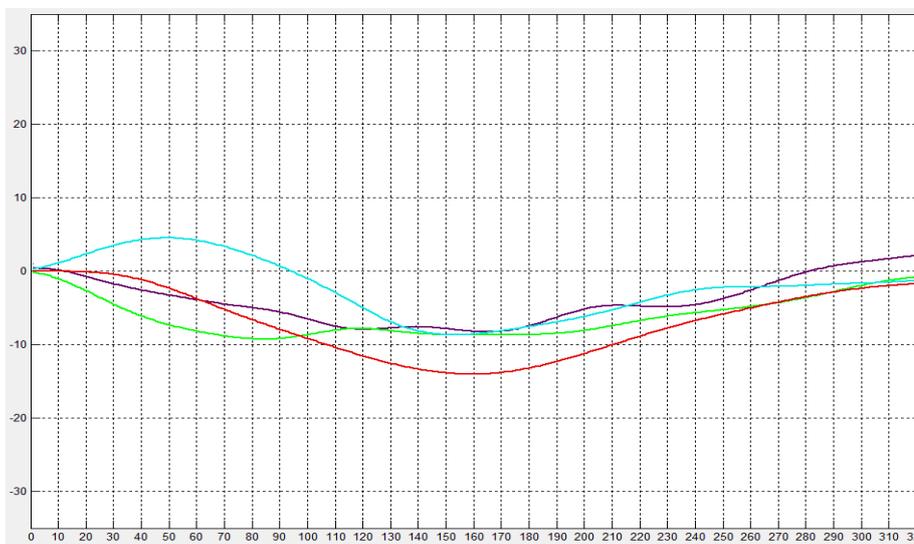
RS098



RS099



RS100



Appendix N: QAREL Checklist for IV-RoMmax and initial attainment rate repeatability studies

Quality Appraisal of Diagnostic Reliability (QAREL) Checklist

Item	Yes	No	Unclear	N/A
1. Was the test evaluated in a sample of subjects who were representative of those to whom the authors intended the results to be applied? (DEF: 3, 4, 5, 7, 8, 9)	<input checked="" type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	
2. Was the test performed by raters who were representative of those to whom the authors intended the results to be applied? (DEF 3, 4, 6, 7, 8, 9)	<input checked="" type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	
3. Were raters blinded to the findings of other raters during the study? (DEF 10)	<input checked="" type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
4. Were raters blinded to their own prior findings of the test under evaluation? (DEF 11)	<input checked="" type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
5. Were raters blinded to the results of the reference standard for the target disorder (or variable) being evaluated? (DEF 12)	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input checked="" type="checkbox"/>
6. Were raters blinded to clinical information that was not intended to be provided as part of the testing procedure or study design? (DEF 13)	<input checked="" type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
7. Were raters blinded to additional cues that were not part of the test? (DEF 14)	<input type="checkbox"/>	<input type="checkbox"/>	<input checked="" type="checkbox"/>	
8. Was the order of examination varied? (DEF 15, 16)	<input checked="" type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
9. Was the time interval between repeated measurements compatible with the stability (or theoretical stability) of the variable being measured? (DEF 17)	<input checked="" type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	
10. Was the test applied correctly and interpreted appropriately? (DEF 18)	<input checked="" type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	
11. Were appropriate statistical measures of agreement used? (DEF 19, 20, 21)	<input checked="" type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	
TOTAL				

DEF numbers relate to items on the QAREL Data Extraction Form
To access the Data Extraction Form, please go to <http://qarel.org>

Note: The inter-marker study sample included a wider age range than used in the main study, and so it could be argued that this element of the repeatability study did not meet the requirements of item 1. All participants were however healthy controls.

Appendix O: QAREL Checklist for RMS sEMG amplitude repeatability study

Quality Appraisal of Diagnostic Reliability (QAREL) Checklist

Item	Yes	No	Unclear	N/A
1. Was the test evaluated in a sample of subjects who were representative of those to whom the authors intended the results to be applied? (DEF: 3, 4, 5, 7, 8, 9)	<input checked="" type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	
2. Was the test performed by raters who were representative of those to whom the authors intended the results to be applied? (DEF 3, 4, 6, 7, 8, 9)	<input checked="" type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	
3. Were raters blinded to the findings of other raters during the study? (DEF 10)	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input checked="" type="checkbox"/>
4. Were raters blinded to their own prior findings of the test under evaluation? (DEF 11)	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input checked="" type="checkbox"/>
5. Were raters blinded to the results of the reference standard for the target disorder (or variable) being evaluated? (DEF 12)	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input checked="" type="checkbox"/>
6. Were raters blinded to clinical information that was not intended to be provided as part of the testing procedure or study design? (DEF 13)	<input checked="" type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
7. Were raters blinded to additional cues that were not part of the test? (DEF 14)	<input type="checkbox"/>	<input type="checkbox"/>	<input checked="" type="checkbox"/>	
8. Was the order of examination varied? (DEF 15, 16)	<input checked="" type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
9. Was the time interval between repeated measurements compatible with the stability (or theoretical stability) of the variable being measured? (DEF 17)	<input checked="" type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	
10. Was the test applied correctly and interpreted appropriately? (DEF 18)	<input checked="" type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	
11. Were appropriate statistical measures of agreement used? (DEF 19, 20, 21)	<input checked="" type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	
TOTAL				

DEF numbers relate to items on the QAREL Data Extraction Form
To access the Data Extraction Form, please go to <http://qarel.org>

Appendix P: Strobe checklist

STROBE Statement—checklist of items that should be included in reports of observational studies

	Item No.	Recommendation	Page No.	Relevant text from manuscript
Title and abstract	1	(a) Indicate the study's design with a commonly used term in the title or the abstract	1	Line 2
		(b) Provide in the abstract an informative and balanced summary of what was done and what was found	2	Lines 26-56
Introduction				
Background/rationale	2	Explain the scientific background and rationale for the investigation being reported	3-4	Lines 61-105
Objectives	3	State specific objectives, including any prespecified hypotheses	3-4	Lines 100-105
Methods				
Study design	4	Present key elements of study design early in the paper	4	Lines 108-110
Setting	5	Describe the setting, locations, and relevant dates, including periods of recruitment, exposure, follow-up, and data collection	4	Lines 113-114
			5	Line 123
Participants	6	(a) <i>Cohort study</i> —Give the eligibility criteria, and the sources and methods of selection of participants. Describe methods of follow-up	4	Lines 111-121
		<i>Case-control study</i> —Give the eligibility criteria, and the sources and methods of case ascertainment and control selection. Give the rationale for the choice of cases and controls		
		<i>Cross-sectional study</i> —Give the eligibility criteria, and the sources and methods of selection of participants		
		(b) <i>Cohort study</i> —For matched studies, give matching criteria and number of exposed and unexposed	N/A	
		<i>Case-control study</i> —For matched studies, give matching criteria and the number of controls per case		

Variables	7	Clearly define all outcomes, exposures, predictors, potential confounders, and effect modifiers. Give diagnostic criteria, if applicable	6	Lines 160-184
Data sources/ measurement	8*	For each variable of interest, give sources of data and details of methods of assessment (measurement). Describe comparability of assessment methods if there is more than one group	6	Lines 160-184
Bias	9	Describe any efforts to address potential sources of bias	N/A	
Study size	10	Explain how the study size was arrived at	4	Lines 116-118
Quantitative variables	11	Explain how quantitative variables were handled in the analyses. If applicable, describe which groupings were chosen and why	6-7	186-201
Statistical methods	12	(a) Describe all statistical methods, including those used to control for confounding	6-7	186-201
		(b) Describe any methods used to examine subgroups and interactions	6-7	193-201
		(c) Explain how missing data were addressed	7	204-205
		(d) <i>Cohort study</i> —If applicable, explain how loss to follow-up was addressed	7	204-205
		<i>Case-control study</i> —If applicable, explain how matching of cases and controls was addressed		
		<i>Cross-sectional study</i> —If applicable, describe analytical methods taking account of sampling strategy		
		(e) Describe any sensitivity analyses	N/A	
Results				
Participants	13*	(a) Report numbers of individuals at each stage of study—eg numbers potentially eligible, examined for eligibility, confirmed eligible, included in the study, completing follow-up, and analysed	7	204-205
		(b) Give reasons for non-participation at each stage	7	204-205
		(c) Consider use of a flow diagram	Not done	

Descriptive data	14*	(a) Give characteristics of study participants (eg demographic, clinical, social) and information on exposures and potential confounders	7	204-209
		(b) Indicate number of participants with missing data for each variable of interest	7	204-205
		(c) <i>Cohort study</i> —Summarise follow-up time (eg, average and total amount)	N/A	
Outcome data	15*	<i>Cohort study</i> —Report numbers of outcome events or summary measures over time	N/A	
		<i>Case-control study</i> —Report numbers in each exposure category, or summary measures of exposure	N/A	
		<i>Cross-sectional study</i> —Report numbers of outcome events or summary measures	N/A	
Main results	16	(a) Give unadjusted estimates and, if applicable, confounder-adjusted estimates and their precision (eg, 95% confidence interval). Make clear which confounders were adjusted for and why they were included	7	212-215
		(b) Report category boundaries when continuous variables were categorized	N/A	
		(c) If relevant, consider translating estimates of relative risk into absolute risk for a meaningful time period	N/A	
Other analyses	17	Report other analyses done—eg analyses of subgroups and interactions, and sensitivity analyses	Not done	
Discussion				
Key results	18	Summarise key results with reference to study objectives	10	Lines 269-275
			10	Lines 297-300
Limitations	19	Discuss limitations of the study, taking into account sources of potential bias or imprecision. Discuss both direction and magnitude of any potential bias	11	Lines 344-357
Interpretation	20	Give a cautious overall interpretation of results considering objectives, limitations, multiplicity of analyses, results from similar studies, and other relevant evidence	10-12	Lines 268-365

Generalisability	21	Discuss the generalisability (external validity) of the study results	11	Lines 345-348
Other information				
Funding	22	Give the source of funding and the role of the funders for the present study and, if applicable, for the original study on which the present article is based	13	Lines 393-394

RESEARCH ARTICLE

Open Access



Relationships between lumbar inter-vertebral motion and lordosis in healthy adult males: a cross sectional cohort study

Alister du Rose^{1,2*} and Alan Breen^{1,2}**Abstract**

Background: Intervertebral motion impairment is widely thought to be related to chronic back disability, however, the movements of inter-vertebral pairs are not independent of each other and motion may also be related to morphology. Furthermore, maximum intervertebral range of motion (IV-RoMmax) is difficult to measure accurately in living subjects. The purpose of this study was to explore possible relationships between (IV-RoMmax) and lordosis, initial attainment rate and IV-RoMmax at other levels during weight-bearing flexion using quantitative fluoroscopy (QF).

