An investigation into the spinal kinematics and lower limb impacts during cricket fast bowling and their association with lower back pain.

by

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Abstract

Cricket fast bowlers have been highlighted as having increased risk of injury when compared to the rest of the team. Lower back injury and more specifically, pain results in more time away from cricket than any other injury in the fast bowling population, with juniors displaying even greater risk compared with senior fast bowlers. Whilst lower back injury (confirmed musculoskeletal diagnoses, usually radiographically) in fast bowlers has been repeatedly investigated. Lower back pain (LBP), defined as pain resulting in time away from matchplay or training with or without a formal diagnosis, (highlighted to display a different relationship to injury) has received little attention in fast bowling literature. High bowling workloads (usually recorded in overs or days bowled) and the immature spine of junior fast bowlers have been highlighted as significantly increasing risk of injury. However, research regarding specific kinematic and kinetic risk factors requires further attention. Therefore, this study aimed to address current methodological limitations to investigate the association between spinal kinematics and lower limb impacts during fast bowling and risk of LBP in junior and senior fast bowlers.

This study compares bowling kinematics and lower limb impacts in junior and senior fast bowlers and retrospective and prospective LBP risk, to provide additional insight into the clinical biomechanics of fast bowling. This study has shown inertial sensors and accelerometers are a valid (r>0.8 for 79% of variables, RMSEP = 0.3-1.5°) and reliable (ICC's >0.8 and SEM<3.4g and 9°) method of analysing fast bowling lower limb impacts and spinal kinematics and may therefore be an acceptable alternative to current methodologies. Analysis of tibial impacts on different playing surfaces displayed larger impacts on outdoor artificial surfaces (26.6g) compared with grass (24.7g) and indoor rubber (22.0g) and wood (17.8g). Highlighting, large workloads on outdoor artificial surfaces may increase injury risk, with a wooden indoor surface more favourable.

Retrospective and prospective LBP and injury data highlighted that senior fast bowlers with known spinal pathologies displayed four times greater risk of future LBP. However, this was not necessarily the case in junior bowlers. Results highlighted that peak accelerations at back-foot impact were higher in bowlers with no history of LBP, as well as bowlers that did not develop LBP in the follow-up season with differences

between 8-10g seen in peak tibial acceleration. This may be a potential mechanism of reducing load at front-foot impact (which showed few notable differences between groups). Junior bowlers with a history of LBP displayed less contralateral thoracic rotation at back-foot impact and consequently a lower overall range. However, this trend was not displayed in senior bowlers. Senior bowlers, with either a history of LBP or that went on to develop LBP bowled with almost double lumbar extension (9° to 16°) resulting in a 12° increase in thoracolumbar extension at back-foot impact. Therefore, this study suggests that higher magnitudes of fast bowling impacts may not be synonymous with increased risk of LBP, however spinal kinematics at back-foot impact may provide some insight into bowlers' risk of developing LBP.

The effect of these recommendations on fast bowling performance was analysed through a correlation of impact and spinal kinematics with ball release speed. This highlighted that the recommendations to reduced risk of LBP are not likely to affect ball release speed, as only sacral loading rate at back foot impact and thoracic lateral flexion at FFI showed significant correlations with ball release speed (r=.521 and .629 respectively). Overall this study has demonstrated the application of novel technology applied to the live cricket fast bowling situation, overcoming limitations of previous methods. The method was valid, reliable and sensitive enough to determine significant differences in the spinal kinematics which were associated with LBP history or with developing LBP in the follow-up season and these were specific to junior and senior bowlers. These new insights will help to inform surveillance and coaching practices in the quest to reduce the injurious nature of fast bowling.

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Publications

Senington, B., Lee, R. Y. and Williams, J. M., 2018. Are shoulder counter rotation and hip shoulder separation angle representative metrics of three-dimensional spinal kinematics in cricket fast bowling?. *Journal of Sports Sciences*, 36(15), 1763-1767.

Senington, B., Lee, R. Y. and Williams, J. M., 2018. Ground reaction force, spinal kinematics and their relationship to lower back pain and injury in cricket fast bowling: A review. *Journal of Back and Musculoskeletal Rehabilitation*, (Preprint), 1-14.

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Chapter 1

Introduction

1.1 Statement of the problem

Within the cricketing population, fast bowlers have been reported to possess a 10% higher risk of developing a musculoskeletal injury than the rest of the team (Orchard et al. 2002). Previous studies have highlighted that the lower back is the most common site of injury and is an even greater issue among adolescent fast bowlers with prevalence of lumbar stress fractures reported between 24-81%, much higher than the 3-6% prevalence reported in the general population (Elliott et al. 1992; Standaert and Herring, 2000; Ranson et al. 2005; Engstrom and Walker, 2007; Crewe et al. 2012). A senior medical expert at the England and Wales Cricket Board (ECB) stated:

> "Lumbar spine injury in fast bowlers in cricket is a very common and potentially career threatening issue. This can result in significant and repeated periods of time lost from the game. Lumbar injury can strike down a fast bowler at any time but their formative and maturation periods during their late teens and early 20's are particularly critical. The more knowledge we can gain about the causative factors can only benefit the players and sport alike" (personal communication, ECB medical expert, 28 June 2014).

Whilst pathological studies are able to report prevalence of these injuries, the impact of these injuries may go far beyond the initial injury. With the ECB central contract for 2016 showing a retainer fee of £700,000 per player and an average county player earning an average of £50,000 a year, treatment costs and ramifications of time away from the game can become a big financial strain for clubs and players alike (Financial Times, 2014; ECB, 2016). Whilst lower back injuries are some of the most prevalent injuries in the game, due to its comparatively long recovery time, they also account for at least double the amount of cricket missed compared to any other injury (Orchard et al. 2002).

Previous research has hypothesised that excessive bowling workloads (usually defined as overs or days bowled), large magnitudes of force during bowling and weaknesses in key anatomical structures (such as the pars interarticularis) all contribute towards fast bowlers' increased risk of lower back injury and pain (Dennis et al. 2004). However, the relationship between three-dimensional bowling kinematics, kinetics and increased risk of these pathologies remains unclear.

Although previous studies have investigated the relationship between fast bowling kinematics and risk of lower back injury and pain, few variables have been conclusively linked to an increased risk of lower back injury and pain (Morton et al. 2013). Research has highlighted shoulder counter-rotation (SCR) as being strongly linked to an increased risk of lower back injury (Portus et al. 2004; Stuelcken et al. 2010). Significantly higher prevalence of lower back injury has been reported among bowlers displaying > 30° SCR (Portus et al. 2004; Stuelcken et al. 2010). However, this measure fails to describe three-dimensional spinal kinematics, therefore, the pathomechanics of SCR and lower back injury is still unclear. Excessive lateral flexion has also been hypothesised to increase risk of lower back injury and pain, however only one study was able to report a significant relationship (Glazier, 2010; Bayne et al. 2016). This may be due to the heterogeneity of methodologies used making it difficult to make comparisons between published results. Therefore, it may be beneficial to create a standardised method of measuring and reporting spinal kinematics.

Fast bowling kinetics have been more consistently reported than spinal kinematics, as studies have all used force plates to measure ground reaction forces (GRF) at the last stride before ball release (delivery stride) (Portus et al. 2004; Crewe et al. 2013; Worthington et al. 2013). Consequently, GRF during fast bowling is better understood and it has been hypothesised that high GRF contributes towards an increased risk of lower back pain and injury (Crewe et al. 2013). However, few studies have looked at the effect of fast bowling on forces further up the body, therefore the relationship between fast bowling kinetics and lower back injury and pain remains unclear (Crewe et al. 2013; Bayne et al. 2016).

In order to gain a better understanding of the mechanisms of injury regarding spinal kinematics and kinetics, more work needs to be carried out to consolidate previous findings as well as, investigate apparent gaps in the research. Gaining a clearer picture of the pathomechanics of lower back injury and pain in fast bowling will allow coaches to implement preventative interventions to decrease risk of these injuries and decrease time lost from the game. Improving education of how to decrease risk of these injuries, as well as development of more widely accessible methods of technique screening, may not only prevent the need for potentially costly treatments, but allow a greater number of young players to continue unhindered development to elite levels of cricket and elite cricketers to prolong their professional playing careers. Therefore, this research project

aims to identify specific variables relating to spinal kinematics and delivery stride impact characteristics that may increase the risk of lower back pain in the fast bowling population. Furthermore, this project will aim to develop a valid and reliable method of 'in-field' analysis of these variables to allow the assessment of live bowling.

1.2 Purpose of the study

One purpose of this thesis was to investigate the effect of impacts during bowling on risk of lower back pain in junior and senior fast bowlers. This study assessed impacts during fast bowling using novel inertial sensing technology. Sensors consisting of accelerometers, gyroscopes and magnetometers placed on the tibia and sacrum were utilised to overcome limitations to existing laboratory-based methodologies. Initially, reliability and validity of this technology for the analysis of fast bowling impact characteristics during the delivery stride was explored. The portable nature of this technology enabled the development of methodologies to analyse playing surface properties and their effect on tibial impact characteristics during fast bowling. These sensors were also utilised to investigate the relationship between impacts at the tibia and sacrum and back pain both retrospectively and prospectively in both junior and senior elite level fast bowlers.

Additionally this thesis assesses the effect of three-dimensional spinal kinematics during bowling on risk of lower back pain in junior and senior fast bowlers. Inertial sensors located at the T1, L1 and S1 spinous processes were used to analyse lumbar, thoracic and thoracolumbar spinal kinematics in three dimensions during live fast bowling. Reliability and validity of these methods were also investigated. These methods were further developed to assess spinal range of motion during fast bowling; these values were then used to explore the relationship between shoulder counter rotation (a variable previously reported as increasing risk of lower back injury when excessive ranges are observed) and three-dimensional spinal kinematics. Consequently, this study comments on the suitability of current methods of kinematic assessment in relation to risk of lower back pain and injury, as well as address limitations highlighted in this analysis in previous studies. This thesis further investigates the relationship between spinal range of motion during bowling and back pain retrospectively and prospectively in junior and senior elite level fast bowlers.

Furthermore, this thesis investigates the effects of LBP reduction recommendations produced from the above data on fast bowling ball release speed. This data provides coaches with the knowledge of whether interventions aimed at reducing risk of LBP would also reduce match performance. This will therefore comment on the likely uptake of the suggested interventions.

This thesis presents results of the studies described above. Tibial accelerometry may provide practitioners with easy to understand data comparable to currently reported ground reaction force measures. Whilst sacral accelerations may provide a more representative measure of risk of lower back injury than the more commonly reported ground reaction force measures, which can only describe interaction between the foot and the floor. It is believed that the use of accelerometers may enable coaches to carry out portable and comparatively inexpensive analysis of fast bowling kinetics outside of a laboratory environment. Thus, new recommendations centred on 'in-field' analysis of live fast bowling are able to be produced. The relationship between thoracic and thoracolumbar kinematics in relation to fast bowling injury risk represent novel findings within the current body of literature, and hence, may provide new insight into the relationship between spinal kinematics and injury.

In gaining a greater understanding of how impact characteristics, spinal kinematics and history of back pain correlate with future episodes, coaches may be able to assess the risk of their players contracting further injuries. Therefore, this study has the potential to provide vital knowledge to enable coaches to implement interventions that could lower the risk of sport specific back injuries, without affecting performance. The applications of new technologies for the analysis of fast bowling used in these studies may provide coaches and practitioners with further solutions to assessment and implementation of interventions away from laboratory environments. The application of quantitative assessment methods based on recommendations provided by these studies may prove invaluable for coaches and support staff in regards to injury prevention and rehabilitation.

1.3 Organisation of Thesis

Chapter one provides an introduction to the thesis by highlighting the current problem and need for the study. This chapter also outlines the organisation of the thesis.

Chapter two provides a summary of current literature relating to fast bowling and biomechanical analysis. It highlights key research investigating fast bowling pathologies and the association with biomechanics of fast bowling. It continues by analysing key methodologies used to evaluate kinematic and kinetic variables during fast bowling and other relevant movements.

Chapter three provides detailed descriptions of the development of the methods and all methods utilised in this thesis. This includes details on the participants used, instrumentation, procedures, data and statistical analysis for each study.

Chapter four reports the results and discussion around the supporting studies used to inform the studies used to address the primary focus of this thesis; the investigation of the relationship between fast bowling biomechanics and lower back pain risk. This chapter assesses the reliability and validity of accelerometry and inertial sensors for analysis of impacts and spinal kinematics during fast bowling. Studies in this chapter also investigate how different playing surfaces may affect fast bowling impacts and the relationship between shoulder counter-rotation (a commonly reported risk factor associated with spinal pathology) and three-dimensional spinal kinematics.

Chapter five investigates fast bowling impacts and kinematics and how they may impact on the risk of lower back pain. A comparison between junior and senior bowlers' impact characteristics and spinal kinematics is presented as well as the relationship between fast bowling impacts and kinematics and risk of lower back pain in junior and senior fast bowlers.

Chapter six continues to investigate the impact and spinal kinematic variables explored in chapter five, assessing how they affect fast bowling performance. This chapter explores the relationship between these variables and ball release speed as well as how orientation of the back leg at back foot impact may affect these variables. This chapter aims to highlight whether proposed interventions in chapter five are likely to affect performance.

Chapter seven provides a general discussion on the findings of this thesis.

Chapter eight presents conclusions and recommendations for future work.

Chapter 2
Review of Related
Literature

2.1 Introduction

This chapter reviews current literature relating to analysis of fast bowling and risk of injury and pain. Studies analysing pathology and lower back pain in cricket and within the fast bowling population are reviewed (2.2). In order to suitably address the risks of injury and understand their impact on the game, the current hypotheses for mechanisms of injury and pain must be understood. This chapter continues with a comprehensive analysis of fast bowling kinetics as reported by current research (2.3). This section reviews study results and appraises current methods of fast bowling kinetic data collection. Section 2.4 then reviews literature reporting fast bowling kinematics and critiques the methods of data collection used in these studies. Whilst fast bowling kinetics and kinematics are often analysed as separate questions, it is important to understand the association between the two in relation to injury risk. Thus, section 2.5 amalgamates key findings relating to kinematics, kinetics and risk of lower back injury and pain, highlighting gaps in current literature and recommendations for further work. Section 2.6 summarises this review, outlining the need for this study and the research questions being addressed.

2.2 Pathology and Lower Back Pain in Cricket Fast Bowlers

2.2.1 Introduction

Lower back pain and injury in sport has been highlighted as a major issue (Morton et al. 2013; Mcanany et al. 2016; Riley Jr and Micheli, 2016). This can lead to activity limitations, substantial time out of competitive and training situations and very often pain or discomfort (Johnson et al. 2012). Particularly high prevalence of lower back pain and injury has been reported in sports such as tennis, golf, gymnastics and cricket (Alyas et al. 2007; Kruse and Lemmen, 2009; Glazier, 2010; Cole and Grimshaw, 2014). These sports tend to display similar spinal motions; therefore, research has aimed to identify the pathomechanics of these injuries (Cholewicki and McGill, 1996; Reeves and Cholewicki, 2003; Cristofolini et al. 2013). A wide range of research investigating prevalence of lower back pain and injury in cricket has been undertaken (Elliott et al. 1992; Hardcastle et al. 1992; Dennis et al. 2005; Johnson et al. 2012; Kountouris et al. 2012; Morton et al. 2013). However, whilst in vivo and in vitro analyses are able to assess mechanical factors that may increase risk of injury; studies looking specifically at lower back injury and pain in cricket fast bowlers have not been able to clearly define the mechanisms of injury (Morton et al. 2013). Therefore, in order to fully understand the pathomechanics, populations that may be at increased risk of lower back pain and injury must first be identified. This review will aim to consolidate and synthesise research looking at incidence of lower back pain and injuries in cricketers, and more specifically the fast bowling population.

2.2.2 Incidence of musculoskeletal injuries in cricket

Musculoskeletal injury in elite level cricket has an unusually high occurrence, despite the fact it is a non-contact sport (Orchard et al. 2006). Incidence of seasonal injuries has been reported at 17.2 injuries per team (Dennis et al. 2003). Fast bowlers alone have been reported to account for 44% of injuries sustained within the team (Stretch, 2014). Stretch (2014) reports that 38% of injuries sustained are to the lower limb, 33% back and trunk and 26% upper limbs. However, whilst these incidences are able to give us an indication of risk, they do not take into account the severity of injuries sustained. Thus, an injury with high incidence may not result in an equally high number of games missed. For example, wrist and hand injuries (10.6% incidence) resulted in 106 missed games, whereas lumbar spine injuries (10.4% incidence) resulted in 296 missed games

over 6 seasons (Orchard, 2002). Mount et al. (2014) recorded 73 injuries over the course of two seasons that resulted in 'time-loss', however 152 injuries were recorded that resulted in no time away from cricket. Consequently, 'missed games' may be a better measure of the impact of specific injuries on the cricketing population. Figure 2.2.1 displays the six injuries that caused the most missed games and which players sustained these injuries (Orchard, 2002).

Figure 2.2.1 supports the observation that fast bowlers in particular, have been shown to display the highest incidence of injury in the cricketing population (Elliott, 2000). Humphries et al. (2015) reported a risk ratio of 10.0 for a fast bowler's risk of receiving an abdominal wall injury when compared to the rest of the team. These higher risks of injury have also been observed in general injury rates; with fast bowling injury prevalence reported at 8% with an incidence of 3 injuries per 100 playing days (Mount et al. 2014). This high risk has been attributed to a multitude of factors, including excessive workloads, exposure to large magnitudes of force and weaknesses in key physiological structures such as the pars interarticularis (Dennis et al. 2004; Johnson et al. 2012). Consequently, research has attempted to analyse the relationship between these proposed risk factors and common injuries during fast bowling (such as those displayed at the lower back) (Elliott, 2000; Portus et al. 2004; Johnson et al. 2012; Morton et al. 2013).
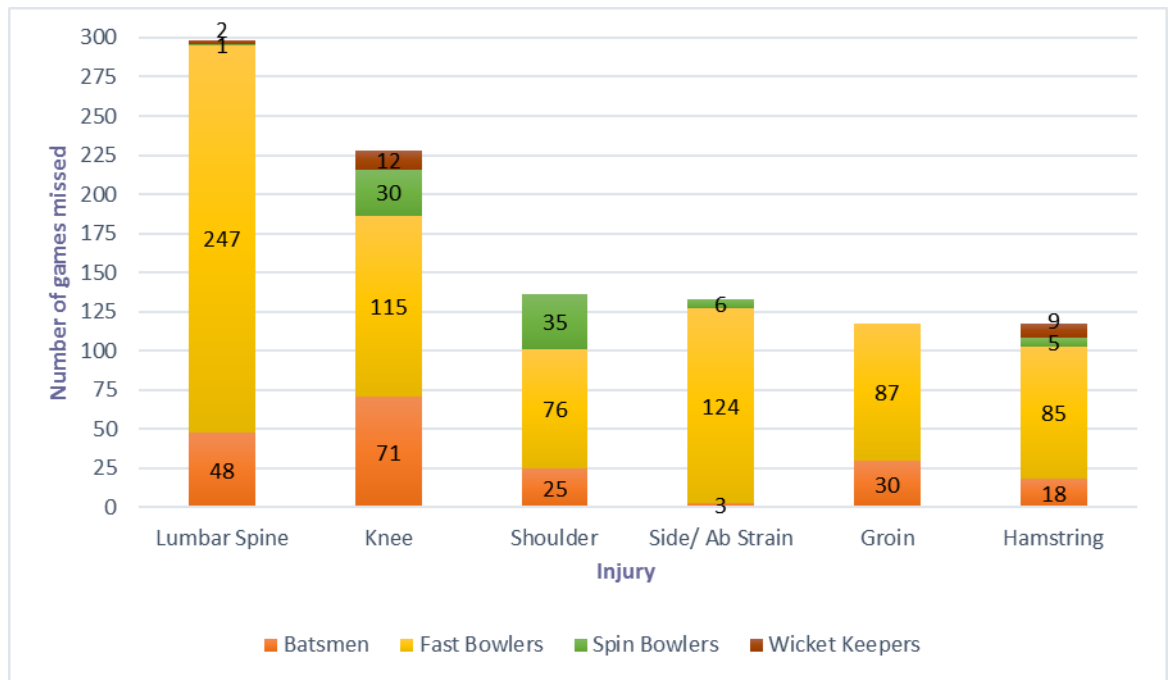
Figure 2.2.1. Data from Orchard et al. (2002) highlighting the number of games missed due to different injuries across 5 seasons for different playing positions.