Methods: Continuous QF motion sequences were recorded during controlled active sagittal flexion of 60° in 18 males (mean age 27.6 SD 4.4) with no history of low back pain in the previous year. IV-RoMmax, lordotic angle, and initial attainment rate at all inter-vertebral levels from L2-S1 were extracted. Relationships between IV-RoMmax and the other variables were explored using correlation coefficients, and simple linear regression was used to determine the effects of any significant relationships. Within and between observer repeatability of IV-RoMmax and initial attainment rate measurements were assessed in a sub-set of ten participants, using the intra-class correlation coefficient (ICC) and standard error of measurement (SEM).

Results: QF measurements were highly repeatable, the lowest ICC for IV-RoMmax, being 0.94 (0.80–0.99) and highest SEM (0.76°). For initial attainment rate the lowest ICC was 0.84 (0.49–0.96) and the highest SEM (0.036). The results also demonstrated significant positive and negative correlations between IV-RoMmax and IV-RoMmax at other lumbar levels ($r = -0.64-0.65$), lordosis ($r = -0.52-0.54$), and initial attainment rate ($r = -0.64-0.73$). Simple linear regression analysis of all significant relationships showed that these predict between 28 and 42 % of the variance in IV-RoMmax.

Conclusions: This study found weak to moderate effects of individual kinematic variables and lumbar lordosis on IV-RoMmax at other intervertebral levels. These effects, when combined, may be important when such levels are being considered by healthcare professionals as potential sources of pain generation. Multivariate investigations in larger samples are warranted.

Keywords: Spine kinematics, Fluoroscopy, Lordosis, Reliability, Agreement

Background

Movement of the lumbar spine requires the participation of multiple segments and the relevant contributions of segments are a function of their own mechanical properties [1]. Aberrant spinal movement patterns are widely thought to be related to musculoskeletal pain and

dysfunction [2–4], and as such they are used to inform surgical and conservative clinical decision making [1, 5–7], and as indicators of spinal stability [3, 8–10]. As a consequence of their wide variation in both low back pain and healthy populations however, the clinical importance of factors such as inter-vertebral range of motion (IV-RoM) remains unclear, and the identification of biomechanical factors that may contribute to low back pain, remains a challenge [11]. Information about how IV-RoM may interact with other biomechanical factors may therefore help provide a better understanding of how variations in lumbar

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inter-vertebral kinematics may affect prognosis and treatment outcomes.

The starting point for this should be the collection of detailed normative quantitative data with respect to *in vivo* inter-vertebral motion and morphologic parameters [12]. Quantitative fluoroscopy (QF) has been shown to be an accurate and reliable 2D method of doing this [11, 13, 14]. Recent technological advances have enabled the acquisition of 3D lumbar kinematic data *in vivo* [15], however it has been demonstrated that there is only minimal axial rotation and lateral bending associated with movements in the sagittal plane [16–19], and in terms of QF inter-vertebral measurements, out of plane motion of up to 10° does not significantly affect accuracy [20]. Therefore, the greater expense and dose associated with current 3D techniques against the clinical and research benefits, perhaps justify the use of 2D QF technology, particularly in the sagittal plane. Indeed, the investigation of spinal mechanical behaviour has been outlined as a priority for future QF research [21], which begins with the relationships between IV-RoM and other kinematic variables in healthy, pain-free control populations. Such normative information should provide insights into the possible biomechanical consequences of changes within each.

Previous dynamic studies using fluoroscopy have highlighted contrasting ranges and patterns of angular rotation between the upper and lower lumbar motion segments [12, 22–24], which make different contributions to movements such as sagittal flexion. There is also evidence to suggest that lordosis may relate to an individual's spinal flexibility [25]. Indeed, a recent MRI study that investigated the intrinsic shape of the lumbar spine concluded that lumbar spinal shapes may be related to an individual's risk of injury [26].

IV-RoM is the most common measure of inter-vertebral motion [11, 13, 27] and attainment rate (defined as the velocity with which IV-RoM is reached), has been identified as a reflection of intervertebral restraint [11, 28, 29]. Initial attainment rate is a refinement of this which measures the slackness of an inter-vertebral motion segment in its initial phase of rotation [30, 31]. This parameter has been shown to correlate with the dynamic neutral zone [32], and is therefore also believed to be of importance when considering the stability of motion segments.

Relationships between these and other kinematic and morphologic variables have not been previously explored. This study examined the relationships between IV-RoMmax at lumbar inter-vertebral levels from L2 to S1 and lordosis, initial attainment rate and IV-RoMmax at other lumbar spine levels during forward bending in healthy controls. It also assessed the intra and inter-observer repeatability of the QF measurement of IV-RoMmax and initial attainment rate.

Methods

Study design

This was a cross-sectional, laboratory based cohort study of the relationships between L2-S1 IV-RoMmax and lordosis, initial attainment rate and IV-RoMmax at other levels.

Participants

The eligibility criteria for the study are shown in Table 1. Twenty male participants were recruited from the Anglo-European College of Chiropractic (AECC) student population over a 5 month period between May and September 2014. National Research ethics Service (NRES) approval was acquired (Bristol 10/H0106/65) and prior to the collection of data, written informed consent was obtained from each participant. A participant number of 20 was selected, as a sample size ≥ 12 has been recommended as sufficient for the precision around the measurement to be used in an exploratory study [33].

Data collection and processing

All data collection was conducted at the radiology department of the AECC. Fluoroscopic images of the lumbar spine were collected at 15 Hz using a Siemens Arcadis Avantic VC10A digital fluoroscope (CE0123) and a motion frame which acted to both stabilise the participants and guide their bending motion. Participants were asked to stand in a neutral upright position with their right side against the motion frame (Fig. 1a), and shadow a rotating arm rest which guided them during continuous fluoroscopic imaging, through a standardised range of 60° of forward flexion and return to upright, over a period of approximately 20 s. A review of spinal ranges of motion in controls proposed that the lumbar spine has an overall range (inclusive of both flexion and extension components) of approximately 80°, with 60° of this attributable to the flexion component [34]. It was therefore theorised that the majority of each participant's lumbar inter-vertebral movement would be captured within this range.

Prior to image acquisition, participants were taken in 20° stages through to the full 60° to safeguard that they were able to tolerate the movement. The movement of the motion frame was recorded by electronic feedback from its motor drive, and synchronised with the fluoroscopic imaging. To minimise bending from the hip joints, the pelvis was stabilised using a strap secured around the anterior superior iliac spine bilaterally, and attached to an appendage of the motion frame directly posterior to the participant (Fig. 1b).

A lead apron was worn to shield the gonads, and participants were verbally reminded to maintain a neutral bending position during the flexion cycle. The position of the central ray was targeted at L4 to make sure

Table 1 Eligibility criteria

Inclusion	Exclusion
Males aged 20–40 years	Inadequate understanding of English
An ability to understand written information	Currently receiving treatment for osteoporosis
Willing to participate and able to give informed consent	A history of recent abdominal or pelvic surgery
Consent to General Practitioner being informed	A history of previous lumbar spine surgery
A BMI of <30	A BMI of >30
No history of low back pain that prevented normal activity for at least 1 day in the previous year	Any medical radiation exposure in the past year or exposure in the past 2 years with a dose greater than 8 mSv
	Involvement in any other ongoing research study

that all vertebrae (L2-S1) were included in the image field Fig. 2.

The fluoroscopic sequences were then transferred to a desk top computer for analysis using bespoke image processing codes written in Matlab (The Mathworks, Cambridge). Using the screen cursor, the outlines of each vertebra from L2-S1 in the first image of each sequence were marked-up manually with an electronic template. In order to increase precision, this process was replicated five times for each sequence and the results were averaged. In all subsequent image frames the bespoke software tracked each vertebra automatically, creating a continuous measurement of its movement throughout the flexion and return bending sequence. To ensure that template tracking was maintained throughout the sequence, visual checks were made using video playback.

The data collected comprised of range of motion (IV-RoM), initial attainment rate, and lordosis and the reliability and agreement of the first two of these were assessed as part of the study [35]. The technique used to measure changes in inter-vertebral angle is discussed in detail elsewhere [36], and is shown in Fig. 3.

IV-RoMmax for each inter-vertebral level (L2-S1) was calculated as the maximum angular range reached at any point throughout the 60° flexion and return cycle (Fig. 4). Initial attainment rate for each level was calculated as the ratio of the slopes of motion frame movement and the inter-vertebral rotation over the first 10° immediately following the onset of inter-vertebral motion. The calculation of this variable has been outlined in detail elsewhere [31], and is also shown in Fig. 5. Lordosis was measured as the sum of all inter-vertebral angles (L2-S1), from the first image in the sequence. All participant data were anonymised.

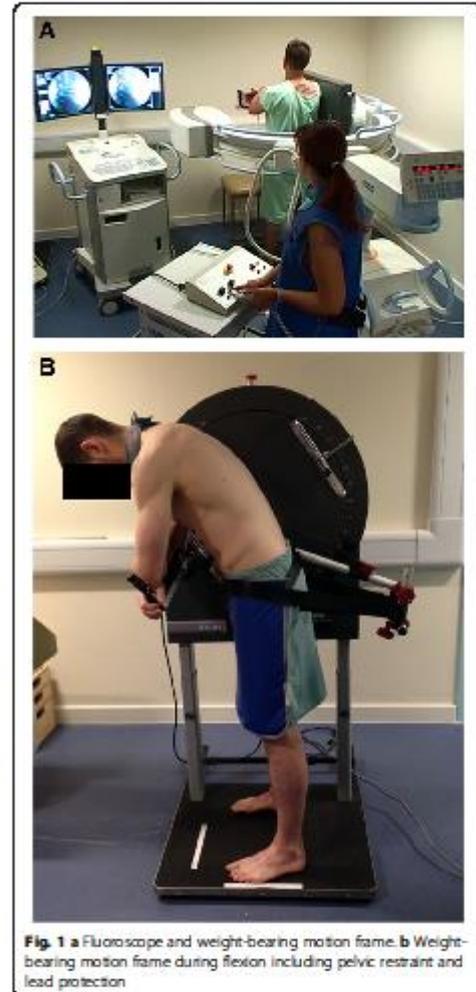
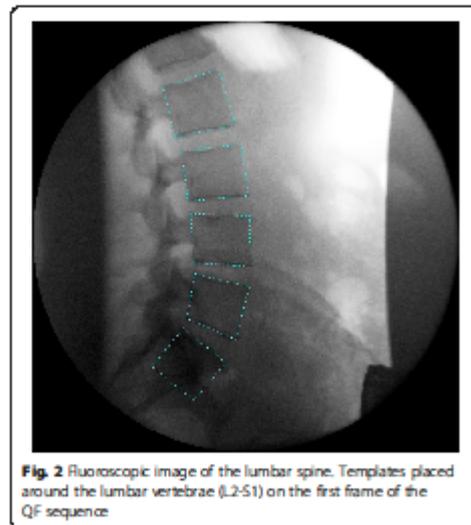


Fig. 1 a Fluoroscope and weight-bearing motion frame. b Weight-bearing motion frame during flexion including pelvic restraint and lead protection

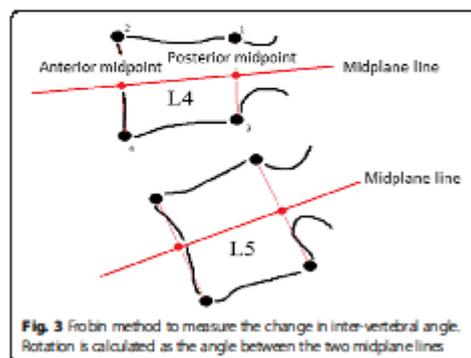
Reliability and agreement

A convenience sample of ten participants was used for the intra- and inter-observer repeatability studies. The intra-observer study was conducted, with a 6 week separation between image mark-ups. The inter-observer study images were processed by two independent observers. The first observer (template marker) was a medical physicist, and the second was ADR. The observers were blinded to the others’ results, and had 3 and 1 year(s) experience of template marking respectively.



Data analysis

The normality of all data were tested using the Shapiro-Wilk test. Relationships between IV-RoMmax and other biomechanical variables, from normally distributed data were analysed using the Pearson product-moment correlation coefficient, and non-normal data using the Spearman's Rank Correlation. Any significant relationships (*p* values < 0.05) were also analysed using simple linear regression. Intra- and inter-observer reliability and agreement of both IV-RoMmax and initial attainment rate measurements, were assessed using intra-class correlations (ICC 3, 1), and the standard error of measurement (SEM) respectively. Statistical analysis was performed using IBM SPSS (version 21).



Note: This study conformed to the STROBE checklist for reports of observational studies [37] (Additional file 1).

Results

Twenty males satisfying the eligibility criteria consented to participate. Template tracking failure occurred in two participant's sequences, and their data were removed. The mean (SD) age, height, and body mass Index (BMI) were 27.6 years (4.4), 1.8 m (0.06), and 24 (2.2) respectively. The average radiographic exposure factors for the group were documented as 79.7 kV SD (5.4) and 55.4 mA SD (3.4). ICRP103 conversion software PCXMC (Monte Carlo Simulation Package) was used to calculate the mean effective dose as 0.143 mSv.

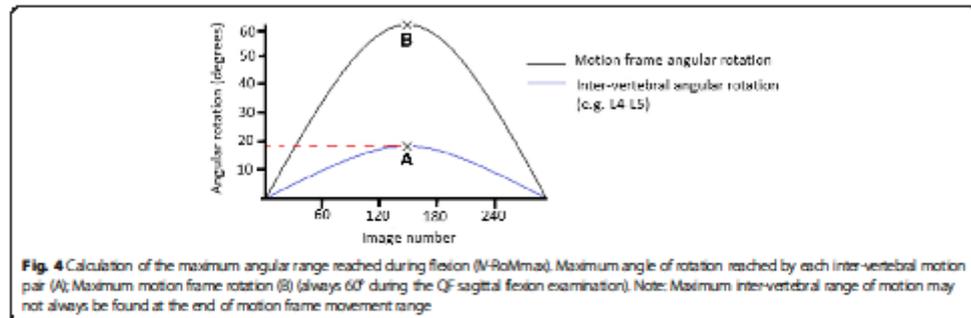
Reliability and agreement

IV-RoMmax

The ICC's (reliability) and SEM's (agreement) for both intra- and inter-observer IV-RoMmax studies are shown in Table 2. The results suggest excellent reliability with the smallest ICC being 0.96 (95 % CI 0.82–0.99) and 0.94 (95 % CI 0.80–0.99) for the intra- and inter-observer studies respectively. When comparing intra- and inter-observer repeatability, it was expected that intra-observer comparisons would demonstrate better reliability and agreement [11, 14]. This trend was not observed in these weight-bearing samples however, and ICC's were the same or slightly better in the inter-observer group for two out of the four inter-vertebral levels. Agreement was found to be better than 1° at all levels, for both intra- and inter-observer studies.

Initial attainment rate

The ICC's (reliability) and SEM's (agreement) for both intra- and inter-observer weight-bearing initial attainment rate studies are also shown in Table 2. The reliability of initial attainment rate measurements was also acceptable, being more than 0.81 [38] in both intra- and inter-observer studies at all inter-vertebral levels. The smallest ICC was 0.84 (95 % CI 0.49–0.96) at the level of L3-L4 in the inter-observer study, and the largest was 0.98 (95 % CI 0.92–1.0) in the intra-observer study at the same level. The intra-observer study demonstrated consistently better reliability (including narrower confidence intervals) than that of the inter-observer study. The agreement of initial attainment rate measurements is also acceptable in both intra- and inter-observer studies. In the upper inter-vertebral levels (L2-3 and L3-4) SEM's are comparatively lower in the intra-observer study, however in the lower levels (L4-L5 and L5-S1) SEM's are comparatively higher.



Correlations

A summary of the correlations between all biomechanical variables and IV-RoMmax is given in Table 3. Significant correlations were found between IV-RoMmax and at least one other variable at all inter-vertebral levels. These were consistently of mid-level strength (r - values ranging from -0.64 to 0.73). Lordosis was positively correlated with IV-RoMmax at L2-L3 and negatively with L4-5 ($r = 0.54$ and -0.52 respectively). In terms of IV-RoMmax versus IV-RoMmax at other levels, correlations were found between all levels except L5-S1. L2-L3 range was shown to be positively correlated with that of L3-4, but negatively correlated with L4-5. The strongest relationship was found between initial attainment rate at L3-4 and L5-S1 IV-RoMmax ($r = 0.73$). Indeed, initial attainment rate showed examples of strong correlations with range at all levels.

Simple linear regression analysis

The coefficients of determination (r^2) for each of the significant correlations are shown in Fig. 6 (a-h). The values range from (0.28 to 0.42) and demonstrate that IV-RoMmax, can be influenced by lordosis, the IV-RoMmax at other lumbar levels, and initial attainment

rate. Figure 6a for example shows that 41 % of the variability in L4-L5 IV-RoMmax can be accounted for by the range of L2-L3 IV-RoMmax.

Discussion

Agreement and reliability

The agreement and reliability of IV-RoM and initial attainment rate measurements using continuous QF image data has previously been assessed in recumbent participants [11]. As anticipated, the reliability and agreement of IV-RoM measurements during recumbent sagittal flexion were found to be similar to those found in the current work, with 'substantial' reliability [38], and acceptable error (i.e. $<1^\circ$) demonstrated at all levels for both intra- and inter-observer studies. It has been demonstrated that reliability and agreement are typically decreased in the inter-observer group [11, 14], however these differences were shown to be minimal in the current study, and there were notable exceptions to the trend (Table 2). Although ICCs were very similar between intra- and inter-observer groups, generally the width of the CIs and the SEMs were larger in the latter. It appears that whilst errors arising from the use of different observers did have a small impact, inter-observer agreement and reliability is still acceptable.

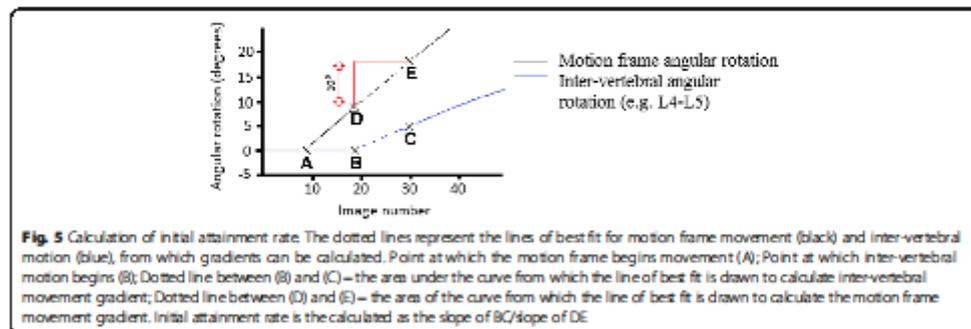


Table 2 Intra- and inter-observer reliability and agreement of IV-RoMmax and initial attainment rate measurements during weight-bearing flexion and return *n* = 10

Inter-vertebral level	Intra-observer ICC (95 % CI)	Inter-observer ICC (95 % CI)	Intra-observer SEM (°)	Inter-observer SEM (°)
IV-RoMmax				
L2-L3	0.98 (0.92–1.0)	0.94 (0.80–0.99)	0.45	0.76
L3-L4	0.99 (0.96–1.0)	0.99 (0.67–1.0)	0.23	0.24
L4-L5	0.99 (0.97–1.0)	0.98 (0.93–1.0)	0.39	0.59
L5-S1	0.96 (0.82–0.99)	0.99 (0.94–1.0)	0.54	0.61
Inter-vertebral level	Intra-observer ICC (95 % CI)	Inter-observer ICC (95 % CI)	Intra-observer SEM _{max}	Inter-observer SEM _{max}
Initial attainment rate				
L2-L3	0.95 (0.78–0.99)	0.95 (0.80–0.99)	0.026	0.036
L3-L4	0.98 (0.92–1.0)	0.84 (0.49–0.96)	0.02	0.033
L4-L5	0.92 (0.71–0.98)	0.91 (0.70–0.98)	0.032	0.018
L5-S1	0.95 (0.81–0.99)	0.88 (0.53–0.97)	0.023	0.019

It was anticipated that there might be a marginal decrease in the repeatability of measurements at inter-vertebral levels closer to the edge of the image field (i.e. L2-3 and L5-S1). This predicted difficulty (due to superimposition of the ilia) in template marking/tracking of L5-S1 was the reason cited by Mellor et al. [11] for its exclusion, and it makes sense that tracking problems could be more likely to occur in templates that may partially leave the image field i.e. L2-3 and L5-S1. The current study's results have shown however, that reliability and agreement of QF IV-RoMmax measurements at all levels, including L2-L3 and L5-S1, can be achieved in a weight-bearing protocol, at an acceptable level.

The initial attainment rate measurements were also highly repeatable; there was however a clearer distinction between the confidence intervals of intra- and inter-observer groups, being notably wider in the latter at L3-4 and L5-S1 levels (Table 2). These may be best explained by differences in the experience levels of the observers [14]. Differences in marking experience may also be the

reason for the relatively increased measurement error in the inter-observer group at upper inter-vertebral levels, and improved agreement at the lower levels.