2.2.3 Lower back injury diagnosis

Epidemiological studies have investigated the relationship between fast bowling and lumbar spine pathologies; concluding that fast bowlers are at increased risk of developing spinal pathological abnormalities (Ranson et al. 2005). However, a variety of different methods of screening have been cited in the fast bowling literature including X-ray, magnetic resonance imaging (MRI) and computerised tomography (CT) scans (Annear et al. 1992; Millson et al. 2004; Dennis et al. 2005; Crewe et al. 2012; Kountouris et al. 2012) . In order to gain an understanding of the impact and reliability of these studies an appraisal of screening methods is warranted.

X-rays are typically the simplist tool for the determination of lower back pathology, due to the speed and cost of this procedure. Consequently, many older studies have reported prevalence and incidence of lower back injuries (such as spondylolysis and spondylolithesis) using this method (Payne et al. 1987; Mackay et al. 1988; Annear et al. 1992; Millson et al. 2004). Whilst x-rays allow for a quick simple analysis, they are only able to provide a one-dimensional picture and cannot be used for soft tissue analysis (Patel, 2004). Thus, any abnormalities 'out-of-plane' or to the surrounding soft

tissue may be easily missed. However, as spondylyses are common lower back injuries in fast bowling x-rays remain a valid but limited screening tool.

MRI has been used in more recent studies looking at fast bowling pathologies, due to it's ability to more accurately and clearly define the scanned structures including soft tissue (Ranson et al. 2005; Engstrom and Walker, 2007; Crewe et al. 2012; Kountouris et al. 2012). Consequently, MRI is considered extremely useful for studies looking at intervertebral disc abnormalities (Patel, 2004). Additionally, unlike x-rays and CT scans, MRI scans operate without exposure to harmful radiation and as such repeated scans are more ethically possible. However, the cost associated with MRI scans is larger than that of x-rays and the lower accessibility to scanners means turnover may also be slower (Patel, 2004).

In addition to the introduction of MRI, CT scans have been reported less often, with only a few studies using CT scans in isolation (Foster et al. 1989; Millson et al. 2004). CT scans are optimal to image the boney structures however CT scanning is associated with a signifiacnt radiation dose. However, MRI is able to perform these functions to a similar degree of diagnostic accuracy and without exposure to harmful radiation, so CT scanning may only be the preferred choice with patients that may not be candidates for MRI (Patel, 2004).

The above appraisal demostrates that although reported results may be compared, it is important to consider method of screening as a consideration of accuracy and robustness. It may be the case that x-ray studies have screened a number of bowlers with intervertebral disc pathologies or more subtle bony abnormalities that have been overlooked as a result of the limitations, while MRI study numbers may be lower as a result of cost implications. A summary of fast bowling pathology studies' methodologies and findings can be seen in table 2.2.1.

2.2.4 Lower back injury risk in the fast bowling population

Studies invesitgating spinal pathologies have focussed primarily on the incidence of spondylolysis and disc degeneration; usually located at the L4/5 vertebra (Ranson et al. 2005; Ranson et al. 2010). Prevalence of bony abnormalities has been placed at between 24-55%; 74% of these abnormalities are accounted for by spondylolysis and 24% by

spondylolisthesis (calculated using weighted averages) (Elliott et al. 1992; Hardcastle et al. 1992; Stretch 1992; Engstrom and Walker, 2007). Hardcastle et al. (1992) placed disc degeneration prevalence of a sample of junior bowlers (16-18yrs) at 64%, with 36% displaying grade one degeneration; 21% grade two and 43% displaying grade three disc degeneration.

2.2.4.1 Spondylolysis prevalence

Spondylolysis has been defined as stress fractures located at the pars interarticularis, typically found at the lumbar vertebrae at L4 and L5 (Crewe et al. 2012). Fractures on the non-dominant side (non-bowling arm side) having a much higher prevalence than dominant side or bilateral fractures (Ranson et al. 2005; Morton et al. 2013). An example of spondylolysis can be seen in figure 2.2.2 (Hardcastle et al. 1992).
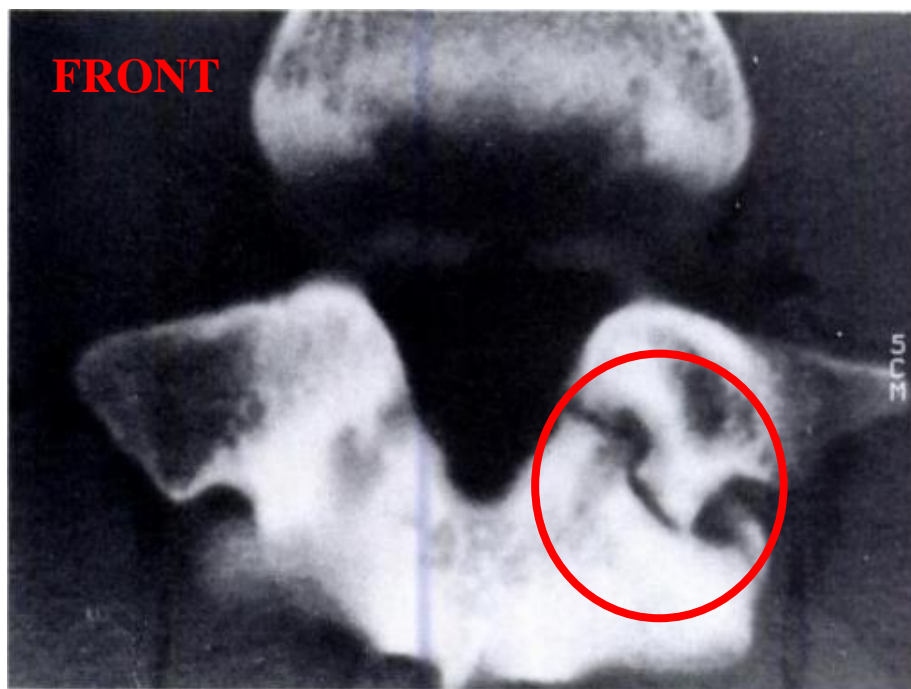
Figure 2.2.2. Reverse gantry CT scan showing spondylolysis (Hardcastle et al. 1992).

Spondylolysis has been reported as having high prevalence among fast bowlers (Hardcastle, 1993). Prevalence of around 22% in the fast bowling population has been observed; significantly higher than the 6-7% prevalence reported in the general male population (Engstrom and Walker, 2007). However, prevalence as high as 67% has been reported in some studies using x-rays (Hardcastle et al., 1992). Research has highlighted that pain may not always be a symptom of spondylolysis, consequently,

some disparity in results may exist due to cases not being identified or differences in methods of reporting injury (as some studies use self-reporting injury criteria) (Millson et al. 2004). This may also highlight the issue of players not reporting pain as a result of selection pressures (Crewe et al. 2012). Whilst, pain in certain cases may not occur, spondylolysis remains a significant issue in regards to time loss from the game (Orchard et al. 2002). MRI scans have enabled more recent studies to be able to more effectively identify players that have developed this pathology and in earlier stages, thus decreasing time loss as a result of this injury (Crewe et al. 2012; Kountouris et al. 2012). Spondylolysis has been widley reported in the literature and as such, professional coaches and players are becoming more aware of the risk it poses to the fast bowling population, resulting in more tailored training sessions and match strategies to reduce this risk. This is evident in junior bowling restrictions discussed later (ECB, 2016b).

2.2.4.2 Spondylolisthesis prevalence

Spondylolisthesis is typically developed from spondylolysis and can be defined as the anterior translation of one vertebrae relative to another (Jacobsen et al. 2007). In the fast bowling population, prevelence has been placed at between 2-25% compared with a 3% prevalence in the general male population (Annear et al. 1992; Elliott et al. 1992; Hardcastle et al. 1992; Engstrom and Walker, 2007; Jacobsen et al. 2007). Spondylolithesis is usually also observed at the L4/5 vertebra. A radiograph of spondylolisthesis can be seen in figure 2.2.3 (Kornblum et al. 2004).

Research has highlighted the importance of effective screening methods for fast bowlers 'playing through pain' in order to reduce the prevalence of spondylolisthesis, as recovery time from this condition is much longer than spondylolysis (Crewe et al. 2012). However, this recommendation may be problematic, due to the dissociation between spondylolysis and pain (Millson et al. 2004). As a result, in many cases the only method of tracking injury status is regular scans, which may become costly as MRI scans are likely to be the only viable option for this volume of repeated screening (Patel, 2004). This emphasises the need for further guidelines to highlight 'at risk' groups within the fast bowling population, so more accurate and costly injury surveillence methods can focus on these groups to prevent the development of more serious pathologies such as spondylolisthesis.
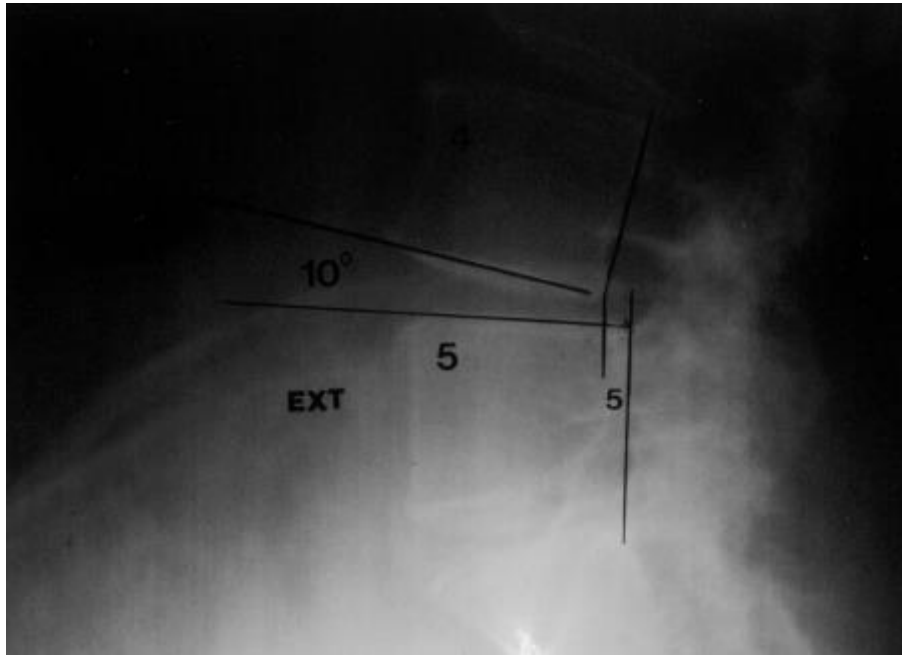
Figure 2.2.3 Radiograph showing spondylolisthesis at the L4 vertebrae (Kornblum et al. 2004).

2.2.4.3 Intervertebral disc degeneration

Research has analysed intervertebral disc degeneration, which although naturally occuring, it is more pronounced in fast bowlers (Elliott and Khangure, 2002). Intervertebral disc degeneration can be described as the progressive dehydration of a disc located between two vertebra (seen in figure 2.2.4) (Hadjipavlou et al. 2008). Imaging studies have placed prevalence of lumbar disc degeneration at between 14-65% in the fast bowling population (Elliott et al. 1992; Hardcastle et al. 1992; Elliot et al. 1993; Burnett et al. 1996; Ranson 2005; Crewe et al. 2012; Morton et al. 2013). In addition, Crewe and colleagues (2012) reported 75% of bolwers with intervertebral disc degeneration displayed a disc bulge and 17% of these participants also displayed an annular fissure (a breaking or separation of the outer layer of the intervertebral disc). Imaging studies have reported the most common sites of intervertebral disc abnormalities are at the L4-5 and L5-S1 discs, with 38% of individuals displaying disc degeneration at more than one site (Crewe et al. 2012).
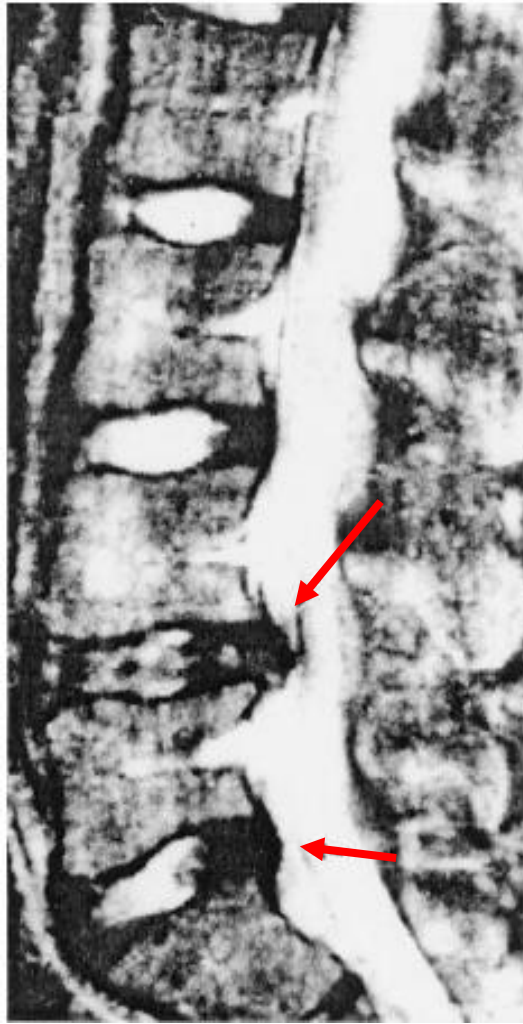
Figure 2.2.4. Intervertebral disc degeneration located at the L4-5 and L5-S1 discs (Hadjipavlou et al. 2008).

Table 2.2.1. An overview of study results and methodology investigating incidence and prevalence of common fast bowling pathologies and lower back pain.

Author	Participants (n)	Method of Diagnosis	LBP	Spondylolysis	Spondylolisthesis	Disc Degeneration	Comments
Kountouris et al. 2012	38 (mean age 15.5yrs)	MRI	45%	21%			
Crewe et al. 2012	46 (13-18yrs)	MRI		33%	4%	35%	
Engstrom & Walker 2007	51 (aged 13-17yrs)	MRI		24%			
Dennis et al. 2005	44 (mean age 14.7yrs)	Self/physiotherapist reported	52%	11%			
Ranson et al. 2005	36 (mean age 26yrs)	MRI	0%	81%		61%	Free of LBP inclusion criteria
Portus et al. 2004	30 (mean age 22.4yrs)	Mixed (injury history reported)		30%			
Millson et al. 2004	10 (mean age 18.7yrs)	CT Scan	90%	80%			Bowlers recruited with LBP
Gregory et al. 2004	39 (mean age 19.7)	Self-reported	100%	67%			Bowlers recruited with LBP
Orchard et al. 2002	12 (No ages given)	Mixed		33%			
Burnett et al. 1996	19 (mean age 13.6)	MRI	53%			58%	
Elliott et al. 1993	24 (mean age 13.7yrs)	MRI				21%	
Annear et al. 1992	20 (mean age 48yrs)	X-ray, CT scan and MRI.		33%	10%		Former fast bowlers
Elliott et al. 1992	20 (mean age 17.9yrs)	CT scan and MRI	40%	20%	25%	65%	

Study	n	Imaging				
Hardcastle et al. 1992	22 (mean age 17.9yrs)	CT scan and MRI	64%	36%	18%	14%
Foster et al. 1989	82 (mean age 16.8yrs)	CT scan		11%		
Mackay et al. 1988	72 (age 15-19yrs)	X-ray		20%		
Payne et al. 1987	12	X-ray		50%		

LBP, low back pain; n, number of participant; yrs, years; MRI, magnetic resonance imaging; CT, computerised tomography.

2.2.4.4 Non-technique based risk factors of spondylolysis and spondylolithesis

Spondylolysis has been well reported in literature and consequently its risk is well know to researchers, players and coaches, however the factors that affect this risk are still poorly understood (Morton et al. 2013). Pathological studies have highlighted age as a significant risk factor, reporting adolescent fast bowlers at an increased risk of spondylolysis due to immature structures in the spine and typically, weaker stabilising muscles around the lower back (Crewe et al. 2013). Gray et al. (2000) examined the reported ages of fast bowlers with spondylolysis or spondylolithesis in previous research, stating that the mean age of these bowlers was under 18 years old in four out of five studies analysed (Payne et al. 1987; Mackay et al. 1988; Foster et al. 1989; Annear et al. 1992; Elliott et al. 1992; Hardcastle et al. 1992). Whilst these studies are dated, similar findings are reported in more recent studies; suggesting that whilst research may have increased the understanding of the problem to some degree, the underlying causes and how to reduce this risk remain unanswered (Johnson et al. 2012; Morton et al. 2013).

As well as age, workload has also been emphasised as having a strong correlation with increased risk of injury (Orchard et al. 2006). Fast bowling workload is generally determined from number of balls or overs bowled, or 'days of bowling'. Research has suggested that bowling in excess of every 3.5 days significantly increases risk of injury (Dennis et al. 2005). Consequently, governing bodies have introduced restrictions on the number of overs adolescent fast bowlers can bowl and coach education at junior and senior level advises close monitoring of bowling workload in games and training in order to minimise risk. Table 2.2.2 shows recommendations set out by the England and Wales Cricket Board (ECB) for junior bowlers (ECB, 2016b). The ECB further recommends that no junior fast bowler should bowl more than 4 days in a week and no more than 2 consecutive days (ECB, 2016b).

Table 2.2.2. ECB fast bowling over limitations for junior fast bowlers

Bowlers Age	Max Overs Per Spell	Max Overs Per Day
Under 13yrs	5	10
14-15yrs	6	12
16-17yrs	7	18
18-19yrs	7	18

The relationship between risk of lumbar spine injury and quadratus lumborum (QL) asymmetry has also been explored, however research reports conflicting results between adult and junior fast bowlers (Engstrom and Walker 2007; Ranson et al. 2008; Kountouris et al. 2012). Larger dominant-side QL asymmetry was reported to increase risk of lumbar stress fractures in junior fast bowlers, however the same association between QL asymmetry and adult fast bowlers was not found (Engstrom and Walker 2007; Ranson et al. 2008). Consequently, it is still unclear as to whether QL symmetry is a significant risk factor, or if age and maturation have confounded reported results. Furthermore, it may be the case that QL asymmetry may be a result of injurous behaviours (such as spinal kinematics), but may not directly contribute to the mechanism of injury.