Correlations

The results showed evidence of relationships between kinematic variables at multiple levels of the lumbar spine. IV-RoMmax at all inter-vertebral levels was significantly correlated, positively or negatively, with at least one other kinematic or morphological variable, and there appear to be trends in these relationships in terms of the regions of the lumbar spine. Previous studies have demonstrated that during sagittal flexion, upper lumbar segments typically move first [24], and have the greatest ranges of motion [39]. In this study, L2-3 and L3-4 were strongly positively correlated, suggesting that they tend to work in tandem. However, they were also both negatively correlated with the IV-RoMmax of L4-5 and its initial attainment rate. This suggests a compensatory function between them, and if it is assumed that instability results from reduced restraint, then it may be suggested that reduced motion at

Table 3 Correlations between kinematic variables and IV-RoMmax at all inter-vertebral levels *n* = 18 (Significant relationships are highlighted in bold)

Kinematic variable	L2-L3 IV-RoMmax		L3-L4 IV-RoMmax		L4-L5 IV-RoMmax		L5-S1 IV-RoMmax	
	r	p	r	p	r	p	r	p
Lordosis	0.54	0.021	0.401	0.099	-0.52	0.026	-0.02	0.973
L2-L3 IV-RoMmax	-		0.65	0.003	-0.64	0.004	-0.35	0.157
L3-L4 IV-RoMmax	0.65	0.003	-		-0.29	0.234	-0.12	0.636
L4-L5 IV-RoMmax	-0.64	0.004	-0.29	0.234	-		0.15	0.558
L5-S1 IV-RoMmax	-0.35	0.157	-0.12	0.636	0.15	0.558	-	
L2-L3 Initial attainment rate	0.20	0.419	0.14	0.58	-0.09	0.713	0.21	0.403
L3-L4 Initial attainment rate	-0.18	0.465	-0.11	0.668	0.17	0.512	0.73	0.001
L4-L5 Initial attainment rate	-0.53	0.023	-0.64	0.004	0.59	0.009	0.02	0.949
L5-S1 Initial attainment rate	0.05	0.852	-0.02	0.938	0.07	0.776	0.42	0.079

reduction in lumbar lordosis, which has implications for proposed surgical stabilisation of upper levels. Conversely, while attainment rate and IV-RoMmax at L4-5 were positively correlated, L4-5 attainment rate was negatively correlated with the IV-RoMmax at L2-3 and with L3-4 above. As both attainment rate and IV-RoMmax are expressions of intervertebral restraint, these relationships may be regarded as compensatory, contributing to the attenuation of stress throughout the lumbar spine linkages. Thus there are indications of interactions and effects between kinematic and morphological variables at different levels.

Finally, there is an increasing awareness of the importance of sagittal parameters when planning surgical strategy [41, 42], correcting sagittal balance, or when considering more conservative treatment options. The ability to accurately assess and measure sagittal kinematic and morphological values may be important, as we attempt to understand their potential clinical utility [43]. The existence of intrinsic links between morphological variables such as lordosis have been described before [44], however we are the first to use continuous *in vivo* inter-vertebral motion to investigate its links with IV-RoMmax and initial attainment rate. These results provide clues as to what may happen when kinematic or morphological changes are imposed through conservative treatment or surgery, both as local and regional effects. The apparent inter-dependency may assist in building rationales for treatments, and highlights the need to account for factors such as lordosis when conducting kinematic studies. If the results are re-affirmed by multivariate investigations in larger samples, future longitudinal studies are recommended to investigate the effect of interventions in low back pain populations, that have been informed by the relationships described in this study.

Limitations

The study's results are only representative of one young, healthy, male population and replication with larger and more extensive populations would be required to explore the relationships in wider age groups and in females. In light of this, any discussions relating to the investigation and management of wider LBP populations warrant careful consideration. Furthermore, it was also not possible to address the impact of loading on spinal behaviour, although every effort was made to standardise the population sample and study protocol for body mass index. In this research all measurements were made during weight-bearing, and therefore the effect of muscle activity is also a consideration. A concurrent study conducted by our research group examines the relationships between lumbar paraspinal muscle activity and the kinematic and morphological variables described here [45]. Future studies may also wish to consider the use of dynamic stereo x-ray imaging [15], especially if investigation of rotation in the

transverse or coronal planes is required, where associated out of plane movements are more prominent.

Conclusions

IV-RoMmax and initial attainment rate measurements made using a QF weight-bearing sagittal plane protocol demonstrated acceptable reliability and agreement. Significant correlations were found between IV-RoMmax, IV-RoMmax at different inter-vertebral levels, lordosis and initial attainment rate. The study demonstrated weak to moderate effects of these variables on IV-RoMmax. The potential prognostic and treatment effects of these relationships merit exploration with multivariate studies in larger samples, potentially leading to longitudinal investigations in back pain populations.

Additional file

Additional file 1: STROBE Statement—checklist of items that should be included in reports of observational studies (DOCX 40 kb)

Abbreviations

AECC: Anglo-European College of Chiropractic; BMI: body mass index; CI: confidence interval; IBM SPSS: International Business Machines Corporation Statistical Package for the Social Sciences; ICC: Intraclass correlation coefficient; IV-RoM: inter-vertebral range of motion; M-RoMmax: maximum inter-vertebral range of motion; MRI: magnetic resonance imaging; msv: millisievert; n: number; NRES: National Research Ethics Service; QF: quantitative fluoroscopy; SD: standard deviation; SEM: standard error of measurement.

Competing interests

The authors declare that they have no competing interests.

Authors' contributions

ADR served as project lead, performed the statistical analysis, conducted the image analysis and wrote the first draft. AB operated the motion table. Both authors contributed to the study design, procedural protocol, data acquisition and the drafting of this paper. Both authors read and approved the final manuscript.

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Article

Relationships between Paraspinal Muscle Activity and Lumbar Inter-Vertebral Range of Motion

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Abstract: Control of the lumbar spine requires contributions from both the active and passive sub-systems. Identifying interactions between these systems may provide insight into the mechanisms of low back pain. However, as a first step it is important to investigate what is normal. The purpose of this study was to explore the relationships between the lumbar inter-vertebral range of motion and paraspinal muscle activity during weight-bearing flexion in healthy controls using quantitative fluoroscopy (QF) and surface electromyography (sEMG). Contemporaneous lumbar sEMG and QF motion sequences were recorded during controlled active flexion of 60° using electrodes placed over Longissimus thoracis pars thoracis (TES), Longissimus thoracis pars lumborum (LES), and Multifidus (LMU). Normalised root mean square (RMS) sEMG amplitude data were averaged over five epochs, and the change in amplitude between epochs was calculated. The sEMG ratios of LMU/LES LMU/TES and LES/TES were also determined. QF was used to measure the maximum inter-vertebral range of motion from L2-S1, and correlation coefficients were calculated between sEMG amplitude variables and these measurements. Intra- and inter-session sEMG amplitude repeatability was also assessed for all three paraspinal muscles. The sEMG amplitude measurements were highly repeatable, and sEMG amplitude changes correlated significantly with L4-5 and L5-S1 IV-RoMmax ($r = -0.47$ to 0.59). The sEMG amplitude ratio of LES/TES also correlated with L4-L5 IV-RoMmax ($r = -0.53$). The relationships found may be important when considering rehabilitation for low back pain.

Keywords: spine kinematics; fluoroscopy; surface electromyography; reliability; agreement

1. Introduction

Optimal control of the spine during voluntary trunk bending requires fine-tuned coordination of numerous trunk muscles [1]. This dynamic control is believed to be modulated by communication between three sub-systems, the passive (vertebrae, discs, and ligaments), the active (muscles and tendons), and the control (central nervous system and nerves) systems [2,3]. Investigating the interplay between sub-systems however is difficult, as the spine is a complex structure; and a hidden kinematic chain. Several different technologies are therefore typically required, each with their own limitations.

In order to directly investigate the passive and active sub-systems of the spine, there have been many efforts to concurrently measure spinal kinematics and muscle activity [4–12]. The majority of these studies have used surface electromyography combined with skin surface kinematic measurement techniques such as Fastrak [8,13], Isotrak [9,11,12], or cameras [4,5,7]. These are limited to the investigation of gross spinal motion. To include segmental data usually requires invasive techniques

such as the surgical insertion of intra-osseous pins. In this way Kaigle *et al.* (1998) investigated the reduction in lumbar muscular activity during full flexion (flexion relaxation) and spinal kinematics at an inter-vertebral level [10]. However, typically only single motion segments were considered, and EMG was also only recorded from one level (e.g., lumbar longissimus thoracis) [10].

1.1. Contemporaneous Monitoring of Inter-Vertebral Passive and Motor Control Systems

Study of the integrated function of the joints and muscles of the spine requires contemporaneous multi-level kinematic and electromyographic monitoring throughout the motion. This is necessary to incorporate timing, magnitude, and segmentation in the two systems to characterise control. Multi-level surface electromyography fulfils these requirements for motor control and quantitative fluoroscopy measures a range of continuous inter-vertebral motion variables [14]. Contemporaneous recording of these measures therefore provides an integrated assessment of the passive and active systems of the spine, and it is proposed that this may be useful when assessing patients with low back pain (LBP) [4,15]. This study therefore deployed quantitative fluoroscopy (QF), and surface electromyography (sEMG) of the lumbar spine together for the first time. The study investigated the biomechanics of the lumbar spine in a healthy control population in order to potentially better understand the significance of biomechanical changes in LBP populations.

1.2. Variable Selection

In order to investigate relationships between segmental kinematics and local muscle activity, suitable variables from each must be identified. While responses to perturbation [16], and the flexion relaxation phenomenon (an absence of paraspinal muscle activity during full sagittal flexion (FRP)) have been investigated [17,18], few studies have included sEMG amplitude changes throughout the cycle, be they increases or decreases. This study therefore addressed these parameters. QF measures continuous intervertebral rotation and translation in the coronal and sagittal planes during weight-bearing or recumbent motion and can also extrapolate the instant axis of rotation (IAR) and rotational range attainment rate from this. However, the need to also compare intervertebral range of motion (IV-RoM) with sEMG in the present studies, dictates the need for continuous motion information. Therefore IAR rotation and attainment rate were not likely to be so useful. In addition, the small ranges of translation make this measure unsuitable for numerical comparisons, leaving maximum rotational motion as the preferred measure.

To investigate the relationships between lumbar muscle activity and inter-vertebral restraint during bending requires access to the maximum IV-RoM (IV-RoMmax). Continuous intervertebral rotation data allows both temporal comparisons with other variables and the actual maximum IV-RoM (IV-RoMmax), rather than IV-RoM at the limit of voluntary trunk bending, to be extracted. Recording in the standing orientation allows these comparisons.

1.3. Enhanced Functional Assessment

Sanchez-Zuriaga *et al.* (2015) suggest that there are only subtle differences between various low back patient groups and healthy controls in terms of paraspinal muscle activity and regional lumbar movement [4]. This means that either muscle activity has no effect on the range of motion, or that we are missing the detail of what is happening at individual levels. For example it may be that whereas there is an increase in paraspinal activity in recurrent LBP patients during flexion, but no difference in RoM, the share of RoM may have shifted between levels at different stages in the motion. The primary role of the paraspinal muscle during flexion is to resist inter-vertebral motion [19] and so it may be that the motion is restricted at a specific level, and compensated for elsewhere, be this at other lumbar levels, or in the thoracic spine or pelvis. It is essential therefore, when attempting to understand the relationships between functional impairments and LBP that specific inter-vertebral levels are assessed both in terms of kinematics and associated muscle activity.

1.4. Repeatability

The development of QF techniques has seen its use in LBP research become more common [20–22]. IV-RoM has been the most common QF measure of inter-vertebral motion [22–24], where it has been shown to be accurate and reliable [22,25]. It is known however that sEMG recordings, by contrast, are inherently variable [26,27]. Therefore, a sub-study was conducted to assess the intra and inter-session repeatability (reliability and agreement) of the mean normalised root mean square (RMS) sEMG amplitude recordings from the entire flexion and return cycle.