2.2.4.5 Non-technique based risk factors of disc degeneration

As with spondylolysis, young fast bowlers are placed at the highest risk of intervertebral disc abnormalities (Crewe et al. 2012). This may be due to increased elasticity in adolescent intervertebral discs, which do not resist shear forces (which have been reported to lead to increased disc degeneration) as effectively as matured, less elastic discs (Gray et al. 2000). Elliott and Khangure (2002) comment on an increase in prevelence of 44% (from 21%) between the ages of 13 to 18 years old. Due to it's progressive nature, high levels of degeneration at a young age is likely to continue to affect fast bowlers later in their career and quality of life after retirement. Prevelences of 70% were observed in retired fast bowlers (Annear et al. 1992). Due to its progressive nature, high workload has been highlighted as one of the most significant risk factors to lumbar disc degeneration (Dennis et al. 2005). Findings show that repetitive impacts, causing microtraumas below the critical limit of the intervertebral discs, cause a degenerative effect over time (Hreljac, 2004). This effect is experienced in the fast bowling population due to very high workloads during the season, may result in a significantly faster degenerative effect, or a more severe injury response.

2.2.5 Lower back pain

Foster et al. (1989) first linked fast bowling to lower back injury, stating that 11% of fast bowlers scanned showed stress fractures to the lumbar vertebrae, most often located at the L5 vertebrae. However, less research has focused on the incidence of lower back pain including that without a recognised diagnosis of injury (Orchard et al. 2006). LBP

has been reported to result in 247 missed games in Australian domestic and international matches, between 1995-2001, with 22.4% of all time missed being due to LBP in the fast bowling population (Orchard et al. 2002). Studies investigating LBP associated with fast bowling have reported incidence of between 40-64% (Elliott et al. 1992; Hardcastle et al. 1992; Dennis et al. 2005; Kountouris et al. 2012). This compares to incidences of between 6-36% reported in the general population (Biering-Sørensen, 1982; Croft et al. 1999; Hestbaek et al. 2003; Cassidy et al. 2005; Mustard et al. 2005).

Studies have highlighted that spinal abnormalities may exist without the sensation of pain, therefore lower back pain (LBP) should be investigated as a separate entity (Millson et al. 2004). Debilitating pain is generally quantified in studies as: pain that causes the bowler to have time out of games or training sessions (Dennis et al. 2005). Conversely, lower back injury (bony or soft tissue) typically can only be conclusively quantified via radiological analysis. Thus, LBP may include any pain, with or without a formal diagnosis of injury that causes time out of training or matches. Whereas lower back injury, including bony and soft tissue, requires a formal diagnosis. Consequently, there is likely to be significant overlap between these two groups. Hardcastle (1992) found that 82% of bowlers with pars interarticularis defects experienced pain as a symptom. Hardcastle (1992) explored the ratio of injury to pain, stating that 45% of the young bowling population suffer pain because of injuries such as spondylolysis. Burnett et al. (1996) further commented on the correlation between disc degeneration and incidence of pain, concluding this relationship to be inconsistent. Despite these findings, few studies have reported an in-depth analysis of LBP independent of pathological diagnosis and the risk factors involved. Hence, while the exact mechanisms of lower back injury remain ambiguous, the relationship between fast bowling and LBP remains even more unclear (Morton et al. 2013).

Both acute and chronic lower back pain are common in fast bowling (Dennis et al. 2005). It has been highlighted that the most common cause of lower back pain in 'under 20's' is stress fractures, even though other abnormalities may exist, they will not necessarily be the cause of pain (Millson et al. 2004). Conversely, more mature fast bowlers have been shown to suffer pain as a result of musculoskeletal tears, including rib-tip syndrome, facet dysfunction and lumber intervertebral disc protrusions. In addition, it must also be noted that pain may manifest at different stages of activity. Many bowlers will experience pain during the end of a long spell, while many may not

experience pain until much later, after they have finished bowling. In opposition, in some cases, pain may be so severe that it is impossible to bowl due to pain (Millson et al. 2004). As a result of this, statistics relating to incidence of injury and pain may be skewed, as pain and injury criteria may only record debilitating pain. Furthermore, more advanced radiographical studies may report early stage injuries in the absence of pain, whilst x-rays may not be able to highlight these findings until they become more severe (most likely during times of high bowling workload) (Patel, 2004). Thus, timing of screening is an important consideration when looking at injury and LBP incidence and prevalence.

2.2.6 Technique related risk factors of lower back pain and injury

The relationship between the biomechanics of fast bowling and LBP and injury has received much attention in current literature (Glazier, 2010; Stuelcken et al. 2010; Johnson et al. 2012; Morton et al. 2013). Nevertheless, the correlation between fast bowling technique, injury and pain is still relatively unclear. Research has highlighted shoulder counter-rotation in excess of 30° during bowling may put fast bowlers at increased risk of lower back injury; however, this variable only takes orientation of the shoulders into account and therefore cannot be used as a measure of spinal kinematics (Elliott, 2000; Portus et al. 2004). Excessive lateral flexion has also been reported to increase risk of lower back injury, yet no thresholds defining what may be classed as 'excessive' have been published (Ranson et al. 2008). In addition, some studies have suggested a combination of lateral flexion and extension may increase the risk of pain (Glazier, 2010). These conflicting results may be a result of heterogeneity in the methodologies used to record spinal kinematics during bowling or different mechanisms being displayed as a result of the individual nature of bowling technique and physiological predisposition to lower back injury and pain (Portus et al. 2004). In order to understand the relationship between fast bowling biomechanics, injury and pain, the biomechanics of fast bowling must first be understood. This is explored in greater detail later in this chapter, before any further hypotheses and recommendations can be made relating to injury and pain.

2.2.7 Conclusion

Research has highlighted that fast bowlers are at a significantly higher risk of lower back pain and injury, compared with the rest of the team (Morton et al. 2013).

Furthermore, this risk is increased in adolescent fast bowlers, however despite this elevated risk the cause is yet to be determined, but may related to immaturity of the pars (Hardcastle et al. 1992). Research has investigated lower back injury in greater detail than lower back pain, with the majority of studies reporting spondylolysis prevalence (Portus et al. 2004; Engstrom and Walker, 2007; Crewe et al. 2012). However, it may be more beneficial to look at the relationship with lower back pain and fast bowling in greater detail, as it is pain that will result in missed match and training time. Nevertheless, spinal pathology linked to fast bowling and non-technique based risk factors of lower back injury and pain in cricket are generally well understood (Morton et al. 2013). However, more research is needed to identify the exact relationship between fast bowling technique factors and risk of lower back injury and pain.

2.3 Kinetics during cricket fast bowling

Cricket fast bowling has been described as a repetitive, high impact skill that places significant stress on the lower limbs and spine (Johnson et al. 2012). Due to the nature of cricket, exposure to these impacts can be over an extended period of time. Some formats of the game may require bowlers to bowl over a period of four or five days, often bowling 50-60 overs per week as a result (Dennis et al. 2005). Previous studies have focussed on the relationship between ground reaction forces (GRF) at back and front-foot impact, risk of injury and performance concluding that higher GRF may result in greater ball release speeds but increased risk of injury (Elliott and Foster, 1984; Elliott et al. 1986; Foster et al. 1989; Mason et al. 1989; Elliott et al. 1992; Elliott et al. 1993; Hurrion et al 2000; Portus et al. 2004; Crewe et al. 2013; Worthington et al. 2013; Bayne et al. 2016). However, no significant relationship between bowling impacts and risk of injury has been reported. It is hypothesised that the combination of high workloads with large impacts, may significantly contribute to the cumulative effect that high ground reaction forces are having on the body (Orchard et al. 2002). Consequently, it is vital to obtain a clear understanding of how technique variables may affect these forces. This understanding will enable further analysis of these factors in relation to performance and risk of injury.

2.3.1 Ground reaction force (GRF) and fast bowling

GRF in fast bowling has been frequently reported as a key variable within studies analysing fast bowling performance and risk of injury. The most commonly reported

values are that of peak vertical and horizontal force perpendicular to the wicket, so called braking GRF (Hurrion et al. 2000; Portus et al. 2004; Stuelcken et al. 2009; Crewe et al. 2013; Worthington et al. 2013). These values are typically taken at the front-foot impact phase of the delivery stride, as this is likely to be the point at which peak GRF is greatest. All studies reported GRF normalised to body weight (BW) therefore enabling quantitative synthesis through weighted averages. A total of 378 participants yielded a weighted average (±SD) vertical GRF of 5.8 (1.3) BW and horizontal braking GRF of 3.2 (1.1) BW at front-foot impact. These results can be seen in table 2.3.1. Running studies have reported peak GRF around 2-3 times body weight vertically and 0.5 times body weight for breaking GRF; highlighting that repetitive impacts of these magnitudes may place runners at risk of injury (Gottschall and Kram, 2005; Hall et al. 2013; Zadpoor and Nikooyan, 2011). Consequently, with fast bowlers experiencing around double these magnitudes, it may be hypothesised that risk of repetitive stress injuries is even greater in this population. This has been highlighted previously in this chapter, stating that excessive fast bowling workloads significantly increases risk of injury (Dennis et al. 2005).

Table 2.3.1. Synthesis of results for mean vertical and breaking GRF for front-foot impact with weighted averages (SD) calculated from previous studies.

Author	n	Vertical GRF (BW)	Braking GRF (BW)
Bayne et al. 2016	25	4.9	3.4
King et al. 2016	20	6.7	4.5
Middleton et al. 2016	15	3.5	2.1
	15	4.5	2.8
Spratford and Hicks, 2014	17	6.3	4.1
	17	5.9	4.0
	17	6.1	4.3
Worthington et al. 2013	20	6.7	4.5
Crewe et al. 2013	23	4.9	3.3
Portus et al. 2004	42	7.3	4.5
Hurrion et al. 2000	6	4.8	3.5
Elliott et al. 1993	19	4.8	2.1
	5	5.2	2.6
Elliott et al. 1992	20	6.4	1.9
Mason et al. 1989	15	9	2.0
Foster et al. 1989	82	5.4	2.5
Elliott et al. 1986	15	4.1	1.6
Foster and Elliott, 1985	1	3.8	1.4
Elliott and Foster, 1984	4	4.7	1.8
Weighted Average (SD)	**378**	**5.8 (1.3)**	**3.2 (1.1)**

n, number of participants; BW, body weight; GRF, ground reaction force.

Whilst magnitudes of fast bowling impacts are useful metrics for quantifying impact characteristics, in isolation these values fail to address the rate dependant nature of many musculoskeletal injuries and performance indicators, and thus may be the reason studies have not been able to link large impacts with increased risk of injury. Thus, it is unclear whether high loading rates affect risk of chronic injuries, such as stress fractures, or if it is only a factor in acute injuries. Hreljac (2004) suggests it may be a factor in both chronic and acute injuries in runners, however this has not been explored in cricket fast bowlers. This has led to some studies also reporting time-to-peak GRF (Hurrion et al. 2000; Portus et al. 2004; Crewe et al. 2013; Worthington et al. 2013). Weighted averages of 49 (4) ms vertically and 55 (6) ms horizontal braking were calculated (seen in table 2.3.2). However, loading rate (reported in running studies and other impact analyses) may be a better measure for this purpose; incorporating both peak and time-to-peak values in its calculation, nonetheless few fast bowling studies have reported this metric (Gottschall and Kram, 2005; Crewe et al. 2013).

Table 2.3.2. Synthesis of results for mean time to peak vertical and braking GRF for front-foot impact with weighted averages (SD) calculated from previous studies.

Author	n	Time to peak vertical GRF (ms)	Time to peak braking GRF (ms)
Worthington et al. 2013	20	30	30
Crewe et al. 2013	23	34	37
Portus et al. 2004	9	60	70
	11	80	80
	7	80	80
	6	90	110
Hurrion et al. 2000	6	26	
Weighted Average (SD)	**82**	**49 (4)**	**55 (6)**

n, number of participants; GRF, ground reaction force; ms, milliseconds.

Fewer studies have reported GRF at the back-foot impact (BFI) phase of the bowling action (Mason et al. 1989; Saunders and Coleman, 1991; Elliott et al. 1992; Hurrion et al. 2000). However, as it is the start of the delivery stride, it may display an important relationship with mediating GRF at front-foot impact; a relationship which at present has not been widely reported (Hurrion et al. 2000). Current studies reporting GRF at BFI have compared magnitudes with FFI, and it's use in defining the start of the delivery stride, with little further explanation on how BFI may affect the delivery stride. Consequently, this relationship may be important to consider when analysing variables related to performance and injury risk. Studies that have reported GRF at back-foot

impact have reported peak vertical GRF between 2-2.9 times body weight and peak braking GRF between 0.94-1.1 times body weight, with no back-foot impact time-to-peak reported (Mason et al. 1989; Saunders & Coleman, 1991; Elliott et al. 1992; Hurrion et al. 2000). Consequently, the rate-dependant relationship between back-foot impacts and injury risk cannot be explored at present.

2.3.2 GRF and fast bowling performance

A small number of studies looking at GRF have attempted to directly link GRF variables to fast bowling performance measures such as ball release speed (Portus et al. 2004; Worthington et al. 2013). However, these findings are not clear as to the exact relationship of GRF and performance. Portus et al. (2004) found that the fastest bowlers produced the highest peak forces and loading rates. However, Worthington and colleagues (2013) contradict these findings stating that there was a positive correlation between the total horizontal impact at front-foot impact and ball release speed as opposed to peak vertical GRF and loading rate. These mixed conclusions may be a result of variations in bowling technique, as it has been acknowledged that a multitude of kinematic and kinetic factors contribute to variations in ball release speed (Worthington et al. 2013). Consequently, studies only looking at GRF in isolation may not be appropriate to accurately describe relationships with performance. Thus, only monitoring GRF when considering performance may not be appropriate.

2.3.3   GRF and fast bowling injury and pain

As highlighted above large impacts are seen at both back and front-foot impact during the fast bowling delivery stride (Crewe et al. 2013; Worthington et al. 2013; Bayne et al. 2016; King et al. 2016). Hence, studies have hypothesised that large GRF increases risk of stress related injuries (Portus et al. 2004). With lower back injury highlighted as a significant issue within the fast bowling population, research has focussed on the relationship between impact characteristics during the delivery stride and injuries such as spondylolysis (Johnson et al. 2012; Morton et al. 2013). At present no studies have reported a significant relationship between GRF and injury risk; this may suggest that large impacts alone may not contribute to increased injury risk. It has been proposed that large impacts may be a contributing factor to increased injury risk when coupled with other variables such as high workload (which has been linked to increased lower back injury risk) as well as bowling kinematics that may place the spine in a position of

weakness during time of high loads (Dennis et al. 2005; Orchard et al. 2010; Morton et al. 2013; Orchard et al. 2015). Moreover, the fact that most studies have only reported GRF limits the information that can be reported relating to loads experienced further up the body. Whilst force plates have been validated and reported to display high degrees of reliability and sensitivity; without kinematic measures, they are not able to collect data for impacts further up the body (Walsh et al. 2006). Consequently, a synchronised three-dimensional motion analysis system (as seen in more recent fast bowling studies) is required (Crewe et al. 2013; Worthington et al. 2013; Bayne et al. 2016). Crewe et al. (2013) reports lumbar loading during the delivery stride highlighting larger loads with higher ball release speed and a straighter front knee, but no changes in load throughout a bowling spell. However, as no injury data was collected the relationship between these findings and risk of lower back injuries is unclear. Bayne and colleagues (2016) is the only study to date to report lumbar load alongside lower back injury data. Findings suggest that larger peak lumbar flexion and lateral flexion moments may increase risk of lower back injury. Whilst lateral flexion range of motion has been hypothesised to increase risk of lower back injury (discussed later in this chapter), greater flexion and lateral flexion moments displayed in the injury group is a novel finding and thus further investigation exploring this proposed mechanism of injury is needed to consolidate findings (Glazier, 2010).

Limited evidence exists exploring the relationship between impacts further up the body and lower back injury; however due to the reported dissociation between lower back injury and lower back pain, lower back pain should be explored as a separate issue (Millson et al. 2004). However, research has not investigated the relationship between GRF and lower back pain, thus future research exploring this relationship is warranted.

2.3.4 Factors affecting GRF in fast bowling

Once the effect of impact characteristics on performance and injury risk are known, it is necessary to discover which variables can be manipulated in order to lower this risk without compromising performance, or increase performance without increasing injury risk. Research has previously reported factors that may affect GRF: Velocity of the colliding objects and the material properties of the dampening elements (soft tissue, shoes and surface of contact) have been shown to contribute significantly to the resulting GRF (Nigg, 1983). Therefore, understanding the contribution of each of these

variables on GRF may be valuable in understanding the pathomechanics of fast bowling injury.

<u>2.3.4.1 Approach Speed and GRF</u>

Few studies have assessed the relationship between approach speed and magnitude of GRF, with more studies assessing the relationship between approach speed and ball release speed (Hurrion et al. 2000; Worthington et al. 2013). Hurrion et al. (2000) suggests that higher peak braking GRF may be associated with higher approach speeds, due to the need to slow down for an effective ball release. Findings by Worthington et al. (2013) supports this suggestion reporting that a faster run up speed produced a significantly higher peak vertical and horizontal GRF. These findings suggest that a lower approach speed may be more advantageous as less braking forces would be required to slow down to the required speed. However, it is likely higher approach speeds elicit a greater transfer of momentum through the body resulting in higher ball release speeds and therefore a technique that can dissipate higher GRF without reducing transfer of momentum may be favourable.

63% of peak vertical ground reaction force variance has been reported to be explained by initial foot angle, plant angle and approach speed with initial foot angle and run-up speed also explaining 31% of variance in peak horizontal GRF (Worthington et al. 2013). Whilst it may be pivotal to ball release speed, Nigg (1983) suggests faster approach speeds may be a significant factor in the resulting magnitude of ground reaction force. Thus a compromise between an approach speed that produces a high ball release speed without producing excessive GRF may be required. Consequently, initial foot angle at front-foot impact and ankle angle when the foot is flat may be of interest to coaches wishing to mediate GRF magnitudes without sacrificing momentum. However, current studies do not define what may be considered generically 'excessive' as it is likely to be closely linked to workload and physiological factors and thus, a personalised value (Elliott, 2000; Orchard et al. 2006). As such coaches do not have the required information to monitor this 'compromise'. This highlights the need for normative impact charcateristics to be collected alongside workload. However, currently force plate technology used to collect this data is not typically accessible to most clubs, and thus alternative methods may need to be explored to address this.

The relationship between bowling kinematics and GRF has been widely reported in the literature (Foster et al. 1989; Portus et al. 2004; Worthington et al. 2013). Studies have typically focused on the effect of lower leg kinematics on magnitude of peak and time to peak GRF (Elliott, 2000; Portus et al. 2004; Worthington et al. 2013). Earlier studies have reported that greater knee flexion may decrease peak and time-to-peak GRF, due to the increased capacity of impact absorbtion when compared with a fully extended, or 'braced' front leg (Elliott, 2000; Portus et al. 2004). However, it has also been stated that increased knee flexion may decrease ball-release speed as the hip cannot be used as effectively as a lever (Portus et al. 2004). Thus, a technique consisting of a flexed knee on impact which is then extended at the point of ball release has been suggested to decrease risk of injury whilst maximising ball-release speed (Bartlett et al. 1996). Conversely, more recent studies have observed that higher degrees of knee flexion were synonymous to quicker time-to-peak GRF (Worthington et al. 2013). This was hypothesised to be a result of players not being able to withstand higher loading rates with an extended knee at initial contact and flexing their knee in order to reduce the impact force experienced (Worthington et al. 2013). These differences in findings may be a result of differing methodologies. Portus et al. (2004) used video analysis at a sampling frequency of 50Hz which cannot provide an analysis as comprehensive as more recent studies analysing front-foot impact at 300Hz using three-dimensional motion analysis systems. Due to the highly ballistic nature of fast bowling, 50Hz may not be adequate to avoid aliasing of relevant data points (such as point of initial contact) and as such this may explain the discrepancies with more recent studies.

Although most studies have focused on the relationship between lower limb kinematics and GRF, the measurement of GRF is limited to the description of the interaction between the playing surface and foot (and footwear). Consequently, research has attempted to quantify impact force experienced further up the body (Crewe et al. 2013; Bayne et al. 2016). Ball-release speed has been reported to produce greater lumbo-pelvic anterior-posterior force (r = 0.529) and medio-lateral force (r = 0.391) (Crewe et al. 2013). However, the value of these results in isolation may be questioned, as coaches and players may be unwilling to sacrifice bowling speed in order to lower risk of injury and they do not provide an insight into any technique variables that may be altered to decrease these forces. Bayne et al. (2016) reported greater pelvis rotation was associated

with higher lumbar loads, which may suggest increases in shoulder counter-rotation is a result of whole body rotation (not just at the shoulders) and the increase in lumbar loads may not be a result of differences in spinal kinematics, rather an effort to increase ball release speed (Elliott et al. 2005; Crewe et al. 2013; Bayne et al. 2016). However, this still does still not provide coaches with a practical target for decreasing lumbar loads. Thus, further research is required to highlight technique variables that may aid dissipation of bowling impacts and consequently, decrease risk of lower back pain and injury.

2.3.4.3 Footwear and GRF

Few studies have explored the effect of footwear on GRF during bowling. Bishop et al. (2010) reports a case study of different shoe types. Although, findings are novel, the relevance of these results to the larger fast bowling population may be questioned. No significant difference existed between GRF experienced with a standard cricket shoe and a bowling (high top) shoe (Bishop and Thewlis, 2010). However, the bowling shoe did significantly increase plantarflexion angle at front-foot impact and therefore significantly increased knee joint moment at front-foot impact. Nonetheless, no data exploring how this may affect impacts further up the body were reported. Worthington et al. (2013) suggested that a larger plantarflexion angle at impact allows a greater range of dorsiflexion to attenuate forces throughout front-foot impact, however Bishop et al. (2010) observed no such difference in ankle range of motion with differing intial plantarflexion angles and no difference in peak or time-to-peak GRF was observed between different shoes.

Although there is a lack of cricket-based research looking at the effect of footwear, other sports such as running have been explored in more detail (Wright et al. 1998; Hardin et al. 2004; Squadrone and Gallozzi, 2009). Typically this research has focussed on comparisons between barefoot and shod running (De Wit et al. 2000; Divert et al. 2005; Squadrone and Gollozzi, 2009). Findings suggest that changing from shod to barefoot conditions causes alterations in running kinematics to reduce impact forces upon landing (Squadrone and Gallozzi, 2009). Thus, although impact conditions were significantly different, no differences in GRF magnitude were observed (Squadrone and Gallozzi, 2009). Shorter et al. (2008) reported that the addition of a 1cm heel raise incorperated with the cricket shoe significantly decreased horizontal loading rates in

four out of six fast bowlers tested. It may therefore be the case that a heel raise might be a practical solution to affect bowling technique without too great an impact on the bowlers performance, however further research is needed in order to confirm this and explore it's effect on different bowling techniques (Shorter et al. 2008).

2.3.4.4 Playing Surface and GRF

Cricket pitches display very high firmness values in order to facilitate the required ball-surface interaction (bounce) and therefore display significantly higher surface firmness than other playing surfaces (Carre et al. 1999). Research reports football pitch hardness to average 42.6g using a clegg hammer classification system; this compares to cricket pitch values ranging between 176-388g dependant on time of year (Cannaway et al. 1990; Baker et al. 1998). Thus, if no other factors are taken into account, it may be assumed that fast bowlers experience signficantly higher GRF than a footballer during a similar movement. However, it has been demonstrated in running that despite varying surface firmness, individuals may actually alter the performance of the task in order to accommadate the surface properties (Hardin et al. 2004). This highlights that consistency of playing surface is needed if testing in the field. Exploring the surface properties in isolation may be unable to provide the necessary information regarding the effects on the actual individual. To this end studying the human-surface interaction may offer new insights into fast bolwing. No fast bowling research has investigated this interaction and thus, the question of how surface properties affect front-foot impact during fast bowling remains unanswered.

All studies accept one investigating GRF in fast bowling have used a laboratory mounted force plate (Elliott and Foster, 1984; Foster and Elliott, 1985; Elliott et al. 1986; Foster et al., 1989; Mason et al. 1989; Elliott et al. 1992; Elliott et al. 1993; Hurrion et al 2000; Portus et al. 2004; Crewe et al. 2013; Worthington et al. 2013). Whilst force plates are a reliable and valid data collection method that enables a wealth of impact characteristics to be recorded, it is limited to describing the foot-force plate interaction without the addition of other technologies. This limits environmental factors that can be analysed such as the effects of different playing surface interactions. Studies have attempted to place material over the force plate (polyflex surface and artificial grass), however no adjustments to the calculation of GRF were made to allow for the added force plate-surface interface (Hurrion et al. 2000; Stuelcken and Sinclair, 2009;

Worthington et al. 2013). Furthermore, although it has been attempted, no simple method of effectively recording fast bowling impacts outside a laboratory environment has been found (Hurrion et al. 2000). Therefore, 'in-field' fast bowling cannot be effectively analysed using current methods. In order to overcome the limitations associated with force plate technology, new technologies and their application to cricket fast bowling should be explored.

Novel methods using accelerometer derived accelerations, have been shown to strongly correlate with GRF during jumping (Sell et al. 2014). Furthermore, studies investigating impact attenuation during activities such as running and falling have successfully utilised accelerometry to produce methodologies that can be implemented outside of a laboratory environment, with no 'landing area' restrictions (Crowell et al. 2010; Theobald et al. 2010). Such methodologies may be useful in a fast bowling environment, however this has not yet been implemented (Bali et al. 2011).

2.3.5 Conclusion

This review has highlighted current literature pertaining to impact characteristics during fast bowling. Although, high impacts and loading rate have been hypothesised to increase risk of injury, no current studies have reported a significant relationship. It has been highlighted that GRF as an isolated predictive measure of injury or performance may not be relevant, due to the multitude of kinematic and physiological variables that affect how these forces are transmitted through the body. Consequently, it may be more appropriate to measure impacts experienced at the physiological landmarks of interest, such as the lumbar spine, rather that limiting analysis to the foot-playing surface interaction by reporting GRF values. Only two studies to date have reported impacts higher up the body during fast bowling, and one of which did not look at the relationship with risk of injury (Crewe et al. 2013; Bayne et al. 2016). This lack of reporting may, in part, provide some explanation as to why the relationship between fast bowling kinetics and injury remains unclear. Further work may also need to be undertaken in order to fully understand the effect of other variables, such as playing surface, on bowling kinetics as these factors have been reported to significantly affect impact conditions but have not been properly explored within fast bowling literature. Moreover, research is needed to explore the relatioship between fast bowling impacts and lower back pain. This review has highlighted the need for analysis of kinematic

variables alongside impacts to gain a true understanding of the mechanisms of fast bowling injury, which at present remain ambiguous. The next section in this chapter aims to explore current literature reporting these kinematic variables.

2.4 The Kinematics of Fast Bowling in Cricket

The previous reviews in this chapter have investigated lower back pain (LBP) and injury in the fast bowling population, as well as currently known findings relating to fast bowling impacts during the delivery stride. These reviews highlighted that whilst these topics have been given much attention in the literature, exact mechanisms of LBP and injury are still unclear. The hypothesis that high fast bowling impacts increases risk of LBP and injury has not been supported by current findings, and thus, it has been suggested that a combination of high impacts and fast bowling kinematics may predispose bowlers to increased risk. Consequently, a review of what is currently known about fast bowling kinematics is warranted.

Kinematic studies have classified the action of fast bowling as an explosive, high-impact activity, as the bowler will typically aim to release the ball with the highest velocity possible (Ranson et al. 2009). This, coupled with the fact that techniques generally require bowlers to use a combination of movements in all three planes of motion, highlights issues with fast bowling studies that have only been able to analyse kinematics using two-dimensional video analysis (Foster et al. 1989; Burnett et al. 1995; Hurrion et al. 2000; Portus et al. 2004; Elliott et al. 2005). Whilst this has provided some insight into fast bowling technique, it is not able to explore movements and mechanisms in three-dimensions. However, classification of bowling technique has remained predominantly two-dimensional even with the introduction of three-dimensional analyses (Johnson et al. 2012).

2.4.1 Classification of Bowling Action

Most studies have suggested three main sub-categories of fast bowling technique; front-on, side-on and mixed action, however more recent studies have proposed a fourth 'semi-open' action (Johnson et al. 2012). Although older studies report bowling action classification solely on shoulder alignment, more recent studies typically classify bowling action by degrees of shoulder counter-rotation, shoulder and hip orientation (seen in figure 2.4.1) (Elliott et al. 1992). Thus, this gives some descriptive data relating to kinematics of the spine during bowling. Nonetheless, due to limitations and heterogeneity of data capture methods, comparisons between studies outlining spinal kinematics are difficult. Characteristics of these actions can be seen in table 2.4.1.
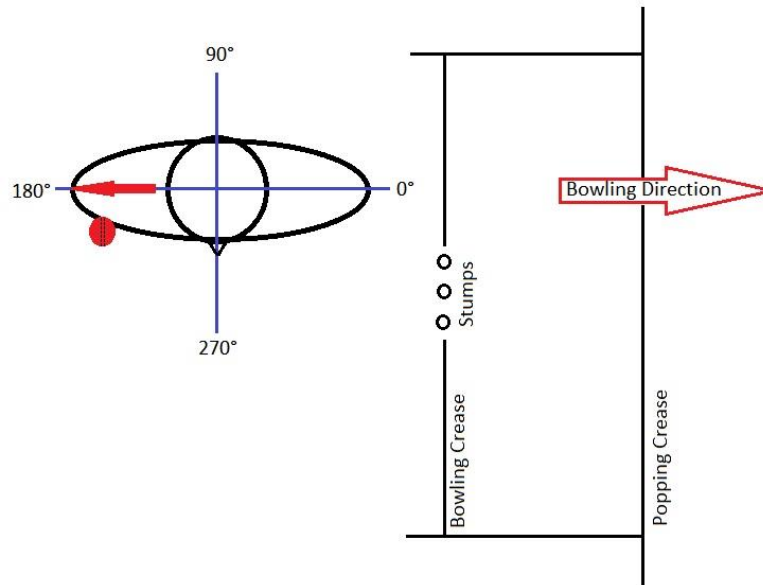
Figure 2.4.1. Method of defining shoulder and hip orientation measurement

Table 2.4.1. Bowling action classifications (Elliott et al. 1992).

Action Classification	Shoulder Orientation	Hip Orientation	Shoulder Counter-rotation	Hip-shoulder Separation
Front-on Action	>240°		<30°	<30°
Semi-open Action	210-240°		<30°	<30°
Side-on Action	<210°		<30°	<30°
Mixed Action	Front or Side On	Side or front on (opposite to shoulders)	>30°	>30°

°, degrees.

Shoulder and Hip Orientation: Orientation of the shoulders is typically taken using the popping crease (from which the bowler bowls) as zero degrees, therefore a fully side-on action would display a shoulder and hip orientation on 180° (Bartlett et al. 1996).

Shoulder Counter-Rotation: Shoulder counter-rotation is determined by subtracting the minimum shoulder alignment angle, from shoulder alignment at back-foot impact (Ranson et al. 2009).

Hip-Shoulder Separation: Hip-shoulder separation angle is calculated from subtracting hip alignment angle from shoulder alignment angle. Therefore, a positive separation

angle refers to the shoulders being in a more front-on alignment than the hips (Portus et al. 2004).

2.4.1.1 Front-on action

The front on action has been adopted by a large percentage of modern fast bowlers, although an exact percentage has not been reported (Ferdinands et al. 2009). This may be due to recommendations made in previous studies stating that front-on or side-on actions may reduce risk of back injury (Foster et al. 1989; Stockill and Bartlett, 1992; Burnett et al. 1996). This may be due to the relatively low rotation of the lumbar spine displayed during this action as a result of both the hips and shoulders facing the batsmen (as seen in figure 2.4.2a).

2.4.1.2 Side-on action

As with the front on action, a side-on technique has been reported to produce low lumbar rotation due to both shoulders and hips orientated 'side-on' to the batsmen (seen n figure 2.4.2b). This technique however is less commonly seen than the front-on technique, as it is characterised by a slower approach speed, which may be deemed to produce a lower bowling speed (Elliott and Foster, 1984; Foster et al. 1989).

2.4.1.3 Semi-open action

This classification has been adopted relatively recently, developed as a 'safe' alternative to the mixed action (Johnson et al. 2012). Shoulder and hip orientation is between a front-on and side-on orientation with very little shoulder counter-rotation displayed in this action. This can be seen in figure 2.4.2c.

2.4.1.4 Mixed action

This technique adopts a mix of front-on and side-on techniques, with either 'front-on' hips with 'side-on' shoulders or 'side-on' hips and 'front-on' shoulders (seen in figure 2.4.2d) (Portus et al. 2004). Consequently, this action has been classified as displaying the greatest shoulder counter-rotation angles and therefore, may place bowlers at increased risk of back injury based on our current understanding of risk (Portus et al. 2004). However, this technique has been identified as the most prevalent action among modern fast bowlers, as it is easiest to produce higher bowling speeds (Elliott et al. 2005).
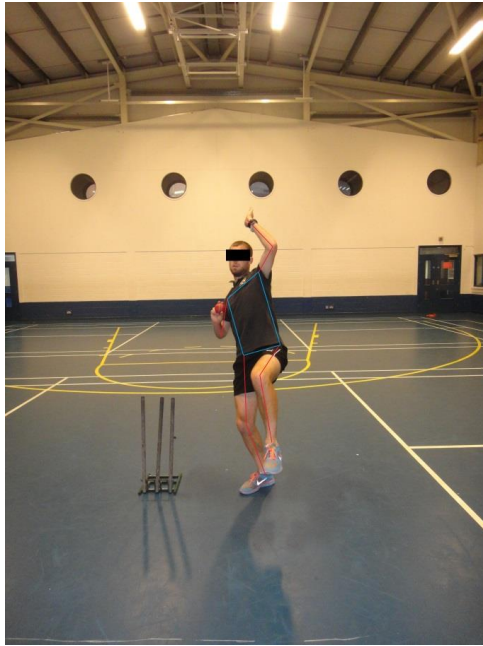
Figure 2.4.2a. Front-on bowling action defined with hip and shoulder alignment.
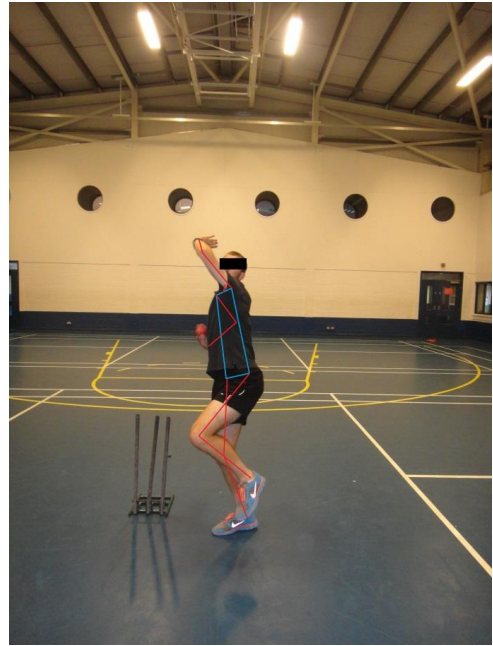
Figure 2.4.2b. Side-on bowling action defined with hip and shoulder alignment.
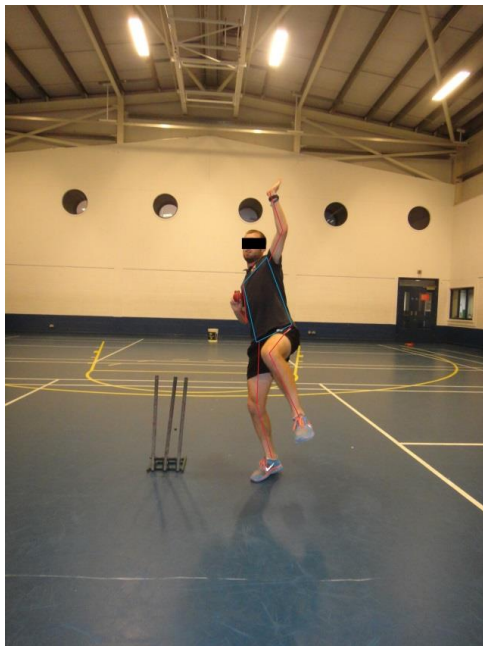
Figure 2.4.2c. Semi-open bowling action defined with hip and shoulder alignment.

Figure 2.4.2d. Mixed bowling action defined with hip and shoulder alignment.

Research typically divides the fast bowling delivery into four sections: Pre-delivery/run-up, pre-delivery stride, delivery stride and follow through (Bartlett et al. 1996).

2.4.2 Pre-delivery/run-up

2.4.2.1 Run-up Distance

Very little research has been undertaken regarding an optimal run-up distance, as the main function is to allow the bowler to reach the required run-up speed for their specific style of bowling (Mason et al. 1989). Consequently, run-up distances vary from bowler to bowler. The run-up has been defined as the distance travelled from when the bowler passes their marker and begins to jog, to the pre-delivery stride (Bartlett et al. 1996). Davis and Blanksby (1976) report that regardless of the style of bowling a "14-pace" run-up will be sufficient to release the ball at optimal velocity. However, the theory that faster run-up speed is not directly proportional to a faster ball release speed is emphasised throughout other studies, as effective technique is a key issue (Elliott and Foster, 1984). Davis and Blanksby (1976) produced conflicting findings; stating that of 17 fast bowlers tested, the 6 fastest bowlers reported mean approach distances 2.14m longer than the 6 slowest bowlers, although as no set distances were reported it is difficult to compare with other findings. However, as highlighted above a generic optimal run-up distance is not achievable due to the physical and technical variations seen in fast bowlers, thus run-up speed may be a more indicative measure of how pre-delivery characteristics may affect fast bowling.

2.4.2.2 Run-up Speed

Although the importance of correct run-up distance is emphasised, studies have placed more emphasis on analysing run-up speeds (Elliott et al. 1992; Elliott et al. 1993; Burnett et al. 1995). Run-up speeds have been shown to vary depending on which bowling action is being performed. A front-on action typically displays higher run up speeds than a side-on action, as a side-on action requires more time during the pre-delivery stride to re-orientate the shoulders and hips (Elliott and Foster, 1984; Bartlett et al., 1996). Consequently, the impact of run-up speed may provide some insight into which bowling technique may allow higher ball release speeds. Run-up speeds have been well documented and can be seen displayed in table 2.4.2. However, most studies have only taken measurements at the beginning of the pre-delivery stride (Elliott et al. 1986; Elliott et al., 1992). From this data we cannot make any assumptions regarding

the effect of run-up distance on run-up speed. Mason and colleagues (1989) suggests that peak running velocity is reached approximately 8-16m before the crease, as bowlers typically slow down in preparation for delivery. Consequently, it may be argued that a shorter run-up may be advantageous if a constant running velocity was to be maintained, conserving the bowlers' energy, whilst still reaching the same run-up speed at the pre-delivery stride.