1.5. Aim of the Study

The purpose of this study was to quantify the relationships between IV-RoMmax during flexion of the lumbar spine with the accompanying paraspinal muscle activity.

1.6. Specific Objectives

To determine the inter- and intra-session reliability and agreement of normalised sEMG amplitudes during weight-bearing sagittal flexion and return.

- To determine whether ratios of inter-level lumbar paraspinal sEMG amplitudes are related to the IV-RoMmax at lumbar inter-vertebral levels.
- To determine whether changes in sEMG amplitudes during different phases of the forward bending cycle are related to IV-RoMmax at lumbar inter-vertebral levels.

2. Experimental Section

2.1. Participants

The eligibility criteria for the study are shown in Table 1. Twenty male participants from the Anglo-European College of Chiropractic (AECC) student population were recruited. National Research Ethics Service (NRES) approval was gained for the study (Bristol 10/H0106/65) and written informed consent was obtained from all participants prior to data collection. The QF and sEMG data collection was conducted concurrently. In order to minimise the potential impact of variations in parameters such as soft tissue thickness (STT) and spinal degeneration (e.g., reduced disc heights), recruitment was restricted to young adult males.

Table 1. Eligibility criteria.

Inclusion	Exclusion
Males aged 20–40 years	Poor understanding of English
Able to understand written information	Having treatment for osteoporosis
Willing to participate and able to give informed consent	Recent abdominal or pelvic surgery
Consent to GP being informed	Previous lumbar spine surgery
BMI < 30	BMI > 30
No history of low back pain that prevented normal activity for at least one day in the previous year	Any medical radiation exposure in the past year or exposure in the past two years with a dose greater than 8mSv
	Current involvement in any other research study

2.2. Kinematic Data Collection and Processing (Quantitative Fluoroscopy)

Lumbar spine fluoroscopic images were collected at 15 Hz using a Siemens Arcadis Avantic VC10A digital fluoroscope (CE0123) and an upright motion frame, which stabilised participants and guided their bending motion. Participants were asked to stand with their right side against the motion frame (Figure 1), and follow a rotating arm rest which guided them through a range of 60° of forward

flexion and a return to upright during continuous fluoroscopic imaging over a period of 20 seconds. A range of 60° was selected on the basis that the lumbar spine has an overall range of 80° (Flexion and extension components) [28]. The motion frame apparatus could be fully adjusted in accordance with the participant's stature, and the central ray was positioned at L4 to ensure that all vertebrae (L2-S1) were included in the image field.

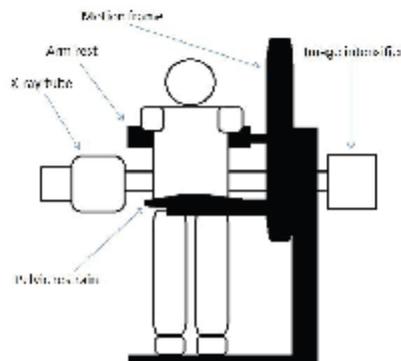


Figure 1. Fluoroscope and motion frame set-up.

Before image acquisition commenced, participants were taken in 20° increments through to the full 60° to ensure that they were able to tolerate the motion. The movement of the motion frame was recorded by electronic feedback from its motor drive and synchronised with the fluoroscopic imaging. To avoid bending at the hip joints, the pelvis was stabilised using a belt secured around the anterior superior iliac spine and secured to a bracing pad placed against the lower sacral segments. A lead apron was worn to shield the gonads.

Flexion and return sequences were then transferred to a desktop computer for analysis using bespoke image processing codes written in Matlab (The Mathworks, Cambridge) [14]. The vertebral outlines from L2-S1 in the first image in each sequence were manually marked with an electronic template using the screen cursor. This process was repeated five times for each sequence and the results averaged to increase precision. In each subsequent image frame the software programme automatically tracked each vertebra, producing a continuous measurement of its movement throughout the bending sequence [14]. Template tracking was checked visually via video playback to ensure the templates maintained the correct alignment throughout the sequence.

The data extracted comprised the continuous inter-vertebral angle in flexion and the IV-RoMmax. IV-RoMmax for each inter-vertebral level (L2-3, L3-4, L4-5 and L5-S1) was calculated as the maximum angular range reached at any point throughout the 60° trunk flexion and return cycle.

2.3. Electromyography

Prior to the commencement of data collection, participants lay prone in order for 12 electrode sites to be marked on their backs with a skin pencil. In preparation for this, the skin over their lower backs was prepared for sEMG electrode application by light abrasion, cleaning with an alcohol swab, and when necessary, shaving of the area. Disposable pre-gelled self-adhesive Ag-AgCl electrodes were then applied over three bilateral muscle groups with a 20 mm centre-to-centre inter-electrode distance as follows: Thoracic erector spinae (TES) (5 cm lateral to the T9 spinous process) [12,29], the lumbar erector spinae (LES), and lumbar multifidus (LMU) (2 cm lateral to the L2 and L5 spinous processes) [18,30] whilst the participant was in slight flexion (Figure 2).

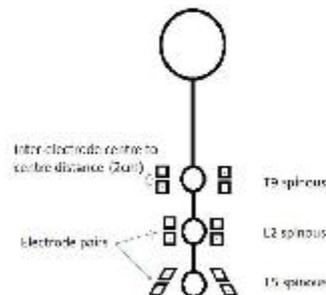


Figure 2. Electrode positioning sites. (Note: T9 spinous refers to the spinous process of the ninth thoracic vertebra, L2 to the second lumbar vertebra and L5 to the fifth lumbar vertebra.)

Although cross talk from multiple muscles will inevitably contribute to the signal recorded at each electrode site, cross-sections of the spine at each electrode site showed that the muscles that will predominate at T9 (TES) and L2 (LES) is longissimus thoracis, and at L5 (LMU) multifidus [31]. Three Biopac wireless transmitters (Bionomadix Dual Channel Wireless EMG) were then placed on the lower back attached by self-adhesive Velcro pads. There was no significant difference between the normalised mean sEMG amplitudes recorded over left and right sides during the flexion and return cycle. Therefore, an average of the mean amplitudes from both sides was used for all analysis [15].

2.4. Electrode Positioning Accuracy

Electrode application accuracy is dependent on the subjective identification of bony anatomical landmarks, and current methods used are therefore limited by human subjectivity and variation in individual anatomy [32–35]. It has been suggested however that accuracy can be improved significantly when techniques are combined [36]. This investigation was integrated into a larger ongoing normative database study, which required recumbent QF imaging before weight-bearing imaging commenced. In order to improve electrode positioning accuracy, an electrode was placed over the spinous process of L3 during the recumbent protocol. This provided an improved anatomical reference point for the application of the electrodes (Figure 3).



Figure 3. An electrode placed over the spinous process of L3.

2.5. The sEMG Equipment

The sEMG signal data were recorded at a sampling rate of 2000 Hz using a common-mode rejection ratio (CMRR) of 110 dB and an input impedance of 1000 MOhms.

The six signals were band pass filtered at 10–500 Hz and full wave rectified. The root mean square (RMS) amplitude was calculated for individual participant cycles and normalised during post-processing to sub-maximal voluntary contractions expressed as a percentage of the sMVC.

2.6. Reference Contraction

When data collection had been completed, and in order to provide a sub-maximal reference contraction (sMVC) [37], participants were asked to lie prone on a padded bench with their hands behind their head. They were then required to raise their torso off the couch and hold this position for five seconds whilst their legs and pelvis were stabilised. This process was repeated three times and the average sMVC was used as a reference. This technique was selected over a normalisation to a peak, primarily due to the even loading of the investigated muscle groups, but also to avoid the problem of variations in participant's muscle activation patterns in order to produce the same movement.

2.7. Synchronisation

The QF motion frame controller recording and the sEMG data recording were co-ordinated using a trip switch attached to the motion arm of the frame. This registered a data point on the sEMG timeline (Figure 4).



Figure 4. Synchronisation of the motion frame movement and sEMG recordings.

2.8. The sEMG Amplitude Repeatability Study

A separate convenience sample of 10 participants was used for the sEMG amplitude intra- and inter-subject repeatability studies. These studies were done without QF imaging. The acquisition cycle was repeated four times (several minutes apart) at baseline and follow up. Intra-session results compared cycles 1 and 2 (of the four), whereas inter-session results were calculated as an average of the four mean (left and right) normalised amplitudes recorded over the cycle duration. All analysis was conducted by ADR.

2.9. Data Analysis

sEMG ratios [38,39] were calculated from the mean left-right normalised sEMG (RMS) amplitudes during the flexion phase only as follows, LMU/LES, LES/TES and LMU/TES. In order to calculate sEMG changes at different stages of the flexion cycle, the forward bending phase was divided into five epochs for each participant [15]. The change in mean sEMG between epochs was then calculated (e.g.,

the change during the early stage of flexion was calculated as (epoch 1–2) for each of TES, LES, and LMU). This was repeated to determine changes between all epochs at all levels.

All data were tested for normality using the Shapiro-Wilk test. Relationships between IV-RoMmax and sEMG ratios and changes from normally distributed data were analysed using the Pearson product-moment correlation coefficient, and non-normal data using the Spearman's Rank Correlation. Significant relationships (p values < 0.05) were further analysed using simple linear regression. Intra-subject reliability and agreement of the mean normalised RMS sEMG amplitudes throughout the flexion and return cycle were assessed using intra-class correlations (ICC 3, 1) [40], and the standard error of measurement (SEM) respectively [41]. Statistical analysis was performed using IBM SPSS (version 21).

3. Results and Discussion

3.1. Results

Twenty males with no history of low back pain over the previous year consented to participate. Failed template tracking occurred in two participants' sequences, and their QF and sEMG data were therefore discarded. The mean (SD) age, height, and body mass Index (BMI) were 27.6 years (4.4), 1.8 m (0.06), and 24 (2.2), respectively. Average radiographic exposure factors for the group were recorded as 79.7 kV SD (5.4) and 55.4 mA SD (3.4). The mean effective dose was calculated using ICRP103 conversion software PCXMC (Monte Carlo Simulation Package) [42], and was 0.143 mSv. A complete motion sequence of the lumbar spine therefore requires less radiation than a single traditional radiograph [14]. Mean normalised RMS sEMG during the flexion cycle ranged between 3% and 21% for the TES, 2% and 31% for the LES and 13% and 40% for the LMU.

3.1.1. Reliability and Agreement

Intra- and inter-session reliability and agreement for normalised muscle activity during the bending sequence was high for all muscle levels (Table 2). The highest ICC was for LMU intra-session (ICC = 0.990, 95% CI 0.961–0.998), and the lowest SEM was 0.5% for TES intra-session. The lowest ICC was for LES inter-session (ICC = 0.872, 95% CI 0.508–0.968) and the highest SEM was for LES inter-session (SEM = 3.9%).

Table 2. Intra- and inter-session reliability and agreement for normalised RMS sEMG amplitudes during the weight-bearing sagittal plane QF protocol ($n = 10$).