Research has attempted to quantify the effectiveness of run-up distance and speed via correlation with a performance measure (Worthington et al. 2013), typically ball release speed. However, the degree to which the speed of run-up contributes to ball release speed may vary between bowlers, due to the wide variability in bowling actions (Elliott and Foster, 1984). As a result, correlating ball release speed with run-up speed becomes difficult. Studies have attempted to calculate percentage contribution of run-up speed using the following calculation seen in equation 2.4.1 (Davis and Blanksby, 1976; Elliott et al. 1986):

$$\left(\frac{Ball\ Release\ Speed - Bowlers\ Mass\ Centre\ Speed}{Final\ Ball\ Speed} \right) \times 100 = Percentage\ Contribution$$

Equation 2.4.1. Calculation for percentage contribution of run-up speed to ball release speed.

However, this equation assumes that no other variables are involved, which as stated above, is not the case. Consequently, this equation may be regarded as too simplistic. Studies have also aimed to establish a relationship between run-up speed and ball release speed via manipulation of run-up speed (Brees, 1989). Brees (1989) concluded that there was a positive correlation between run-up speed and ball release speed but the correlation between run-up speed and accuracy was negative. These changes have been attributed to the changes in bowling kinematics observed (Brees, 1989). An increase in run-up speed was seen to be related to decreased trunk lateral flexion and flexion, whilst an increase in knee flexion was observed. However, these kinematic changes may not be representative of the kinematics displayed during match situations, as the testing protocol required trained bowlers to perform a bowling action that they are not accustomed to, such as an excessively slow or fast run-up. Although it has been acknowledged that run-up speed is conceptually likely to have some contribution to ball release speed; due to the lack of substantiated evidence produced in the current

literature, the relationship between run-up speed and the remaining bowling action is unclear.

2.4.2.3 Pre-delivery Stride

The pre-delivery stride has been defined by the Marylebone Cricket Club as the stride that separates the run up from the delivery stride. It consists of a jump off of the leading leg, allowing enough time to orientate the body for the rear-foot impact in the delivery stride (M.C.C., 1976). Few studies have reported the impact of the pre-delivery stride on the kinematics during the delivery stride; however, it may be assumed that the length of the pre-delivery stride will vary depending on the bowling action being performed. A side-on bowler may need more time to rotate 180º for back-foot impact (BFI) compared to a front-on bowler who does not need to rotate to enter the delivery stride (Bartlett et al. 1996). Davis and Blanksby (1976) reported that faster bowlers' delivery stride was 22% longer than the previous run-up stride. Slower bowlers were reported to have only increased stride length by 5% in the pre-delivery stride. This may be explained by the increased need to decelerate, to facilitate an effective bowling action (Davis and Blanksby, 1976).

2.4.3 Delivery Stride

BFI is commonly used as a marker to signify the start of the delivery stride, with the end of front-foot impact (FFI) marking the completion of the delivery stride (Figure 2.4.3) (M.C.C., 1987). Most biomechanical research has focussed on this stage of the bowling action, as highest impact forces are experienced at this point, as well as kinematic variables during this point being highlighted as displaying the greatest contribution to performance and injury risk (Crewe et al. 2013; Bayne et al. 2016; King et al. 2016; Middleton et al. 2016).
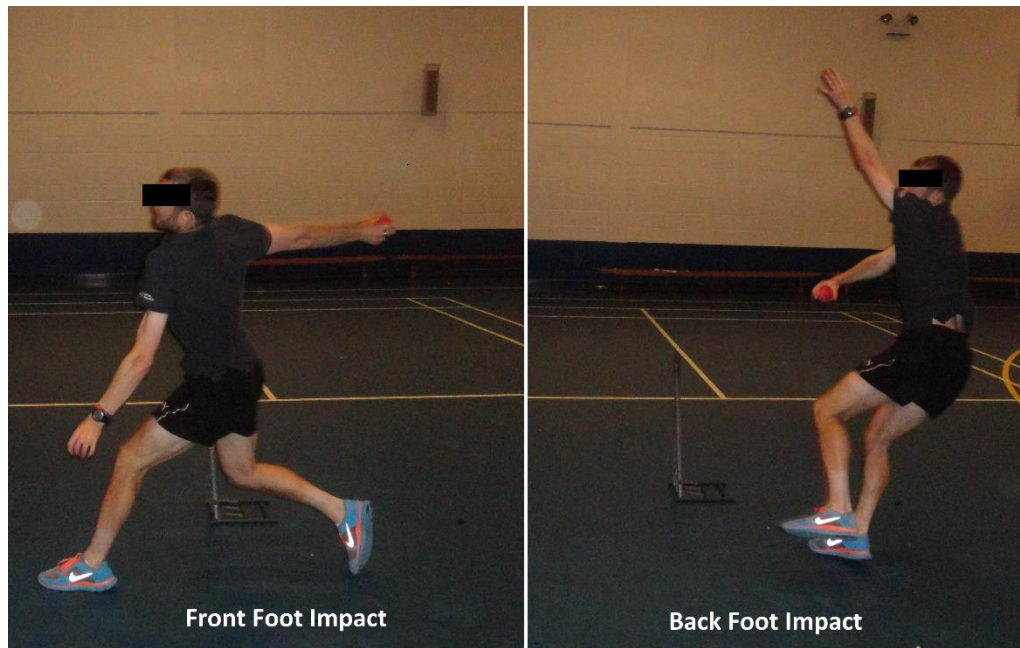
Figure 2.4.3. Defining the delivery stride using back and front foot impact.

2.4.3.1 Stride length and alignment

Stride length of the delivery stride has been reported, but not emphasised as a major contributing factor in relation to injury or performance (Stockill, 1994). Stride length is commonly reported as a percentage of bowlers height in order to standardise the measurement (Burnett et al. 1995). Elliott and colleagues (1992) provide a recommended stride length of approximately 75-85% of height. An appropriate stride length is necessary to maintain a balanced action and enable the speed generated from the run-up to be effectively controlled (Elliott and Foster, 1984). Consequently, it has been reported that an increased approach speed typically results in a larger delivery stride, however, if approach speed is too great, it will lead to a shortened and 'unbalanced' delivery stride (Elliott et al. 1986). Nevertheless, few studies have reported these effects in any detail so no clear trend can be established.

2.4.3.2 Lower limb kinematics

Whilst less attention has been focussed on stride length and alignment, lower limb kinematics such as knee and ankle kinematics have been reported to play a major role in regulating magnitude of impacts and ball release speed (Nigg and Liu, 1999; Crewe et al. 2013). An extended leg has been reported to produce higher ball release speeds (Elliott et al. 1986). This is due to the increased stability of the lower body, producing a more effective lever to which the upper body can pivot around, consequently increasing

ball-release speed (Elliott and Foster, 1984). Although the 'straight leg' technique has been shown to maximise ball release speed, assuming that other kinematic variables remain the same, the risk of injury is likely to be increased as a result of the diminished capacity for impact attenuation (Hurrion et al. 2000). Although studies have shown that there is an increase in impact forces displayed when the straight leg technique is used, due to few studies reporting impacts further up the body, there is little conclusive evidence to support the claim that this also results in decreased shock attenuation further up the kinetic chain.

Research has also outlined the effects of front-foot impact when the leading leg displays the 'flexed knee' technique (Elliott et al. 1986). Impact forces displayed upon front-foot impact were reported as significantly smaller than bowlers bowling with an extended knee (Crewe et al. 2013). However, as a result of a bent knee, ball release speed is compromised (Portus et al. 2004). Therefore, research has recommended a third technique in regards to knee flexion: an action that displays knee flexion upon impact, then proceeds to extend to straight for ball release has been deemed advantageous as peak impact force is reduced, whilst the effectiveness of a straight leg lever is not completely lost, resulting in higher ball release speed with less risk of injury (Stockill and Bartlett, 1992). However, the above assumptions on the relationship between front leg technique and GRF has been questioned in a recent study highlighting that bowlers displaying higher GRF also displayed greater knee flexion (possibly as a mechanism of impact attenuation) (Worthington et al. 2013). These differences may be a result of older studies analysing knee flexion at lower sampling frequencies, in a single plane of motion thus introducing the possibility of aliasing or parallax errors. This ambiguity, highlights that more research is needed looking into mechanisms of impact attenuation in order to understand its relationship with injury risk and performance.

The disparity between more recent studies and the body of more dated two-dimensional analyses has also been highlighted in the reporting of the effect of ankle angles on performance and injury risk. Older studies have typically not reported ankle angle (possibly due to the difficulty in defining different time points with lower sampling frequencies), whilst more recent studies have highlight the ankle as playing a significant role in mediating GRF (Worthington et al. 2013). Bowlers who first landed on their heel at FFI and then consequently displayed greater plantar flexion when the foot was flat on the floor experienced smaller impacts and slower time-to-peak values (Worthington et

al. 2013). This may suggest that a technique that allows scope for greater range of motion is more effective at being able to dissipate impacts. However, as addressed later in this review the opposite has been suggested for impacts experienced at the spine, where larger ranges of motion may place the spine in an increased position of weakness (Burnett et al. 2008; Glazier et al. 2010). Therefore, it may be necessary to see if similar restrictions exist at the knee and ankle, or whether they are better equipped to handle large impacts at end range when compared with the spine. This relationship has not been reported in previous fast bowling literature.

2.4.3.3 Shoulder and Hip Alignment

Further uncertainty exists with regards to the effect of kinematics further up the body on impact attenuation. Studies utilising two-dimensional video analysis have had difficulty effectively analysing hip and spinal kinematics due to their three-dimensional nature during fast bowling (Foster et al. 1989; Burnett et al. 1995; Hurrion et al. 2000; Portus et al. 2004; Elliott et al. 2005). Thus, studies reporting these kinematics are limited to more recent three-dimensional studies, some of which are only interested in their relationship with performance (Ranson et al. 2008; Ranson et al. 2009; Zhang et al. 2011; Crewe et al. 2013; Bayne et al. 2016; King et al. 2016; Middleton et al. 2016). Research typically agrees that a position between front-on and side-on is generally adopted by the majority of fast bowlers, at BFI (Ranson, 2009). Shoulder alignment angles have been reported between 195º-254º where a completely side-on orientation is 180º (Elliott and Foster, 1984; Foster and Elliott, 1985; Ranson, 2009). Hip alignment, has been underreported at this phase due to difficulty in accurately assessing this using two-dimensional methods. However, it has been reported that hip alignment also tends to remain predominantly 'front-on' (Bartlett et al. 1996; Johnson et al. 2012). It can therefore be hypothesised that less hip rotation is involved in a front-on action as the shoulders and hips are relatively close to the desired alignment at rear foot impact, suggesting that other factors may have a greater contribution to ball release speed in front-on bowling, as opposed to a side-on action where greater angular velocity is generated by the body's rotation (Stockill and Bartlett, 1992). Stockill and Bartlett (1992) stated that ball release speed was not affected by bowling action, suggesting that other factors, such as spinal hyperextension and increased run-up speed, have a higher contribution in front-on techniques to minimise the effects of less angular momentum. However, as mechanisms of injury continue to remain unclear, there is not enough

evidence in the literature to substantiate the impact of increased ball release speed on risk of injury across different bowling techniques without further investigation.

Mean shoulder alignment angles have been reported between 198º-217º at FFI (Elliott and Foster, 1984; Elliott et al. 1986). Some studies report an increased alignment from BFI, contradicting others that display decreased shoulder alignment angle, emphasising the individual nature of bowling technique and thus the difficulty with between cohort comparisons (Elliott and Foster, 1984; Burnett et al. 1995). There has been few studies that have specifically looked at the relationship of shoulder alignment at BFI compared with FFI, none of which have commented on its effect on ball release speed (Stockill and Bartlett, 1992). However, this relationship may be of key importance when looking at lower back injury and pain (Ranson et al. 2008).

Shoulder counter-rotation taken from alignment at BFI and during the delivery stride has been reported by a multitude of studies. However, it must be acknowledged that methods to obtain these results have varied with some studies classifying it by reporting minimum shoulder alignment angle, whilst others use back-foot and front-foot impacts as landmarks; others singularly report angles at front-foot impact (Burnett et al. 1995). Consequently, it is likely both over and underestimations of the true counter-rotation angle have been reported. Foster et al. (1989) reports a guideline for excessive shoulder counter-rotation: shoulder counter-rotation in excess of 40º is deemed to increase the risk of lower back injury, whilst other studies suggest 30° may be the critical value (Portus et al. 2004). Foster et al. (1989) states of 17 bowlers with excessive shoulder counter-rotation 6 displayed stress fractures and 7 contracted soft tissue injuries. Research highlights that excessive shoulder counter-rotation is most common with bowlers who bowl using the mixed action (Johnson et al. 2012). This is due to the combination of a 'front-on' and 'side-on' action; as this action is less linear, mechanically it is more reliant on a higher trunk rotation to generate a higher ball release speed. However, this measure still fails to describe three-dimensional kinematics and consequently how representative shoulder counter-rotation is of true spinal kinematics is currently unknown. Further analysis of this relationship would allow coaches and researchers to understand what shoulder counter-rotation values represent in terms of a possible mechanism of injury.

2.4.3.4 Ball Release

Ball release typically occurs after initial contact of FFI, often used as the point to signify the end of the delivery stride and beginning of the follow-through (MCC, 1976). However, this point has only been used by some studies resulting in heterogeneity in reported results with research reporting ranges of motion between BFI to ball release are likely to report larger values compared with studies reporting values between BFI to FFI or FFI to ball release. Whilst all these time points provide different insights into performance and injury, inconsistencies in reporting limit the ability to pool data between studies and thus gain a deeper understanding of the fast bowling population as a whole.

Ball release has been reported less frequently in relation to injury as it typically occurs after peak impact and consequently kinematic differences at this point may be considered 'less risky' than kinematics during peak impacts typically seen at FFI (Worthington et al. 2013). However, variables such as height at ball release and kinematics at this time are considered more influential in terms of the effect on ball release speed (Bartlett et al. 1996). Worthington and colleagues (2013) highlight that greater upper trunk flexion through to ball release and delayed onset of rotation towards the batsmen is seen in the fastest bowlers and may therefore be key to coaching interventions. Whilst this may be advantageous from a performance perspective, Crewe et al. (2013) suggests that increased ball release speed, along with higher shoulder counter-rotation increases lumbar loads and consequently risk of injury. However, as mentioned previously, reporting of shoulder counter-rotation does little to inform of mechanisms of spinal injuries. Thus, it may be the case that variables highlighted above to increase ball release speed may be the cause of the increased lumbar load seen in faster bowlers. Bayne et al. (2016) has reported larger lumbar flexion and lateral flexion joint moments following FFI may increase risk of injury. This corroborates the theory that larger upper trunk flexion seen in faster bowlers may also increase risk of lower back injury, however this does not link delayed rotation to increase injury risk. Consequently, flexion following FFI may increase ball release speed but also lower back injury risk, however this is speculative and requires further corroboration.

Table 2.4.2. Fast bowling kinematics (mean± SD) reported in previous studies.

Authors	Participants (n)	Run Up Speed (ms^{-1})	Shoulder Alignment Angle (^{o})			Knee Angle at Front-Foot Impact (^{o})	Delivery Stride Length (m or % height)	Delivery Stride Alignment (m)	Ball Release Speed (ms^{-1})
			Back-Foot Impact	Front-Foot Impact	Ball Release				
Crewe et al. 2013	40 juniors (mean age 16.2 yrs)	5.8± 0.7	230±15.6			166± 6.8			29.9±2.8
Elliott et al. 2005	14 (<11 yrs), 11 (<13 yrs), 12 (<15's yrs)					142.3 134.3 125.2			20.0 21.3 23.5
Portus et al. 2004	42 seniors (high performance)					164± 11			34.7± 2.0
Burnett et al. 1995	9 seniors (elite)	5.5±0.5	223± 22	204± 11	301± 13		87± 8%	-0.02± 0.19	32.2± 1.9
Elliot et al. 1993	24 juniors (mean 13.7 yrs)	4.5± 0.6 4.6± 0.6	213± 12 219± 14	209± 18 199± 8	319± 19 312± 12		83± 10% 88± 10%		24.4± 2.1 24.8± 1.4
Stockhill and Bartlett, 1992	17 seniors (elite)	6.8± 1.7	223± 17	209± 10	291± 15				37.4± 1.87
Elliott et al. 1992	20 juniors (mean age 16.8 yrs)	5.1± 0.9	206± 32	201± 16	309		86± 9%	-0.109±0.013	31.7± 1.6
Saunders and Coleman, 1991	7 seniors								
Burden and Bartlett, 1989	10 college players	6.0± 0.6							
Mason et al. 1989	15 juniors	5.6					1.3m		
Foster et al. 1989	82 juniors (mean age 16.8yrs)	5.0± 1.4	219± 21	203± 16					
Elliott et al. 1986	15 seniors (elite)	4.6± 0.8	232± 18	217± 8	300± 8		1.29± 0.1m	0.0032± 0.017	30.6± 2.0
Elliot and Foster, 1985	1 senior (elite)	5.4	200		300		1.37		34.8
Elliott and Foster, 1984	4 Seniors (elite)	4.3± 0.3	195± 11	198± 9	299± 4		1.54± 0.12	-0.014± 0.028	36.3± 1.7
Penrose et al. 1976	6 seniors (elite)	7.5± 1.8							40.0± 2.76

n, number; yrs, years; ms^{-1}, meters per second, $^{\circ}$, degrees; m, meters.

2.4.4 Spinal Kinematics

Spinal kinematics have received much focus in the literature due to the high prevalence of spinal injury and pain in the fast bowling population (Burnett et al. 1998; Ranson et al. 2008; Ferdinands et al. 2009; Ranson et al. 2009; Stuelcken et al. 2010; Crewe et al. 2013). Typically, spinal kinematics have been reported between back-foot impact and front-foot impact, however kinematics at ball release have also been reported (as seen in table 2.4.3). Although, spinal orientation at back and front foot impact is important, it may also be valuable to report range of motion between these landmarks. This has been reported in some studies, however not as frequently as spinal orientations (Stuelcken et al. 2010; Crewe et al. 2013).

Shoulder counter-rotation (SCR) has received more attention than true three-dimensional spinal kinematics in current literature. Crewe et al. (2013) suggests that increased SCR is associated with an increase of medio-lateral force being placed on the lumbar spine. Excessive SCR is most common with bowlers who use the mixed action (Portus et al. 2004; Johnson et al. 2012). However, SCR angles provide limited information regarding spinal kinematics as orientation of the pelvis is not considered, therefore the measurement represents the change in shoulder alignment relative to the wicket; which could be created by spinal rotation or whole body rotation (Chosa et al. 2004). However, disagreement exists in which variables are important, with studies opting to report differing kinematic variables. Previous reviews have attempted to synthesise kinematic information; however, this was limited to shoulder alignment angles which provides little information on the resultant behaviour of the spine in three-dimensions (Bartlett et al. 1996).

Burnett and colleagues (1998) have reported a non-significant trend towards greater contralateral lumbar side-flexion, with the mixed action. Furthermore, contralateral side-flexion and ipsilateral rotation have been reported to occur alongside spinal flexion at front-foot impact, however it is also reported that individual technique can vary significantly (Burnett et al. 1998; Glazier, 2010).

2.4.4.1 Shoulder Counter-rotation and Hip-shoulder Separation Angle

Shoulder alignment angle at back-foot impact has been used by researchers to determine degrees of shoulder counter-rotation (SCR) (Portus et al. 2004). SCR has been shown to vary depending on the bowling action; the mixed action has been reported as showing

the greatest peak counter-rotation angles. SCR has been used in older studies to quantify spinal rotation, however, as this value does not take pelvis orientation into account, spinal kinematics are not represented (Burnett et al. 1995; Glazier et al. 2000; Hurrion et al. 2000; Portus et al. 2004; Elliott et al. 2005). Despite this, recent studies continue to report SCR angles, although hip-shoulder separation angle (reported in some recent studies) may be more representative of overall spinal rotation (Burnett et al. 1995; Glazier et al. 2000; Portus et al. 2004). Mean shoulder counter-rotation angles have been reported between 19-45° in comparison to hip-shoulder separation angles of around 16-18°, highlighting that in some cases, spinal rotation contributes less than half the total rotation seen in shoulder counter-rotation (Burnett et al. 1995; Portus et al. 2004). Consequently, this questions the viability of using shoulder counter-rotation as a measure of spinal kinematics.

Table 2.4.3. Mean (±SD) spinal kinematics, shoulder counter-rotation and data collection methods reported in recent fast bowling studies.

Authors	Participants (n)	Segment Analysed	Phase Analysed	Flexion (°)	Extension (°)	L. Lateral Flexion (°)	L. Rotation (°)
Crewe et al. 2013b	13 18 8	S1-L1 (Lumbar)	FFI - BR			10.2± 3.6 11.8± 3.4 11± 2.9	
Stuelcken et al. 2010	14 senior (LBP) 12 senior (No LBP)	S1-T1 (Lumbo-Thoracic)	BFI- BR	27.2± 12.1 29.4± 10.5	14.2± 9.1 12.5± 8.6	41.9± 5.8 38.4± 6.3	25.6± 6.1 26.8± 5.6
Ferdinands et al. 2009	21 senior	S2-T10 (Lumbar)		38.2± 8	5.5± 2	15.7± 11.3	19.4± 2.4
Ranson et al. 2009	14 junior	S1- T10 (Lumbar)			0± 7	34± 7	29± 9
Ranson et al. 2008	50 senior	S1- T10 (Lumbar)			9± 6	34± 7	32± 8
Burnett et al. 1998	20 senior	S2- L1 (Lumbar)		48.3	10.1	30.25	10.6

LBP, lower back pain; BFI, back-foot impact; FFI, front-foot impact; BR, ball release; L, left.

2.4.5 Summary

This review has highlighted that fast bowling kinematics' relationship to performance variables such as ball release speed is relatively well understood, despite significant variations in bowlers' technique. However, bowling kinematics' relationship with risk of lower back injury and pain is not as clearly understood. Whilst excessive shoulder

counter-rotation has been significantly linked to increased risk, this does not aid the understanding of three-dimensional spinal pathokinematics. Furthermore, the relationship between lower limb impacts and knee flexion has, in recent studies, been questioned. Differences in study methodologies have made it difficult to pool data, resulting in no other significant findings. Studies have hypothesised that larger ranges of lateral flexion and rotation, coupled with large impacts may increase risk as a result of the spine being in a position of mechanical weakness. In order to highlight areas that may aid the understanding of this issue, the current body of literature must be appraised to highlight limitations that may be responsible for the current lack of clarity and thus provide clear recommendations for further work.

2.5 How do fast bowling biomechanics affect risk of lower back injury and pain in cricket?

2.5.1 Introduction

The previous reviews in this chapter have provided an in-depth analysis of fast bowling pathologies and pain, bowling kinetics and kinematics separately. These reviews have highlighted that the current understanding of the relationship between fast bowling biomechanics and lower back injury and pain is ambiguous, despite receiving much attention in the current literature. This review aims to amalgamate and synthesise evidence to address the issue of fast bowling lower back pain by appraising current methodologies and therefore provide recommendations for future studies.

Prevalence of musculoskeletal injury in the cricket fast bowling population is significantly higher than for all other players (Orchard et al. 2006; Johnson et al. 2012). Observation over six seasons demonstrated that 51% of all injuries were sustained by bowlers compared to 24% for fielders, 23% for batsmen and 2% for wicket keepers (Orchard et al. 2002). Fast bowlers missed 14% of games due to injury, whilst spin bowlers only missed 4% with lower back injury resulting in the longest time absent from cricket (Stretch, 1992; Orchard et al. 2002). Therefore it is clear that this is a population that faces great risk of injury, however injury may include demonstrable pathological change (such as spondyloses) or pain. Epidemiological studies have suggested a relationship between cricket fast bowling and spinal pathological change (Ranson et al. 2005). The prevalence of bony abnormalities as a whole has been estimated at 24-55% in fast bowlers compared to just 6-7% in the general male population (Elliott et al. 1992; Hardcastle et al. 1992; Engstrom and Walker, 2007). Spondylolysis (stress fracture of the pars) and spondylolisthesis (fracture of the pars with anterior translation of one vertebrae on another) are the most common bony pathological changes and junior fast bowlers at the greatest risk with estimated prevalence figures around 67%, possibly due to immature bony structures at this age (Hardcastle et al. 1992; Hardcastle, 1993). In addition to these bony changes, a prevalence of lumbar disc degeneration at 35% has been reported in fast bowlers (Crewe et al. 2012; Morton et al. 2013). Once again adolescent fast bowlers appear at particular risk where prevalence of 44% was noted in 13 to 18 year olds (Elliott and Khangure, 2002). In one sample of junior bowlers (16-18 years old), 23% displayed

grade 1 disc degeneration; 13% grade 2 and 28% grade 3 (Hardcastle et al. 1992). Perhaps due to the progressive nature of disc degeneration, prevalence of 70% have been observed in retired fast bowlers (Annear et al. 1992). Therefore it is clear that fast bowlers are more likely to display pathological change in their lumbar spine.

Despite the prevalence of pathological change, it should be acknowledged that spinal abnormalities may exist without the sensation of pain, therefore lower back pain (LBP) should be investigated as a separate entity (Millson et al. 2004). LBP has been reported to result in 247 missed games in Australian domestic and international matches between 1995 and 2001 (Orchard et al. 2002). Indeed, almost a quarter (22.4%) of all playing time missed was attributable to LBP in fast bowlers (Orchard et al. 2002). Studies investigating LBP associated with fast bowling have reported prevalence between 40-64% (Elliott et al. 1992; Hardcastle et al. 1992; Dennis et al. 2005; Kountouris et al. 2012).

Previous literature has suggested that repeated exposure to high magnitudes of ground reaction force (GRF), in conjunction with combined spinal motions may be a significant factor in the pathomechanics of LBP and injury in fast bowlers (Bartlett et al. 1996; Hurrion et al. 2000; Crewe et al. 2013). Despite this, there are no systematic reviews investigating the literature pertaining ground reaction force. Previous reviews have explored intrinsic and extrinsic risk factors for developing LBP in fast bowlers, however these reviews offer little critique of the biomechanical measurement methods used to obtain the data resulting in a faithful representation of the original results (Morton et al. 2013; Oliver et al. 2016). Therefore, little insight into potential bias or flaws in methodological design was gained. The aim of this review was to critically analyse and synthesise the cricket fast bowling literature pertaining to ground reaction force and spinal kinematics during fast bowling to offer new insights into methods and conclusions relating these aspects to back pain and or pathology.

2.5.2 Evidence Acquisition

2.5.2.1 Search Strategy

The following electronic databases were systematically searched during April 2016; MEDLINE (1946-04/2016), Google Scholar, SPORTDiscuss (1985-04/2016), Science Citation Index (1900-04/2016), OAIster (2002-04/2016), CINAHL (1937-04/2016),

Academic Search Complete (1887-04/2016), Science Direct (1872-04/2016) and Scopus (1841-04/2016). The following terms were used during the search employing Boolean search operators where appropriate; biomechanic*, kinematic*, cricket*, fast bowler*, bowl*, lumbar, back, spine, injur*, sport* injur*. Reference lists of relevant articles were also searched to identify additional literature. A PRISMA diagram illustrates the retrieval process (Figure. 2.5.1).

2.5.2.2 Inclusion Criteria

To be included in this review, articles needed to investigate spinal kinematics or GRF during cricket fast bowling. All standards of cricket were considered, as were all ages and genders. Articles had to be in the English language as no funds for translation were available. Material from magazines and editorials were excluded in order to target only peer-reviewed information. Articles reporting just the shoulder alignment were excluded as this provides no insight into spinal kinematics.

2.5.2.3 Data extraction and study appraisal

Articles were initially screened by the principal investigator using title and abstract information. Any doubt over the relevance resulted in retrieval and review of the full-text and resolution achieved through consensus with additional authors. A review of the methodological quality of the studies was completed by the principal investigator using the standardised critical review form and guidelines from Law et al. (1998) as a template. This form was modified by the removal of 'intervention' due to the question of this review and nature of the studies investigated, as well as the inclusion of a mark each for sample bias, measurement bias and performance bias. This resulted in a checklist of thirteen items. These study appraisal checklists can be seen in table 2.5.1 and 2.5.2. Studies were separated into the main topics of GRF, spinal kinematics and injury with synthesis of results completed using odds ratios using MedCalc (V15.2) using a random effects model, where the relevant groups needed for the calculation are presented in the literature. Additionally, weighted averages were calculated as described in Equation 2.5.1 below. Weighted averages enabled data pooling after consideration of sample size.

Weighted Average GRF = $\dfrac{(GRF_1 \times N_1) + (GRF_2 \times N_2) + (GRF_3 \times N_3) + \ldots}{N_1 + N_2 + N_3 + \ldots}$

Where GRF_1= Reported GRF for 'study 1' and N_1= Sample size of 'study 1'

Equation 2.5.1. Calculation of weighted average GRF from extracted studies' results.

PRISMA 2009 Flow Diagram

Records identified through
database searching
(n = 252)

Additional records identified
through other sources
(n = 0)

Records after duplicates removed
(n = 216)

Records screened
(n = 216)

Records excluded
(n = 76)

Full-text articles assessed
for eligibility
(n = 140)

Full-text articles excluded,
with reasons
(n = 84)

Studies included in
qualitative synthesis
(n = 56)

Studies included in
quantitative synthesis
(meta-analysis)
(n = 16)

Identification

Screening

Eligibility

Included

Figure 2.5.1. PRISMA 2009 flow diagram of study retrieval process.

2.5.3 Evidence Synthesis

The systematic search resulted in 140 relevant articles which were reduced to 56 following application of the inclusion criteria. This was comprised of 16 focusing on GRF, 17 on spinal kinematics and 34 on injury. Some studies were included in more than one section.

2.5.3.1 Ground reaction force

Quality appraisal of GRF studies can be seen in table 2.5.1. Studies reporting GRF generally share common methodologies and consequently share similar threats to validity. This is evident in the repeated lack of reporting of detailed sample characteristics with inadequate description of sampling methods, making the determination of selection bias difficult. None of the studies reported a bowling history and therefore it was unclear how long individuals had been bowling or at what level. Justification of sample sizes was only reported in one study reviewed (Middleton et al. 2016). Statistical sample size calculations may offer some reassurance regarding the power of the study, however even a more pragmatic justification of sample size was missing. In light of these issues, the degree to which these results are representative of the fast bowling population is unclear. Moreover, few studies have reported actual p-values making the interpretation of significance due to chance difficult. Studies also scored poorly for bias relating to testing environment. Due to the nature of the studies, a typical laboratory based environment was utilised. It is possible that such an environment may affect bowling style due to physical constraints, such as run-up space, targeting the force plate or awareness of an 'unfamiliar' environment. It is not clear to what extent these factors affect GRF, however such factors could be considered limitations to the reviewed studies.

Despite previous research hypothesising a link between GRF at front-foot impact and risk of LBP and injury in fast bowlers (Portus et al. 2004; Crewe et al. 2012), only one study recorded GRF and LBP (Stuelcken et al. 2010). However, this was in female fast bowlers so it's comparisons to the main body of fast bowling literature may be limited. Over one third of studies reported GRF alongside lower back injury data (Elliott et al. 1992; Elliott et al. 1993; Portus et al. 2004; Bayne et al. 2016) and no results have reported a relationship between GRF and back pain/injury (Elliott et al. 1989; Elliott et al. 1992; Elliott et al. 1993; Portus et al. 2004; Bayne et al. 2016). All studies reported

GRF normalised to body weight, enabling the calculation of weighted averages as a method of data synthesis. A total of 378 bowlers resulted in a weighted average (±SD) vertical GRF of 5.8 (1.3) BW and horizontal braking GRF of 3.2 (1.1) BW. Time to peak GRF was also synthesised for 82 bowlers, with a weighted average of 49 (4) ms vertically and 55 (6) ms horizontal braking.

Table 2.5.1. Quality appraisal of fast bowling ground reaction force studies (based on Law et al. 1998)

Author	Citation	Purpose	Literature	Design	Appropriateness of design	Bias: Sample Measurement Performance	Sample	Outcomes	Interventions	Results	Drop outs	Clinical implications	Score (/13)
Elliott and Foster, 1984	Yes	Yes	Yes	(No) Not identified	Yes	(No) Sample: No detail on recruitment. No bowling history. (Yes) measurement (No) Performance: Lab based.	(No) No demographics. No sample size calc. No consent mentioned.	Yes	NA	(No) No statistical values reported	Yes	Yes	8
Foster and Elliott, 1985	No	(No) Not identified	Yes	(No) Not identified	(No) Not identified	(No) Sample: profile of 1 bowler. No bowling history. (No) measurement: not specified. (No) Performance: Not specified.	(No) No consent mentioned.	(No) Not described	NA	(No) No statistical values reported	Yes	Yes	3
Elliott et al. 1986	Yes	Yes	No	(No) Not identified	Yes	(No) Sample: No detail on recruitment. No bowling history. (Yes) measurement (No) Performance: Lab based.	(No) No demographics. No sample size calc. No consent mentioned.	Yes	NA	No No statistical values reported	Yes	Yes	7
Foster et al. 1989	Yes	Yes	Yes	Yes	Yes	(No) Sample: No bowling history (Yes) Measurement (No) Performance: Lab based	(No) No sample size calc. No consent mentioned.	(No) GRF, no detailed description of procedure	NA	(No) Missing p-values, or f-values.	Yes	Yes	8
Mason et al. 1989	Yes	No	Yes	(No) Not identified	Yes	(No) Sample: No bowling history. (Yes) measurement (No) performance: Lab based	(no) No demographics. No sample size calc.	Yes GRF clearly described.	NA	No actual stats presented	Yes No mention	yes	7
Elliott et al. 1992	Yes	Yes	Yes	(No) Not identified.	Yes	(No) Sample: No detail on recruitment. No bowling history. (Yes) Measurement (No) Performance: Lab based	(No) No sample size calc.	(No) Not clear	NA	(No) No actual stats presented	Yes	Yes	7
Elliott et al. 1993	Yes	Yes	Yes	(No) Not identified	Yes	(No) Sample: sample selected by coaches. No bowling history. (Yes) Measurement (No) Performance: Lab based	(No) No sample size calc.	Yes	NA	(No) No actual stats presented.	Yes	Yes	8

Hurrion et al. 2000	Yes	Yes	yes	(No) Not identified.	Yes	(No) Sample: No detail on recruitment. No bowling history. (Yes) Measurement (Yes)Performance: Outdoors (polyflex surface over Force Plate)	(No) No sample size calc.	Yes	NA	(No) No actual stats presented	Yes	(No) None mentioned.	8
Portus et al. 2004	(No) No full ref. available	Yes	Yes	Yes	Yes	(No)Sample: No recruitment detail. No bowling history. (Yes)Measurement (No)Performance: Not stated. Assumed lab based.	(No) No sample size calc.	(No) No detailed description of procedure	NA	Yes	Yes	Yes	8
Stuelken et al. 2009	Yes	Yes	Yes	(No) Not identified.	Yes	(No)Sample: No recruitment detail. No bowling history. (Yes)Measurement (No)Performance: Lab based (polyflex surface over Force Plate)	(No) No sample size calc.	Yes	NA	(No) No actual stats present.	Yes	No None mentioned.	7
Crewe et al. 2013	Yes	Yes	Yes	(No) Not identified.	Yes	(No)Sample: No bowling history. (Yes)Measurement: (No)Performance: Lab based.	(No) No sample size calc.	Yes	NA	(No) R and p values reported. Only as (p<0.05).	(No) 23 complete.	Yes	7
Worthington et al. 2013	Yes	Yes	Yes	(No) Not identified	Yes	(No)Sample: No bowling history. (Yes)Measurement (No)Performance: Lab based. (Artificial grass over Force Plate.)	(No) No sample size calc.	Yes	NA	Yes	Yes	yes	9
Spratford and Hicks, 2014	Yes	Yes	Yes	(No) Not identified	Yes	(No)Sample: No bowling history. (Yes) Measurement (No) Performance: Lab based.	(No) No sample size calc.	Yes	N/A	Yes	Yes	yes	9
Middleton et al. 2016	Yes	Yes	Yes	(No) Not identified	Yes	(No)Sample: No bowling history. (Yes) Measurement (Yes) Performance: Open Lab to allow adequate space	Yes	Yes	NA	Yes	Yes	No None mentioned	10
King et al. 2016	Yes	Yes	Yes	(No)	Yes	(No) Sample: No bowling	(No)	Yes	NA	Yes	Yes	Yes	9

				Not identified		history. (Yes) Measurement (No) Performance: Lab based.	No sample size calc. No consent mentioned.						
Bayne et al. 2016	Yes	Yes	Yes	(No) Not identified	Yes	(No) Sample: No bowling history. (Yes) Measurement (No) Performance: Lab based.	(No) No sample size calc.	Yes	NA	Yes	(No) 25 complete	Yes	8

2.5.3.2 Spinal Kinematics

Quality appraisal of kinematics studies can be seen in table 2.5.2. Studies measuring spinal kinematics share common methodologies and therefore share common threats to validity. A breakdown of the actual reported spinal kinematics and methodologies used have been outlined in table 2.5.3. The sample used was inadequately described in most studies (94%). Information regarding bowling history and low back pain history is not reported. Justification of sample size was not demonstrated in any of the studies analysed. All studies, with only two exceptions, were completed in a laboratory environment (Hurrion et al. 2000; Middleton et al. 2016). It was common to measure GRF and kinematics simultaneously which may result in the bowler targeting the force plate, the effect of which on spinal kinematics is not known. Further limitations to laboratory based studies have been outlined above. In addition, only 47% of studies reported actual p-values or statistical test values.

It has been hypothesised that specific spinal kinematics, may contribute to increased risk of LBP and injury (Johnson et al. 2012). Although three-dimensional spinal pathomechanics are still relatively unclear; studies have identified significantly greater range of lateral flexion in injury/ LBP groups, as well as greater SCR angles (Foster et al. 1989; Elliott et al. 1993; Portus et al. 2004). However, to date, there is little research correlating three-dimensional spinal kinematics with injury or LBP (Foster et al. 1989; Portus et al. 2004; Stuelcken et al. 2010).