	Intra-Session ICC (3, 1) (95% CI)	Inter-Session ICC (3, 1) (95% CI)	Intra-Session SEM (%)	Inter-Session SEM (%)
TES	0.996 (0.986–0.999)	0.895 (0.606–0.974)	0.5	2.7
LES	0.984 (0.939–0.996)	0.872 (0.508–0.968)	1.2	3.9
LMU	0.990 (0.961–0.998)	0.974 (0.902–0.993)	1.4	2.8

3.1.2. Correlations between Muscle Activity Changes and IV-RoMmax

A summary of all correlations between changes in muscle activity and IV-RoMmax is given in (Table 3). Significant correlations were only found with lower lumbar segmental motion (L4-5 and L5-S1). These were consistently of mid-level strength (r -values ranging from -0.48 to 0.59), and include inter-vertebral relationships with all three muscle levels. The results also demonstrate a number of correlations that approach significance; these did include relationships with motion at upper inter-vertebral lumbar levels (L2-3 and L3-4).

All significant correlations were further analysed using simple linear regression. The effects of muscle activity changes on IV-RoMmax are shown in (Table 4). The table shows that r^2 values range from 0.177 to 0.247.

Table 3. Correlations* between muscle activity changes (three groups, five epochs) and IV-RoMmax at all inter-vertebral levels (n = 18).

		Inter-Vertebral level			
Muscle activity change		L2-L3	L3-L4	L4-L5	L5-S1
YES epoch 1-2	r	0.404	0.316	-0.164	0.224
	p	0.097	0.201	0.516	0.371
YES epoch 2-3	r	0.083	-0.02	0.036	-0.477
	p	0.743	0.938	0.888	0.045
YES epoch 3-4*	r	-0.059	-0.077	-0.171	-0.434
	p	0.817	0.760	0.496	0.072
YES epoch 4-5	r	-0.124	-0.194	-0.134	-0.103
	p	0.625	0.441	0.596	0.683
LES epoch 1-2*	r	-0.203	0.070	0.595	0.391
	p	0.418	0.782	0.009	0.108
LES epoch 2-3	r	-0.045	0.257	0.295	0.497
	p	0.86	0.303	0.234	0.036
LES epoch 3-4	r	-0.117	-0.118	0.211	0.266
	p	0.645	0.642	0.4	0.286
LES epoch 4-5*	r	0.228	0.215	-0.088	-0.055
	p	0.362	0.392	0.729	0.829
LMU epoch 1-2	r	0.14	0.334	0.314	-0.144
	p	0.58	0.176	0.204	0.567
LMU epoch 2-3*	r	0.021	0.062	0.317	0.139
	p	0.935	0.807	0.200	0.581
LMU epoch 3-4	r	-0.039	0.164	0.455	0.273
	p	0.877	0.517	0.058	0.272
LMU epoch 4-5	r	-0.159	0.067	0.429	0.461
	p	0.53	0.793	0.076	0.027

Significant correlations are highlighted in bold italic. Correlations that approach significance are highlighted in bold. * Indicates a row that includes non-parametric data and therefore a Spearman's Rank Correlation was used. All other normally distributed data were analysed using the Pearson product-moment correlation coefficient. r = correlation co-efficient, p = p-value.

Table 4. Simple linear regression analysis: significant correlations.

Variable	Inter-Vertebral Level	r	p	r ²
LMU Epoch 4-5	L5-S1	0.461	0.027	0.212
LES Epoch 2-3	L5-S1	0.497	0.036	0.247
YES Epoch 2-3	L5-S1	-0.477	0.045	0.227
LES Epoch 1-2*	L4-5	0.595	0.009	0.177

* Indicates a row that includes non-parametric data and therefore a Spearman's Rank Correlation was used. All other normally distributed data was analysed using the Pearson product-moment correlation coefficient r = correlation co-efficient, p = p-value and r² = the co-efficient of determination.

3.1.3. Correlations between sEMG Ratios and IV-RoMmax

The correlations between sEMG ratios and IV-RoMmax at all inter-vertebral levels are shown in (Table 5). The only significant relationship was found between the ratio of LES/YES and the IV-RoMmax at L4-5, and is demonstrated by the scatter plot in (Figure 5). This plot highlights the negative correlation between the LES/YES ratio and L4-L5 IV-RoMmax, and shows that when the muscle activity of the LES increases relative to that of the YES, there is a decrease in the IV-RoMmax at L4-L5. The only other correlation to approach significance was between LMU/LES ratio and the IV-RoMmax at L5-S1 (r = 0.37, p = 0.13).

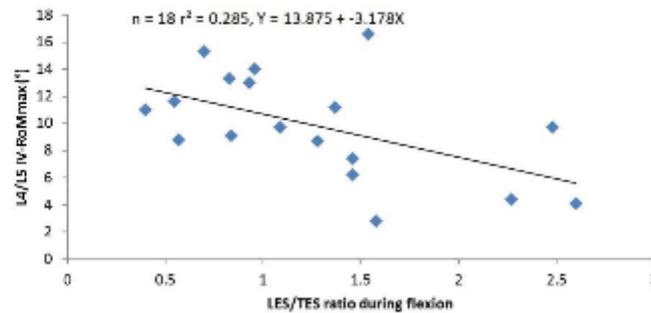


Figure 5. The relationship between the ratio of LES/YES and the IV-RoMmax at L4-5.

Table 5. Correlations between muscle activity ratios and IV-RoMmax at all inter-vertebral levels (n = 18).

Ratio		Inter-Vertebral Level			
		L2-L3	L3-L4	L4-L5	L5-S1
LMU/YES	r	0.046	-0.013	-0.236	0.152
	p	0.856	0.958	0.345	0.548
LMU/LES	r	-0.209	0.04	0.263	0.37
	p	0.405	0.875	0.292	0.13
LES/YES	r	0.095	-0.217	-0.533	-0.242
	p	0.708	0.387	0.023	0.333

r = the Pearson product-moment correlation coefficient, p = p-value.

3.2. Discussion

3.2.1. Reliability and Agreement

It is recommended that any procedures to be used in EMG studies should undergo reliability testing [43]. In this study, intra- and inter-session reliability and agreement was “substantial” for all muscle levels [40]. A common problem with sEMG studies is the great variability in their findings [44,45], therefore the high reliability shown in this study is reassuring. It is usual for a proportion of variability to be attributed to a lack of standardisation, and the method by which EMG variables are normalised [46]. The results however (Table 2) indicate that the standardisation of movement range, speed, and direction provided by the QF protocol may have played an important role in reducing the impact of variability resulting from these causes. It should be observed however that reliability and agreement was relatively poorer in the inter-session group, and of particular note was the increase in SEM for LES (3.9%). As muscle activity changes can be subtle during functional tasks, this may be a limitation for future inter-session studies.

3.2.2. Changes in sEMG Amplitudes at Different Stages of the Flexion Cycle

The results demonstrate that changes in activity of TES, LES, and LMU at various stages of the forward bending cycle, can all be to some degree related to the IV-RoMmax at lower lumbar levels (L4-5 and L5-S1). It has been suggested that intersegmental forces maintain or decrease inter-vertebral motions [47,48], it would seem logical then that if the role of the posterior muscles is to resist sagittal flexion, in order for inter-vertebral movement to occur, there must be a deactivation of this supporting musculature. Figure 6 shows an example of how the muscles most local to the L5-S1 inter-vertebral segment (LMU) demonstrate a significant decrease in activity during the final stage of flexion in a healthy control subject. This corresponds with the phase lag [49] in the initiation of movement at the adjacent inter-vertebral level from the motion graphs. The larger the change in activity between

epochs, (in this case deactivation in the final stages of the flexion cycle) the larger the IV-RoMmax at L5-S1. This is suggestive of a degree of localised control, however, the stabilisation of the pelvis in order to keep the spine in the image frame and avoid hip joint contributions to motion cannot be ruled out as possible external influences. This direct relationship between corresponding levels was not apparent between the LES and the upper inter-vertebral lumbar motion segments (Table 5), and may be suggestive of anatomically specific control at this level. However, the potential importance of LES and TES was also highlighted.

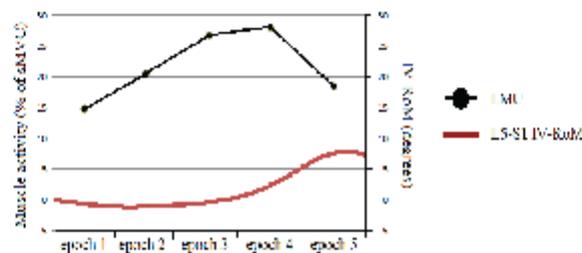


Figure 6. An example of LMU activity and lumbar IV-RoM during sagittal flexion.

Of particular interest is the apparent shift in effect between TES and LES on the IV-RoMmax of L5-S1 (Figure 7). As LES activity decreases between epochs 2 and 3 of the cycle (early mid stage) there is an associated increase in L5-S1 IV-RoMmax, whilst at the same stage of the cycle TES changes (decrease) are significantly associated with a decrease in L5-S1 IV-RoMmax (Figure 8). This indicates possible different roles for TES and LES in terms of the control of the range of motion at a distal motion segment. If there is more movement at L5-S1 there may be less activity of LES, more TES, and vice versa.

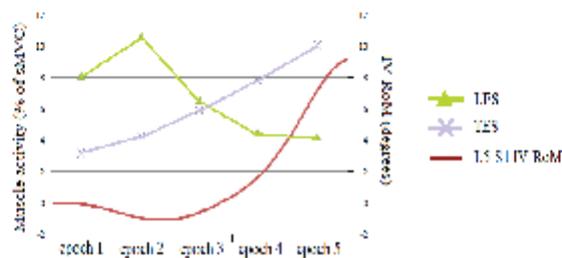


Figure 7. An example of LES and TES activity and L5-S1 IV-RoM during sagittal flexion (An example of a greater IV-RoMmax). Please note that the scales of both Y-axis are slightly different to those seen in Figure 8.

When considering the LES to be local (inter-segmental) and TES to be global (multi-segmental) [50], then these findings may have important clinical implications, as they raise the possibility of level specific stabilisation/control. Conflicting arguments have been put forward regarding the role of local and global muscles in spinal stability, Bergmark suggested that inter-segmental (local) muscles were the chief stabilisers [50], whereas Crisco and Panjabi concluded that the larger multi-segmental (global) muscles were more powerful [51]. In a study investigating the relative contribution of different trunk muscles to lumbar stability, Cholewicki and Van Vliet concluded that whilst inter-segmental and multi-segmental paraspinals had the greatest effect on stabilisation compared to other muscles (psoas and rectus abdominis), no distinction could be made between the two [52].