Studies analysed in this review were subjected to odds ratio calculations where possible, in order to determine the effect of specific technique variables on prevalence of LBP and pathology in the fast bowling population. SCR of greater than 40° has been highlighted in research as significantly increasing risk of LBP and injury (Foster et al. 1989). Consequently, an odds ratio [95% CI Lower, Upper] defining the chance of fast bowlers displaying SCR >40° developing LBP compared with bowlers with <40° SCR was calculated at 0.2 [0.03, 1.1]. However, an odds ratio of 11.9 [3.0, 46.9] was determined for the chance of fast bowlers displaying SCR >40° developing lower back injury (Foster et al. 1989; Portus et al. 2004). This clearly displays the importance of separating lower back injury from LBP. Recent studies have hypothesised that high range of lateral flexion may also significantly increase risk of LBP and injury, however as no clear values for excessively high lateral flexion have been reported, no groupings

could be made and therefore odds ratio calculations cannot be conducted (Stuelcken et al. 2010; Crewe et al. 2012). Any further odds ratio calculations were also made difficult by the lack of 'non-fast bowler' data available from the studies analysed, as typically only a fast bowling sample was used.

Table 2.5.2. Quality appraisal of fast bowling spinal kinematics studies (based on Law et al. 1998)

Author	Citation	Purpose	Literature	Design	Appropriateness of design	Bias: Sample Measurement Performance	Sample	Outcomes	Interventions	Results	Drop outs	Clinical implications	Score (/13)
Foster et al. 1989	Yes	Yes	Yes	Yes	Yes	(No) Sample: No bowling history (Yes) Measurement (No) Performance: Lab based	(No) No sample size calc. No consent mentioned.	(No) GRF, no detailed description of procedure or reliability/validity	NA	(No) Missing p-values, or f-values.	Yes	Yes	8
Elliott et al. 1992	Yes	Yes	Yes	(No) Not identified.	Yes	(No) Sample: No detail on recruitment. No bowling history. (Yes) Measurement (No) Performance: Lab based	(No) No sample size calc.	(No) Not clear	NA	(No) No actual stats presented	Yes	Yes	7
Burnett *et al.* 1995	Yes	Yes	Yes	(No) Not identified	Yes	(No) Sample: No bowling history (Yes) Measurement (No)Performance: Lab based	(No) Some demographics given. No sample size calc.	Yes	NA	Yes	Yes	(No) No detailed implications	8
Burnett *et al.* 1998	Yes	Yes	Yes	(No) Not identified	Yes	(No)Sample: No bowling history. (Yes)Measurement (No) performance: Lab based	(No) No sample size calc.	Yes	NA	(No) No specific p values given	Yes	yes	8
Glazier *et al.* 2000	Yes	Yes	Yes	(No) Not identified.	Yes	(No)Sample: No bowling history (Yes)measurement (No) performance: Lab based.	(No) No sample size calc.	Yes	NA	Yes	Yes	No	8
Portus *et al.* 2000	Yes	Yes	Yes	(No) Not identified.	Yes	(No)Sample: No bowling history. (Yes)measurement (No)performance: Lab based	(No) Few demographics given. No sample size calc.	Yes	NA	Yes	Yes	No	8
Hurrion et al. 2000	Yes	Yes	yes	(No) Not identified.	Yes	(No) Sample: No detail on recruitment. No	(No) No sample size calc.	Yes	NA	(No) No actual stats	Yes	(No) None mentioned.	8

						bowling history. (Yes) Measurement (Yes)Performance: Outdoors (polyflex surface over Force Plate)				presented			
Elliot *et al.* 2005	Yes	Yes	Yes	(No) Not identified.	Yes	(No)Sample: No bowling history. (Yes)Measurement (No) Performance: Lab based.	(No) No sample size calc.	Yes	NA	Yes	Yes	Yes	9
Portus et al. 2004	(No) No full ref. available	Yes	Yes	Yes	Yes	(No)Sample: No recruitment detail. No bowling history. (Yes)Measurement (No)Performance: Not stated. Assumed lab based.	(No) No sample size calc.	(No) No detailed description of procedure	NA	Yes	Yes	Yes	8
Ranson *et al.* (2008)	Yes	Yes	Yes	(No) Not identified.	Yes	(No)Sample: No bowling history. (Yes)Measurement (No)Performance: Lab based.	(No) No sample size calc.	Yes	NA	Yes	Yes	Yes	9
Ranson *et al. (2009)*	Yes	Yes	Yes	(No) Not identified.	Yes	(No)Sample: No bowling history. (Yes)Measurement (No) Performance: Lab based.	(No) No sample size calc.	Yes	NA.	Yes	Yes	yes	9
Ferdinands *et al.* (2009)	Yes	Yes	Yes	(No) Not identified	Yes	(No)Sample: No bowling history. (Yes)Measurement (No)Performance: Lab based.	(No) No sample size calc.	Yes	NA	(No) No actual statistics reported	Yes	Yes	8
Stuelken et al. (2010)	Yes	Yes	Yes	(No) Not identified	Yes	(No)Sample: No bowling history. (Yes) Measurement (No) Performance: Lab based.	(No) No sample size calc.	Yes	NA	Yes	Yes	Yes	9
Zhang *et al.* (2011)	Yes	Yes	Yes	(No) Not	Yes	(No)Sample: no recruitment detail.	(No) No sample	Yes	NA	(No) No P	Yes	(No) None	7

			identified.		No bowling history. (Yes)Measurement (No)performance: Lab based	size calc.				values reported		mentioned	
Crewe *et al.* 2013b	Yes	Yes	Yes	(No) Not identified.	Yes	(No)Sample: No bowling history. (Yes)Measurement (No) performance: Lab based.	(No) No sample size calc.	Yes.	NA	(No) No R and R values reported.	Yes	yes	8
Crewe et al. 2013	Yes	Yes	Yes	(No) Not identified.	Yes	(No)Sample: No bowling history. (Yes)Measurement: (No)Performance: Lab based.	(No) No sample size calc.	Yes	NA	(No) R and p values reported. Only as (p<0.05).	(No) 23 completed.	Yes	7
Bayne et al. 2016	Yes	Yes	Yes	(No) Not identified	Yes	(No) Sample: No bowling history. (Yes) Measurement (No) Performance: Lab based.	(No) No sample size calc.	Yes	NA	Yes	(No) 25 complete	Yes	8

Table 2.5.3. Spinal kinematics, shoulder counter-rotation and data collection methods reported in recent fast bowling studies

Authors	Participants (n (mean age))	Methods	Sample Frequency (Hz)	Marker/ Sensor Placement	Spinal Kinematics (° (±SD))				Shoulder Counter-Rotation (° (±SD))
					Flexion	Extension	Lateral Flexion	Rotation	
Bayne et al. 2016	12 INJ (15.5yrs) 12 NON-INJ (16yrs)	12 cam MAS & FP	250	L1, L5, left & right of L4	20.2 (3.8) 21.2 (4.7)		10.5 (3.6) 10.6 (3.4)	4.1 (1.5) 4.8 (1.6)	35.7 (12.3) 32.5 (11.8)
Crewe et al., 2013	40 (16.2yrs)	12 cam MAS & FP.	250	ASIS, PSIS, L1, left & right of L4, L5.					35.1 (12.2)
Crewe et al., 2013b	39 (16.1yrs)	18 cam MAS & FP.	250	ASIS, PSIS, L1, left & right of L4, L5.			11 (3.3)		
Stuelcken et al., 2010	14 LBP 12 No LBP = 26 (22.5yrs)	8 cam MAS.	120	ASIS, PSIS, shoulder joint centres.	27.2 (12.1) 29.4 (10.5)	14.2 (9.1) 12.5 (8.6)	41.9 (5.8) 38.4 (6.3)	25.6 (6.1) 26.8 (5.6)	39.4 (3.2)
Ferdinands et al., 2009	21 (22.4yrs)	8 cam MAS & FP.	240	S2, T10.	38.2 (8)	5.5 (2)	15.7 (11.3)	19.4 (2.4) (Est. from %)	
Ranson et al., 2009	14 (18.5yrs)	18 cam MAS.	300	ASIS, PSIS, L1, T10.		0 (7)	34 (7)	29 (9)	45 (15)
Ranson et al., 2008	50 (23yrs)	12 cam MAS.	120	ASIS, PSIS, L1, T10.		9 (6)	34 (7)	32 (8)	41 (16)
Elliott et al., 2005	14 (<11yrs) 11 (11-13yrs) 12 (13-15yrs)	2 high speed cams.	200	PSIS, acromion processes.					23.2 21.8 Not reported.
Portus et al., 2004	42 (22.4yrs)	2 high speed cams & 2 FP.	100 & 150	Hip joint centres, shoulder joint centres.					30.75
Portus et al., 2000	14 (23.3yrs)	2 cams.	50	Acromion processes.					44 (15.8)
Glazier et al., 2000	9 (21yrs)	2 cams.	25	Unclear					28 (14)
Burnett et al., 1998	20 (19.1yrs)	ETD	120	S2, L1.	67.5	29	32.9	15.5	

| Burnett et al., 1995 | 9 (18.1yrs) | 3 high speed cams. | 100 | Mid ASIS-PSIS, acromion processes. | | 31 (16) |

Cam, camera; MAS, motion analysis system; FP, force plate; ETD, electromagnetic tracking device; ASIS, anterior superior iliac spine; PSIS, posterior superior iliac spine; SD, standard deviation.

2.5.4 Discussion

2.5.4.1 Fast Bowling and GRF

Peak vertical GRF at front-foot impact reported in fast bowling studies is considerably higher than those reported in running and jump landing (between 2-3 and 3-5 times body weight respectively) (Zadpoor and Nikooyan, 2011; Bates et al. 2013). Time to peak GRF is similar to running literature which reports time to peak around 45 ms, therefore fast bowlers may experience higher loading rates at front-foot impact compared with runners (Bennell et al. 2004). Despite this, the current literature suggests that front-foot impact in bowlers is either effectively attenuated up the body or remains below the injury threshold and thus appears to be unrelated to spinal injury risk. Spratford and Hicks (2014) support these conclusions, reporting increased knee flexion at higher magnitudes of GRF, in opposition to previous literature reporting higher GRF with an extended front knee (Elliott et al. 1986; Hurrion et al. 2000; Spratford and Hicks, 2014). However, whilst Bayne et al. (2016) have looked at lumbar loads using inverse dynamics, no other research has analysed the relationship between GRF and loads further up the body during fast bowling and thus the effects of these kinematic adjustments to higher GRF are still unclear.

2.5.4.2 GRF Research Methodologies

Most studies employed an experimental design which included laboratory based testing. The merits of such an environment mean many confounding factors can be controlled, such as wind and weather. However, it is unclear whether bowling in such an environment accurately reflects the bowling strategies used in 'live situations.' Completing bowling inside a confined space will likely enforce constraints on run-up, which can be long for fast bowlers. Furthermore, in order to land on the force plate, the bowler may 'target' their front-foot landing and while footfall constraints are considered a key aspect of fast bowling technique, psychological differences may still affect ecological validity and may therefore not be truly representative of 'live' bowling. Additionally, bowling in the laboratory environment is often based on the bowler aiming for specific targets, not actual stumps, the effect of this altered visual target on bowling strategy is not known.

Four studies chose to lay material over the force plate (polyflex surface and artificial grass), however no adjustments to the calculation of GRF were made (Hurrion et al.

2000; Stuelcken and Sinclair, 2009; Worthington et al. 2013; King et al. 2016;). The additional damping characteristics of the added surface are likely to have affected the actual GRF values measured. Moreover, this additional material has the potential for allowing movement between the foot and force plate further affecting the values reported.

The limitations of current studies reporting GRF could be overcome by more detailed reporting of the sample used, particularly gaining greater understanding of bowling history. Bowling experience has been reported to affect the technique used as well as influence the magnitude of GRF, therefore detailed reporting of the sample used is imperative (Middleton et al. 2016). The reporting of actual statistics values and p-values provides the reader with additional information regarding the confidence of the statistical results. All studies chose to use a force plate to measure the GRF. It is difficult to integrate such technology into live cricket testing as such a device either sits on the surface of the grass, providing a raised platform onto which the bowler must land or is sunk into the floor. This overcomes the issue with differing heights but defaces the pitch and is not portable. Moreover, the rigidity of the surface onto which the force plate sits significantly affects the GRF, requiring copious recalibration during differing conditions. In light of these limitations, a novel solution should be sought that allows for the measurement of front-foot kinetics in a non-defacing, simple and portable way.

2.5.4.3 Fast Bowling Kinematics

Three-dimensional spinal kinematic analyses in cricket is beginning to be reported (Ranson et al. 2008; Ferdinands et al. 2009; Ranson et al. 2009; Stuelcken et al. 2010). However, even with these more detailed measures of spinal kinematics during bowling, very few new insights into the impact of bowling technique on lower back injury and pain have been reported. SCR has repeatedly been reported as earlier studies have identified it as significantly increasing risk of lower back injury, when in excess of 40°, as commonly seen in 'mixed bowling actions' (Portus et al. 2004; Johnson et al. 2012). However, the ability of SCR to describe three-dimensional spinal motion is questionable as orientation of the pelvis is not considered. The measurement represents the change in shoulder alignment relative to the wicket which could be created by spinal rotation or whole body rotation. Burnett et al. (1998) have reported a non-significant trend towards greater contralateral lumbar side-flexion with a mixed bowling action

compared with other bowling actions (Burnett et al. 1998). Furthermore, contralateral side-flexion and rotation have been reported at front-foot impact, placing the spine in a position of relative weakness (Elliott, 2000; Portus et al. 2004). However, as no values of what may be considered 'excessive' have been established and no evidence in fast bowling literature has conclusively related these variables with increased risk of injury, the precise pathokinematics remain unclear (Chosa et al. 2004; Ranson et al. 2008).

2.5.5.4 Kinematic Research Methodologies

It was noted in the results section that kinematic studies shared common methodologies, however large heterogeneity existed in the actual measurement methods used. These fundamental differences prevented any data synthesis. Seven studies used a multi-camera optoelectronic motion analysis system (Elliott et al. 1993; Chosa et al. 2004; Ranson et al. 2008; Ferdinands et al. 2009; Crewe et al. 2013; Crewe et al. 2013b; King et al. 2016). Such systems allow for a wealth of kinematic information due to the freedom of multiple markers determining many body segments. Rapid sampling rates are achievable which is necessary for highly ballistic movements such as bowling, and are considered the current 'gold' standard. However, these methods are associated with excessive drops outs due to marker occlusion or marker loss due to sweating (Crewe et al. 2013). Furthermore, in order to use marker systems, the subject must be in a state of undress which may not be appropriate for all cricket fast bowlers. It is noted that only one study employed different technology to video based systems, namely an electromagnetic tracking device (Burnett et al. 1998). Such a device is commonplace for the measurement of three-dimensional spinal kinematics and has the distinct advantage of being portable (Van Herp et al. 2000; Ma et al. 2009). Despite this, the study was still conducted in the laboratory environment. Electromagnetic tracking devices have small operating ranges due to the limited magnetic field produced, which can be overcome (as in this study) by mounting the electromagnetic source on the person (Milne et al. 1996). However, whether wearing such a 'large sensor' (dimensions 56mm x 58mm x 56mm) interferes with the bowling technique is unclear. In addition, the possibility of the wires of such a device erroneously moving a sensor has been acknowledged (Burnett et al. 1998).

When analysing lumbar kinematics, it is necessary to define a body segment of interest which has varied in the previous literature. Earlier studies typically measure spinal

kinematics between shoulder and pelvis; thus, describing thoraco-lumbar range of motion (ROM) with the addition of shoulder girdle for studies using markers on the shoulders (Burnett et al. 1995; Portus et al. 2004; Elliott et al. 2005). Other studies have demarcated the spine to just a lumbar 'joint' between S2 and L1 (Burnett et al. 1998). Moreover, some studies have only reported shoulder counter-rotation, which only takes into account contralateral shoulder rotation in relation to minimum shoulder alignment without reference to the kinematics of the hips or pelvis. The absence of a pelvis frame of reference means the values may not represent actual spinal motion. Recent studies continue to report these values in favour of more traditional cardanic descriptions of ROM, making comparisons to the literature outside of cricket difficult.

In order to overcome the limitations associated with laboratory constraints, line of sight difficulties and issues of magnetic field sizes, new technologies and their application to cricket fast bowling should be explored. Rowlands et al. (2009) present a report on the use of inertial sensors within fast bowling practice sessions. Inertial sensors have been used in the clinical analysis of three-dimensional spinal ROM and therefore, may be able to overcome the limitations in current methods (Williams et al. 2013).

2.5.5.5 Practical Implications

This review has highlighted the large heterogeneity in reported kinematic results evident between studies, making it difficult for coaches and health practitioners to make informed decisions on any required interventions. This is also a limiting factor when trying to pool data from multiple studies, making meta-analysis difficult. SCR remains the only variable that significantly affects risk of lower back injury; however, this may be due to consistency in reporting of this value allowing data pooling and therefore analysis of a larger sample of fast bowlers. Whilst a useful and quick measurement for coaches, SCR still fails to describe three-dimensional spinal kinematics, thus, the exact mechanism of injury is still unclear. Nonetheless, until any further guidelines can be produced, coaches should continue to monitor SCR values with an aim to maintain SCR<40°.

Magnitude and time to peak GRF has shown no relationship with risk of lower back pain or injury. However, studies hypothesise that frequency of exposure to high GRF may increase risk of injury. This is in agreement with literature highlighting high bowling workloads as being associated with elevated risk. Consequently, it is advised

that coaches monitor fast bowler's training and match workloads. Fast bowlers should avoid bowling spells of greater than 10 overs to minimise risk of acute LBP, whilst bowling less than 50 overs or 2.5 days a week may decrease risk of chronic LBP and injury. Furthermore, a dramatic increase in bowling workload should be avoided (Foster et al. 1989; Dennis et al. 2005).

2.5.5 Summary

This review has provided a contemporary, systematic analysis of the current literature investigating spinal kinematics and GRF during fast bowling in cricket, as well as identifying the clinical implications. Similar methodologies resulted in similar threats to validity. Spinal kinematics focussed on either shoulder-counter rotation or the cardinal planes, however studies differed in the region of the spine analysed. All kinematic studies were limited to the laboratory setting. Furthermore, reporting data relating to cardinal spinal movements is recommended to aid in comparison with other literature and enable better understanding of injurious kinematics. Studies investigating the links between kinematics and LBP/injury are limited. Future research should focus on measuring GRF and kinematics of fast bowling during live cricket, overcoming the limitations outlined in this review. Linking these findings to LBP and injury is imperative to enhance the understanding of LBP and injury in fast bowling.

2.6 Need for this study

To enhance future knowledge, the need for detailed and accurate bowling intervention guidelines to lower the risk of LBP and spinal injury are important. As stated below, only Bayne et al. (2016) has reported significant relationships between three-dimensional spinal kinematics and lower back injury. In order for this further enhancement in knowledge to occur, limitations to previous studies should be identified and overcome. The review of the literature in this chapter has attempted to provide a contemporary understanding of the problem and highlight limitations as a framework for future studies to enhance the knowledge around cricket fast bowling and LBP and/or injury.