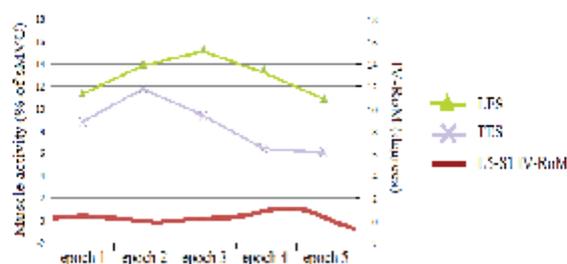


Figure 8. An example of LES and TES activity and L5-S1 IV-RoM during sagittal flexion (An example of a smaller IV-RoMmax). Please note that the scales of both Y-axis are slightly different to those seen in Figure 7.

There are many correlations that approach significance (Table 3), and therefore future studies with a larger sample size may well reveal more statistically important relationships, potentially with upper lumbar inter-vertebral levels.

3.2.3. The sEMG Ratios

Previous work has indicated a clear distinction between the kinematic behaviour of the upper and lower sections of the lumbar spine [53]. It was anticipated therefore that there may be relationships between the IV-RoMmax and the muscle activity ratio of LMU/LES. These were not evident, and suggest that the location of a motion segment within the spinal curvature, or the influence of passive structures (e.g., the strong iliolumbar ligament) may influence such interactions. The ratio of LES/TES however, did reveal a statistically significant negative relationship with the range of motion at L4-L5 (Figure 5 and Table 5).

The ratio of lumbar erector spinae over thoracic erector spinae activity has been investigated in several previous studies [38,39,54–56]. In a musculoskeletal trunk model based on the EMG data collected from two healthy participants, Cholewicki and McGill suggested that the preferential recruitment of the LES over the TES may be a strategy to increase spinal stiffness [54]. A further study comparing the muscle recruitment patterns in healthy controls to those of LBP patients, found higher LES/TES ratios in the latter [38]. These results led to the conclusion that the differences found between groups were likely to be an adaptation designed to enhance spinal stability. This theory was further supported by Van Den Hoorn *et al.* (2012), who also demonstrated a significantly higher LES/TES ratio in LBP patients during gait [55].

Reeves *et al.* also investigated this muscle activation imbalance in varsity athletes, and while maintaining that there was indeed a relationship between muscle imbalance between levels and LBP, the authors also found that in some individuals with a history of LBP, TES activity could be dominant [39]. The authors contend that this may be explained by pathology, e.g., the CNS optimising activation to minimise compression, or by a difference in muscle fibre types between groups in order to compensate for fatigue related pain [39]. Crucially however, there is also the mention of the possibility of the patterns being the result of different types of posture or lordosis, and that further studies may account for this effect.

The results of this study highlight that the ratio of LES/TES can vary in a population with no long term history of low back pain, and would appear to relate to variations in inter-vertebral mechanics in such a population. It has been proposed that lumbar inter-segmental movement is also influenced by spinal morphology [57], but these results provide more level-specific detailed information, and it is apparent that different recruitment strategies are required in accordance with inter-vertebral range changes. A question frequently asked in this field of research is whether these strategies are a cause or a consequence of the related kinematics.

It has been suggested that muscle imbalance between levels does not cause low back injury [39]. It is also suggested that imbalance is not necessarily tantamount to impairment. Therefore correcting muscle imbalance in patients should not be a priority. However if L4-L5 or L5-S1 are the segments of interest, or suspected levels of pain generation and movement at that level is considered to be part of the problem, then reducing the muscle imbalance may be of importance.

In a LBP free population sample, it might be assumed that variations in muscle activity patterns do not represent adaptations to pain. However, that is not to say that particular activity patterns and thus kinematic behaviours may not be risk factors for future LBP episodes. It also questions the conclusions of studies that compare LBP population groups with healthy pain free controls, as muscle activity patterns may not be adaptations to the episode.

It is suggested that achieving sufficient stability is a moving target, and that no single muscle can therefore be considered the best stabiliser, as the most important muscle is transient dependent on the task [58]. The results provide a demonstration of this concept in action during the task of forward bending. Whilst effect sizes are small, inter-vertebral movements have been shown here to be influenced by muscle activity. It would seem that IV-RoMmax depends not only on the relative activation of multiple trunk muscles, but also other biomechanical variables, therefore, the next logical step may be to assess the importance of each. This will require multivariate analysis of larger population samples. If the relative value of each factor can be determined, then better informed decisions regarding model types and inputs may be possible. The diversity of muscle activation patterns within a "normal" sample highlights the problem of using limited participant numbers as a basis for systems models, whereas reductionist approaches are typically weakened by the limitations of the size of the effects of the selected variables. If the variables with the greatest influence on kinematics can be found, then the selective use of these variables in models and LBP/control studies may be beneficial.

Finally, it is a limitation of this study that people with non-specific low back pain were not included, yet it would be important to know to what extent these relationships, which are consistent with maintaining appropriate restraint on vertebrae during bending, are disrupted in patients. If so, this would point to a potential route for patient stratification based on biomechanics. Such studies are now warranted. The study also only investigates a narrow population (*i.e.*, young healthy male adults) and so the results are not generalizable to other groups. It is anticipated that variations in kinematic and morphological parameters that are associated with age related change and gender would also affect IV-RoMmax, and therefore also warrant further investigation. Future investigators may also wish to incorporate measurements such as thoracic kyphosis and pelvic incidence in order to gain insight into changes in kinematic behaviour beyond the lumbar spine.

4. Conclusions

This study found weak to moderate but significant correlations between both muscle activity changes and ratios and IV-RoMmax at various inter-vertebral levels. Of particular interest was the correlation between decreased LMU and increased IV-RoMmax at L5-S1 in the latter stages of flexion, the apparent co-dependency between LES and TES during early to mid-flexion, and the effect of the LES/ TES ratio on the IV-RoMmax at L4-L5. These relationships, when combined with other influencing factors, may be important when specific inter-vertebral levels are considered to be sources of pain generation and when considering rehabilitative or surgical planning. Multivariate investigations in larger samples are warranted, potentially leading to longitudinal outcome studies in LBP groups.

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Author Contributions: Alister du Rose served as project lead, performed the statistical analysis, operated the sEMG equipment, conducted the image analysis and wrote the first draft. Alan Breen operated the motion table. All authors contributed to the study design, procedural protocols, data acquisition, and the drafting of this paper. All authors read and approved the final manuscript.

Conflicts of Interest: The authors declare no conflict of interest.

Abbreviations

The following abbreviations are used in this manuscript:

AB	Alan Breen
ADR	Alister du Rose
AECC	Anglo-European College of Chiropractic
BMI	Body Mass Index
CI	Confidence Interval
CMRR	Common Mode Rejection Ratio
CNS	Central Nervous System
EAC	European Academy of Chiropractic
EMG	Electromyography
FRP	Flexion relaxation phenomenon
GP	General Practitioner
IAR	Instant axis of rotation
ICC	Intraclass Correlation Coefficient
IV-RoM	Inter-vertebral Range of Motion
IV-RoMmax	Maximum Inter-vertebral Range of Motion
LBP	Low Back Pain
LES	Lumbar erector spinae
LMU	Lumbar multifidus
NRES	National Research Ethics Service
QF	Quantitative Fluoroscopy
RMS	Root Mean Square
RoM	Range of Motion
SD	Standard Deviation
SEM	Standard error of measurement
TES	Thoracic erector spinae

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Appendix S: Systematic review reporting checklist (Moher et al. 2009)

Checklist of items to include when reporting a systematic review or meta-analysis (Moher et al. 2009).

Section/topic	Item No	Checklist item
Title	1	Identify the report as a systematic review, meta-analysis, or both
Abstract Structured summary	2	Provide a structured summary including, as applicable, background, objectives, data sources, study eligibility criteria, participants, interventions, study appraisal and synthesis methods, results, limitations, conclusions and implications of key findings, systematic review registration number
Introduction Rationale	3	Describe the rationale for the review in the context of what is already known
Objectives	4	Provide an explicit statement of questions being addressed with reference to participants, interventions, comparisons, outcomes, and study design (PICOS)
Methods Protocol and registration	5	Indicate if a review protocol exists, if and where it can be accessed (such as web address), and, if available, provide registration information including registration number
Eligibility criteria	6	Specify study characteristics (such as PICOS, length of follow-up) and report characteristics (such as years considered, language, publication status) used as criteria for eligibility, giving rationale
Information sources	7	Describe all information sources (such as databases with dates of coverage, contact with study authors to identify additional studies) in the search and date last searched
Search	8	Present full electronic search strategy for at least one database, including any limits used, such that it could be repeated
Study selection	9	State the process for selecting studies (that is, screening, eligibility, included in systematic review, and, if applicable, included in the meta-analysis)
Data collection process	10	Describe method of data extraction from reports (such as piloted forms, independently, in duplicate) and any processes for obtaining and confirming data from investigators
Data items	11	List and define all variables for which data were sought (such as PICOS, funding sources) and any assumptions and simplifications made
Risk of bias in individual studies	12	Describe methods used for assessing risk of bias of individual studies (including specification of whether this was done at the study or outcome level), and how this information is to be used in any data synthesis
Summary measures	13	State the principal summary measures (such as risk ratio, difference in means).
Synthesis of results	14	Describe the methods of handling data and combining results of studies, if done, including measures of consistency (such as I^2 statistic) for each meta-analysis
Risk of bias across studies	15	Specify any assessment of risk of bias that may affect the cumulative evidence (such as publication bias, selective reporting within studies)
Additional analyses	16	Describe methods of additional analyses (such as sensitivity or subgroup analyses, meta-regression), if done, indicating which were pre-specified
Results Study selection	17	Give numbers of studies screened, assessed for eligibility, and included in the review,

		with reasons for exclusions at each stage, ideally with a flow diagram
Study characteristics	18	For each study, present characteristics for which data were extracted (such as study size, PICOS, follow-up period) and provide the citations
Risk of bias within studies	19	Present data on risk of bias of each study and, if available, any outcome-level assessment (see item 12).
Results of individual studies	20	For all outcomes considered (benefits or harms), present for each study (a) simple summary data for each intervention group and (b) effect estimates and confidence intervals, ideally with a forest plot
Synthesis of results	21	Present results of each meta-analysis done, including confidence intervals and measures of consistency
Risk of bias across studies	22	Present results of any assessment of risk of bias across studies (see item 15)
Additional analysis	23	Give results of additional analyses, if done (such as sensitivity or subgroup analyses, meta-regression) (see item 16)
Discussion		
Summary of evidence	24	Summarise the main findings including the strength of evidence for each main outcome; consider their relevance to key groups (such as health care providers, users, and policy makers)
Limitations	25	Discuss limitations at study and outcome level (such as risk of bias), and at review level (such as incomplete retrieval of identified research, reporting bias)
Conclusions	26	Provide a general interpretation of the results in the context of other evidence, and implications for future research
Funding		
Funding	27	Describe sources of funding for the systematic review and other support (such as supply of data) and role of funders for the systematic review

Appendix T: Justification of sample size for reliability studies

It has been suggested that the use of a relevant (i.e. an appropriately selected) ICC is suitable for determining the reliability of measurements (Shrout and Fleiss 1979), and whilst there is little reference in the literature regarding planning for reliability study sample size, Donner and Eliasziw (1987) suggested that “for 2 sets of observations where reliability is based on the intra-class correlation from one-way analysis of variance, 40 paired observations will provide 80% power at the 5% level of significance for the ICC result” (Donner and Eliasziw 1987). Donner and Eliasziw concede that this number is fairly conservative however, and suggest that true sample size may be less when a two-way model of analysis is adopted.

Determining the sample size for the reliability testing was therefore an important part of the design of this study. Unfortunately the recording and processing of QF kinematic and EMG measurements is costly both in terms of finances and time, and so the planned sample size was required to be large enough extract meaningful information, whilst avoiding preventable costs by being larger than was necessary. Many previous kinematic measurement and EMG variable repeatability studies have used smaller sample sizes, i.e. $n = 5$ (Lee et al. 2002; Mannion and Dolan 1994), $n = 8$ (Stokes et al. 1987), $n = 10$ (McGregor et al. 1995; Thuresson et al. 2005; Mellor et al. 2014; Branney and Breen 2014), $n = 11$ (Dankaerts et al. 2004; Dorel et al. 2008), $n = 15$ (Daneels et al. 2001); $n = 20$ (Frobin 1996; Mannion et al. 2004), although there are of course examples of larger samples e.g. $n = 70$ (Ahern et al. 1986). Whether or not the results of such studies are meaningful depends not just on the size of the sample but how the results are produced and reported, and the width of the data’s confidence intervals (CI’s). The CI of an ICC can be taken to represent the range within which it can be certain that the true effect lies. The width of the confidence interval (CI) is affected by the sample size, with larger studies typically providing more precise estimates of effects and therefore narrower CI’s than smaller studies.

A larger sample size does not necessarily equate to a narrow CI if the measurement has poor reliability, but a larger sample would be justified if the CI is especially wide. In a study investigating the intra-examiner between day, and inter-examiner reliability of segmental ranges of flexion using a skin surface measurement device called a spinal mouse, Mannion et al. (2004) showed moderate mean ICC’s of 0.64 and 0.62 respectively. The CI’s were however shown to be as wide as 0.02-0.75 in the lumbar spine, and so the device was shown not to be reliable for lumbar inter-vertebral flexion measurements (Mannion et al. 2004), a

conclusion unlikely to be changed by an increase in sample size. In a similar sized study (i.e. $n = 20$) Frobin et al. 1996 investigated the inter- and intra-observer reliability of inter-vertebral angular ranges marked on lateral lumbar x-rays, stating excellent reproducibility for both (Frobin et al. 1996). The Frobin et al. study however did not report either ICC statistics or CI's and therefore the validity of their conclusions is questionable. This is a common problem in older studies. Ahern et al. 1986 for example investigated the within session reliability of mean EMG amplitude recordings during weight-bearing flexion and return using the Pearson r correlation statistic and not an ICC. Whilst demonstrating excellent within session correlations for both left and right side paraspinal muscle amplitudes ($r = 0.97$ bilaterally), the Pearson r value provides no insight into systematic errors within the measurement, and therefore even a perfect correlation should not be misinterpreted as complete agreement (Bland and Altman 1999; Vaz et al. 2013). In such as case it could be argued that a smaller but better designed study would have been more meaningful.

Even when repeatability studies do incorporate the use of an ICC statistic, the CI's are sometimes not reported (Daneels et al. 2001; McGregor et al. 1995) and are therefore limited in that the range in which the true effect lies is not presented. More recent inter- and intra- observer studies have used the QF technology to determine the repeatability of recumbent lumbar and weight-bearing cervical inter-vertebral ranges of motion during sagittal flexion (Branney and Breen 2014; Mellor et al. 2014). Both investigations included the ICC statistic, the CI's, and the standard error of measurement (SEM), and despite a relatively small sample size ($n = 10$), showed narrow confidence intervals, with the widest range being (0.82-0.99) and (0.68-0.98) respectively. As these confidence intervals are already acceptably narrow, a larger sample size would unlikely make any meaningful difference in the findings. It is argued therefore, that a sample size of $n = 10$ is appropriate for repeatability studies using QF measurement technologies. For consistency the EMG reliability and agreement studies also used a sample of ten.

Glossary

Adjacent segment disease (ASD)	The symptomatic deterioration of vertebral levels adjacent to the site of a previous spinal fusion
Anglo-European College of Chiropractic (AECC)	The collaborating institution
Alister du Rose (ADR)	The author
Analysis of variance (ANOVA)	A statistical method used to test differences between two or more means
Biomed central (BMC)	An online publisher of free peer-reviewed scientific articles
Body mass index (BMI)	$BMI = \text{mass (kg)}/\text{height (m)}^2$
Bournemouth University (BU)	The host institution
Confidence interval (CI)	For a given statistic calculated for a mean, the CI is the range of values around that statistic that are believed to contain within a certain probability, the true value of that statistic
Coefficient of multiple correlations (CMC)	A measure of how well a specified variable can be predicted using a linear function of a set of other variables
Common mode rejection ratio (CMRR)	The measure of rejection by the EMG hardware of undesirable input signals
Central nervous system (CNS)	The part of the nervous system that includes the brain and spinal cord
Chronic non-specific low back pain (CNSLBP)	LBP of no known biological or pathological origin (for longer than 6 weeks)
Coronal plane	A vertical plane that divides the body into ventral and dorsal sections (Appendix B)
Clinical prediction rules (CPRs)	Tools intended to guide clinicians in terms of their decision making
Combined proportional range variance (CPRV)	The combination of the variance of the proportional ranges throughout motion sequences in all measured planes
Computed tomography (CT)	An imaging technique that uses x-ray equipment to generate pictures of internal body structures
Coefficient of variation (CV)	The ratio of the standard deviation to the mean
Diplomate of the American Chiropractic Board of Radiology (DACBR)	Diplomate of the American Chiropractic Board of Radiology
Dual fluoroscopy imaging system (DFIS)	Dual fluoroscopy imaging system

European Academy of Chiropractic (EAC)	The academic division of the European Chiropractors Union (ECU)
Electrocardiography (ECG)	The recording of electrical activity from the heart using electrodes
Eccentric: concentric ratio (ECR)	The ratio of eccentric muscle activity over the concentric muscle activity
Electromyography (EMG)	The study of the function of muscles via the electrical signal associated with muscular contraction
Erector spinae (ES)	Three columns of paraspinal muscles (Iliocostalis, longissimus and spinalis) travelling from the skull to the pelvis
Food and Drug Administration (FDA)	The U.S. Food and Drug Administration
Flexion relaxation phenomenon (FRP)	The phenomenon of reduced paraspinal myoelectrical activity during sagittal flexion of the trunk
Flexion relaxation ratio (FRR)	The ratio of maximal muscle activity during flexion and activity at full flexion
Functional spinal unit (FSU)	The smallest physiological motion unit of the spine to exhibit biomechanical characteristics similar to those of the entire spine
General practitioner (GP)	A medical doctor (U.K)
Graphical user interface (GUI)	Software that works at the interface between the user and a computer, employing graphical elements
International Business Machines (IBM)	An information technology company
Instantaneous axis of rotation (IAR)	The centre about which spinal muscles employ their moment during flexion, extension and rotation
Intraclass correlation coefficient (ICC)	A statistic that assesses the consistency between measures of the same class
Iliac crest level (ICL)	Situated at the same level as the iliac crest
Institute for Musculoskeletal Research and Clinical Implementation (IMRCI)	A research institution within the AECC
The International Society of Electrophysiology and Kinesiology (ISEK)	An organisation devoted to the study of human movement and the neuromuscular system
Inter-vertebral flexion extension (IVFE)	Inter-vertebral flexion extension
Inter-vertebral range of motion (IV-ROm)	Inter-vertebral range of motion
Maximum inter-vertebral range of motion (IV-ROmmax)	Maximum inter-vertebral range of motion
Innervation zone (IZ)	A site within a muscle where nerve terminations and muscle fibres are connected

Kilovolt (kV)	A radiation factor relating to the potential difference between the cathode and anode
Low back pain (LBP)	Pain anywhere between the 12 th rib and the crease of the buttocks
Lumbar erector spinae (LES)	The lumbar section of the ES
Lumbar multifidus (LMU)	Deep segmentally acting back muscles
Milliampere (mA)	A radiation factor relating to the quantity of electrons which pass from the cathode to the anode
Movement system impairment (MSI)	A categorisation technique for low back pain groups
Magnetic resonance imaging (MRI)	A technique employing a magnetic field and radio waves in order to produce images of internal organs and tissues
Maximum voluntary bending angle (MVBA)	Maximum voluntary bending angle
Maximal voluntary contraction (MVC)	The maximum contraction a subject can produce during a given exercise
National Health Service (NHS)	A publically funded national healthcare system (U.K.)
National research ethics service (NRES)	A national ethics approval service, now part of the Health Research Authority
Non-specific low back pain (NSLBP)	LBP of no known biological or pathological origin
Objective spinal motion imaging assessment (OSMIA)	A fluoroscopy imaging technique devised by the IMRCI
Post graduate researcher (PGR)	Post graduate researcher
Patient and public involvement (PPI)	The involvement of patients and the public in various stages of the design and conduct of clinical research
Posterior superior iliac spine (PSIS)	A bony prominence of the posterior ilium serving as an anatomical reference point
Quantitative fluoroscopy (QF)	An imaging technique used to assess continuous inter-vertebral motion
Randomised controlled trial (RCT)	A study where participants are randomly allocated to one of several interventions
Recumbent (REC)	Lying down on the participant's right hand side
Root mean square (RMS)	Reflects the mean power of the EMG signal
Range of motion (RoM)	Range of motion
Reference voluntary contraction (RVC)	A muscular contraction that provides a reference value from which to normalise measurement data

Sagittal plane	A vertical plane which passes from ventral to dorsal dividing the body into left and right halves (Appendix B)
Standard deviation (SD)	A statistic used to estimate the mean variability in a set of data. It is calculated as the square root of the mean
Standard error of measurement (SEM)	A statistic that estimates how repeated measures of an individual using the same instrument are distributed around their true score
Surface electromyography (sEMG)	The study of the function of muscles via the electrical signal associated with muscular contraction, using surface electrodes
Surface Electromyography for the Non-invasive Assessment of Muscles (SENIAM)	A project that is part of the Biomedical Health and Research Program (BIOMED II) of the European Union
Sub-maximal voluntary contraction (sMVC)	A sub-maximal contraction produced during a given exercise
Signal to noise ratio (SNR)	A means of comparing the level of desired signal to the level of background noise
Statistical Package for the Social Sciences (SPSS)	A statistical analysis software package
Soft tissue thickness (STT)	The thickness of subcutaneous tissue between the electrode and the contracting muscle
Thoracic erector spinae (TES)	The thoracic section of the ES
Thoracolumbar fascia (TLF)	A mixture of aponeurotic and fascial planes that form the retinaculum around the lumbar paraspinals, and muscles including transversus abdominus insert into it
Ultrasound (US)	A technique that uses ultrasonic waves to produce images of internal body structures
Weight-bearing (WB)	The participant is in a standing position