This review has highlighted that, junior bowlers are at a significantly higher risk of lower back pain and injury when compared with senior bowlers. Despite significant focus in current literature, the relationship between fast bowling impacts and risk of lower back injury is still unclear. The relationship between fast bowling impacts and LBP has seen less attention and is consequently even more ambiguous. These findings suggest that large impacts may be a contributing factor but may not be the best indicator of injury risk in isolation from other variables such as bowling kinematics or bowling workload. Furthermore, due to limitations in the force plate technology used to measure impacts, some variables that may contribute to differences in impact characteristics (such as playing surface) have not been quantified. Novel portable technologies are now able to measure these impacts and as such, their use in the analysis of the above question may aid in the understanding of bowling impacts.

Only Bayne et al. (2016) has highlighted any significant three-dimensional spinal kinematics that may affect risk of injury, with excessive shoulder counter-rotation (a two-dimensional measure) being more commonly reported to significantly increase risk of lower back injury. The combination of lateral flexion and rotation has been hypothesised to increase risk of lower back injury and pain, which only partly agrees with the conclusions of Bayne et al. (2016) highlighting increased lumbar flexion and lateral flexion as risk factors. Due to a lack of standardised testing procedures, between-study comparisons are difficult and thus, further investigation is needed to clarify the mechanisms of lower back pain and injury. As with fast bowling impacts, bowling kinematics have predominantly been measured in a laboratory environment and as such,

'in-field' measurement of bowling kinematics may provide a more representative analysis of competitive fast bowling scenarios. Future studies should work towards analysis of bowling during 'live play' with novel minimally invasive technologies able to quantify front-foot kinetics and spinal motion in a consistent manner.

Consequently, this thesis aims to address the following questions:

1. Is accelerometry a valid and reliable method of analysing fast bowling impacts at the tibia and sacrum?

2. Are inertial sensors a valid and reliable method of analysing fast bowling spinal kinematics?

3. How does playing surface affect impacts during fast bowling?

4. What is the relationship between shoulder counter-rotation and three-dimensional spinal kinematics?

5. How do impact characteristics and three-dimensional spinal kinematics differ between senior and junior fast bowlers?

6. How do fast bowling sacral and tibial impacts and spinal kinematics affect the risk of lower back pain in junior and senior fast bowlers?

7. May recommendations synthesised from this thesis aiming to decrease lower back pain risk, also affect fast bowling performance?

Chapter 3

Development of Methods

3.1 Development of Methodology

3.1.1 Rationale for Methodology

The previous chapter has provided an in-depth analysis of current literature pertaining to fast bowling biomechanics, as well as methodologies used for their analyses. Current biomechanical analyses of fast bowling has typically been the domain of biomechanics laboratories, employing optoelectronic motion analysis systems and force plates (Bayne, Elliott, Campbell, & Alderson, 2016; Burnett, Barrett, Marshall, Elliott, & Day, 1998; Crewe, Campbell, Elliott, & Alderson, 2013; Elliott & Foster, 1984; Elliott, Hardcastle, Burnett, & Foster, 1992; Foster & Elliott, 1985; Foster, John, Elliott, Ackland, & Fitch 1989; Hurrion, Dyson, & Hale, 2000; Mason, Weissensteiner, & Spence, 1989; Portus, Mason, Elliott, Pfitzner, & Done, 2004; Ranson, Burnett, King, Patel, O'Sillivan, 2008; Ranson, King, Burnett, Worthington, & Shine, 2009; Stuelcken, Ferdninands, & Sinclair, 2010; Worthington, King, & Ranson, 2013). Whilst the laboratory can provide an ideal 'controlled' environment with optoelectronic motion analysis systems offering highly reliable and accurate motion analysis, there are a number of inherent limitations (Bayne et al., 2016; Crewe et al., 2013; Hurrion et al., 2000; King, Worthington, & Ranson, 2016; Middleton, Mills, Elliott, & Alderson, 2016; Portus et al., 2004; Spratford & Hicks, 2014; Stuelcken & Sinclair, 2009; Worthington et al., 2013). It is difficult to know if bowlers in such environmental constraints bowl with an action identical to that seen during live cricket. Furthermore, such technologies are highly costly and the demand for space, if a natural run up is to be achieved, is large. Therefore, in order for coaches to be able to monitor performance routinely during live practice and match play, alternatives measurement options are necessary.

3.1.2 Reliability and Validity

Accelerometers at the tibia and sacrum have previously been validated as a representative measure of kinetic variables (see table 3.1.1), such as peak and time-to-peak acceleration during high impact movements including running, jumping and falling (Crowell, Milner & Hamill, 2010; Sell, Atkins, Opp, & Lephart, 2014; Theobald, Whitelegg, Nokes, & Jones, 2010; Tran, Netto, Aisbett, & Gastin, 2010). Tran and colleagues (2010) reported that accelerometer data resulted in moderate correlations when compared with GRF data during jumping and landing tasks and minimal measurement error, suggesting accelerometers may be a valid method for measuring impacts in the field (Tran et al., 2010). Furthermore, running literature has highlighted a

very strong relationship between tibial accelerations recorded by tibial mounted accelerometers and GRF obtained using force plates ($r^2 = 0.95$) (Hennig, Milani, & Lafortune, 1993). Whilst accelerometers are becoming a more commonly utilised method, they have not yet been used for the analysis of impacts during fast bowling. This technology may offer a solution for real-time in-field analysis of three-dimensional fast bowling impacts that has not previously been available to coaches. The addition of this analysis may provide vital support required for injury prevention, rehabilitation and technique modification that has typically relied on subjective coach observation when in the field.

Inertial sensors (consisting of gyroscopes, accelerometers and magnetometers) have been validated for the use in clinical analysis of three-dimensional spinal kinematics (table 3.1.2) and more dynamic sporting movements (Charry, Umer, & Taylor, 2011; Hu, Charry, Umer, Ronchi, & Taylor, 2014; Swaminathan, Williams, Jones, & Theobald, 2016; van den Noort, Scholtes, & Harlaar, 2009; Williams, Haq, & Lee, 2013; Williams & Bentman 2014). Inertial sensors have been directly compared to more traditional motion analysis systems for the measurement of spinal motion. Correlation coefficients of >0.78 with RMSE <3.1° were reported using inertial sensors some ten years ago (Wong, Wong, & Lo, 2007). With further enhancement of an evolving technology, validation to optoelectronic systems have provided RMSE of <1.9° (Mjosund et al., 2017; Walgaard, Faber, van Lummel, van Dieën, & Kingma, 2016) and correlations of >0.99 (Walgaard et al., 2016). In addition to comparisons with optoelectronic systems, inertial sensors have also been compared to electromagnetic systems. Very strong correlations with values reported from electromagnetic systems in clinical settings (as high as $R^2 = .999$) have been achieved with mean differences <1° (Ha, Saber-Sheikh, Moore, & Jones, 2013; Saber-Sheikh, Bryant, Glazzard, Hamel, & Lee, 2010).

As inertial sensors are not dependant on cameras or line of sight, they offer the potential for 'in-field' data collection. Furthermore, a review analysing the validity of using inertial sensors for human movement highlighted good validity and reliability (Cuesta-Vargas, Galan-Merchant, & Williams, 2010). The reliability of inertial sensors for the analysis of three-dimensional spinal kinematics during fast bowling has not been previously investigated. Variables such as peak and time-to-peak acceleration and three-dimensional spinal kinematics could generate important information in relation to fast

bowling, providing coaches and practitioners with new insights into fast bowling. Prior to the uptake of any new technology, a task specific reliability and validity analysis is warranted. Therefore, this thesis first aims to assess the reliability of novel in-field fast bowling analysis, utilising inertial sensors to analyse three-dimensional tibial and sacral impacts and spinal kinematics during fast bowling in cricket.

Table 3.1.1. Overview of reliability and validity literature for accelerometers measuring impacts during human movement.

Authors	Anatomical Region	Sensor make/type	Sampling Frequency	Movement	Variables Analysed	Comparator	Results
Raper et al. 2018	Tibia	1 3-axis accelerometer (ViPeform)	100Hz	Running	Peak acceleration	Force plate (Kistler)	ICC = 0.974
Simons and Bradshaw, 2016	T1, S1	2 3-axis accelerometers (Catapult)	100Hz	Single and double leg landing tasks	Peak Resultant Acceleration	Force Plate (Kistler)	R= 0.73-0.83 (T1) R= 0.77-0.81 (S1)
Boutaayamou et al. 2015	Foot	2 3-axis accelerometers	200Hz	Walking	Temporal Parameters	Force Plate (Kistler), 3D MAS (Codamotion), Video Camera.	Accuracy of detection< 5ms between systems.
Meyer et al. 2015	Pelvis	2 3-axis accelerometers (Actigraph GT3X+ and GeneActive)	100Hz	Walking, Running, Jumping	Peak Acceleration	Force Plate (Kistler)	R = 0.90 and 0.89, p<0.001 (Actigraph and GeneActive)
Fortune et al. 2014	Lower limbs	7 3-axis accelerometer	100Hz	Walking	Peak acceleration and loading rate	Force Plate (AMTI)	R^2= 0.53-0.75 (vertical) R^2= 0.62-0.70 (resultant) R^2= 0.65-0.82 (loading rate)
Charry et al. 2013	Tibia	2 3-axis accelerometer (ViPerform)	100Hz	Running	Peak acceleration	Force Plate (AMTI)	R^2=0.81
Tran et al. 2010	Trunk	1 3-axis accelerometer (SPI Pro)	100Hz	Jumping and Landing	Peak Acceleration	Force Plate (ACG)	R = 0.45-0.70 Sig. relationships to comparator (p<0.05).
Wixted et al. 2010	L3-L4	1 3-axis accelerometer	500Hz	Running	Temporal Parameters	In-sole pressure sensors (paromed Vetrriebs GmbH & Co. KG)	Clear associations between methods (no quantitive statistics produced)
Elvin et al. 2007	Tibia	1 1-axis (along-tibial) accelerometer (ADXL78)	1000Hz	Jumping	Peak Acceleration	Force Plate (AMTI)	R^2= 0.812, p≤0.01

Table 3.1.2. Overview of reliability and validity literature for the measurement of three-dimensional spinal kinematics using inertial measurement units.

Authors	Anatomical Region	Sensor make/type	Movement	Sampling Frequency	Comparator	Results
Robert-Lachaine et al. 2017	Whole Body	17 Xsens IMUs	Walking and squatting	30Hz	Optotrak	ICC = 0.90-0.94 No sig. diff to comparator.
El-Gohary et al. 2017	Lower lumbar	3 Opal IMUs	Push and Release Test	128Hz	8-cam Optoelectronic (Motion Analysis 120Hz)	R = 0.592-.994
Fasel et al. 2017	Shank, Thigh, Sacrum, Sternum	6 IMUs	Downhill Giant Slalom	500Hz	6-cam (50Hz)	R=0.65-0.99
Muller et al. 2016	Elbow, Wrist	2 Xsens IMUs	Elbow flex/ext, wrist pro/supination		Vicon Optoelectronic system	RMSE = 2.7-3.8°
Cahill-Rowley and Rose, 2017	Wrist	IMUs	Reaching	128Hz	8- Cam Optoelectronic (Motion Analysis 100 Hz)	ICC = 0.72-0.98
Van den Noort et al. 2016	Hand, Finger	11 IMUs	Finger movement tasks		Optotrak	RMS diff = 2.6-7.2°
Lanovaz et al. 2017	Whole Body	6 Mobility Lab IMUs	Walking	128Hz	8-cam Optoelectronic (Vicon 100Hz)	RMS diff = 0.014-0.026s
Walgaard et al. 2016	Trunk	1 DynaPort IMU	Sit-to-walk	100 Hz	Optotrak (200Hz)	RMSE = 0.55-1.66°
Von Marcard et al. 2016	Whole body	10 Xsens IMUs	Walking, running, jumping, punching, arm rotation.	50Hz	8-cam (50Hz)	3D position error = 3.76-5.15cm
Faber et al. 2016	Whole Body	17 Xsens IMUs	Trunk bending	200Hz	Optotrak (50Hz)	R^2= 0.56-0.99
Neville et al. 2015	Postural Stability	1 Motion Intelligence IMU	Balance tests	250Hz	12-cam Optoelectronic (Vicon 120Hz)	R=0.79-0.88

Sabatini et al. 2015	Lower Trunk, Shank	2 Invensense IMUs	Walking	100Hz	5-cam optoelectronic (Vicon 100Hz)	RMS Diff = 0.5-3.2°
Najafi et al. 2015	Lower limb, Trunk	3 LEGSys IMUs	Golf swing	60Hz	5-cam optoelectronic (Vicon 60Hz)	R = 0.9-0.99
Xu et al. 2015	Cervical Spine	Oculus Rift	Neck ROM	60Hz	Optotrak (60Hz)	RMSE = 3.9-9.5°
Bugane et al. 2014	Sacrum	F4A IMU	Sit-to-Walk	100Hz	8-cam Optoelectronic (Vicon 100Hz)	R=0.88-0.95
Bergamini et al. 2014	Lumbar Spine	Opal IMU	Manual tasks and locomotion	128Hz	9-cam Optoelectronic (Vicon 100Hz)	RMSE = 3.5-7.3°
Lambrecht and Kirsch, 2014	Upper Limb	2 InvenSense IMUs	ROM	100Hz	Optotrak (50Hz)	R = 0.988 No sig. diff between IMU and comparator.
Kortier et al. 2014	Hand and Finger	15 IMUs	Finger positions	100Hz	Optoelectronic (PTI VisualEyez)	RMS diff = 2.0-12.4mm
Laudanski et al. 2013	Lower Limb	7 Xsens IMUs	Stair ascent/ descent	120Hz	Optotrak (50Hz)	ROM diff = 3.3±8.1°
Bourgeois et al. 2014	Lower Limb	2 Physilog IMUs	Walking		7-cam optoelectronic (Vicon)	Mean diff = 0.5 ±2.9°
Faber et al. 2013	Upper limb and trunk	3 Xsens IMUs	Reaching task	100Hz	Optotrak (100Hz)	RMSE = 0.5-3.5°
Zhang et al. 2013	Lower Limb	Xsens IMUs	Walking and stair ascent/ descent	100Hz	Optotrak (100Hz)	CMC = 0.5-0.96

3.1.3 Validity and reliability of sensors used in this study

Whilst literature has highlighted the validity and reliability of inertial sensors as a method, previous studies have also successfully used the same 3AMG inertial sensors in sporting and clinical environments as is utilised in this thesis. Identical sensors have been used to measure spinal (Lumbar, Thoracic and Cervical) range of motion during live rugby scrummaging (Swaminathan et al. 2016), and to quantify shoulder joint position sense (Bewes et al., 2015). However, prior to employing them to measure spinal motion in cricket, validity and reliability was established.

3.1.3.1 Company calibration

Prior to shipping the 3AMG sensors were calibrated by THETAmetrix. This process is done to minimise any sensor related error and takes into consideration offsets and gains from mounting of individual sensing elements resulting in a minimisation of on-board sensor errors. In addition a temperature calibration specific to the sensors used in this thesis was applied by the company prior to shipping. The following data pertaining to these specific sensors (serial number 134301) was provided.

Table 3.1.3.Accuracy data achieved following factory calibration of inertial sensors used in this thesis.

Axis	Average Error (^{o})	Standard Deviation of Error (^{o})
Roll	0.20	0.58
Pitch	0.13	0.61
Heading	0.02	1.97

3.1.3.2 Gimbal Rotatory Jig validation

General measurement of sensor accuracy is reported by the company data sheet as 0.5^o for pitch and roll and 1° for heading. However, a bench top procedure was also used to determine sensor accuracy. This has been reported in detail elsewhere (Swaminathan, 2016b). Briefly a bespoke rotatory jig was used to determine accuracy of the sensor outputs. The jig determined orientation through digital encoders (ERN-420, Heidenhain, Sweden) each of which had 3600 lines per revolution with 4 steps per line giving steps of 1/40th of a degree.

Sensors were fixed to the rotatory table using double sided tape and two of the three gimbles were fixed allowing free motion in only one plane. The jig and sensors were rotated about +/- 180 degrees for roll and heading axes and =/- 90° for the pitch axis at a metronome controlled speed of around 30^os^{-1}. Absolute error was calculated for each data point, to determine the mean absolute error for each plane. Lin's concordance correlation coefficient was used to determine data concordance and chosen as it provides a coefficient of how closely two measurement techniques measure the same variable (i.e. a 1:1 ratio in this case). All coefficients of concordance were >0.99 for all axes. Mean absolute percentage errors were 1.4% for pitch, 2.2% for roll and 4% for heading and it was noted that accuracy was much better +/- 60 degrees. The larger heading error is common in inertial sensors due to the relatively greater reliance of the magnetometer for this axis. The magnetometer is known to suffer from hard iron interference, however this is unlikely to have the same effect on the cricket field.

3.1.4 Instrumentation and sensor attachment

Following pilot testing of sensor attachment methods (as seen in appendix 4), three inertial sensors (THETAmetrix, Waterlooville, UK) were attached to the skin over the T1, L1 and S1 spinous processes with double-sided tape and re-enforced with elastic adhesive bandage (figure 3.1.1). Bony landmarks were identified according to the directions outlined in Field and Hutchinson (2013). Sensors contained accelerometers, gyroscopes and magnetometers sampling at 100 Hz. Sensors were wired to a small processor board which communicated wirelessly to a laptop computer. An additional accelerometer ($\pm$200 g) sampling at 750 Hz was also attached to the medial aspect of the mid-tibia on the bowlers front and back leg (right leg for a right handed-bowler) with double-sided tape, vertically aligned to the tibia and secured further with a compressive bandage, as seen in figure 3.1.2 (with bandage removed).
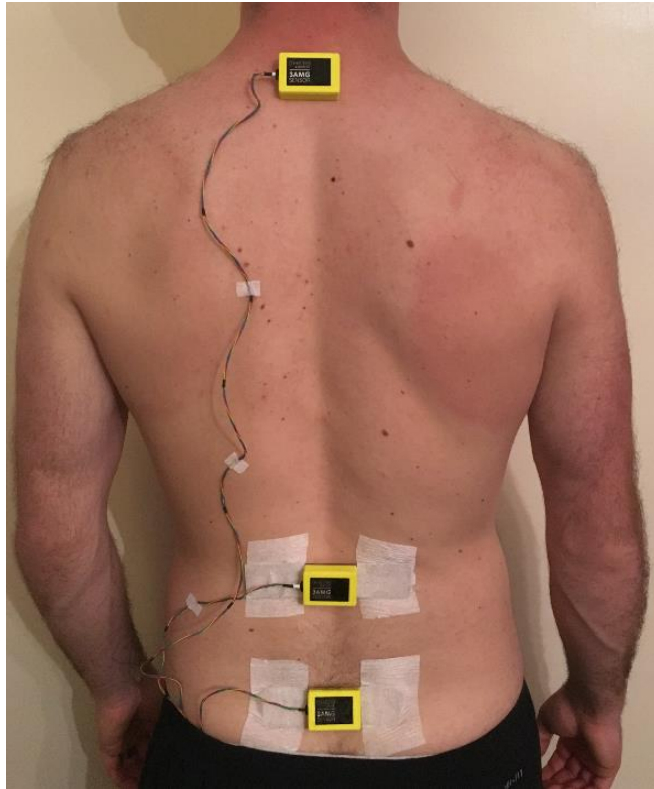
Figure 3.1.1. Placement of spinal inertial sensors on S1, L1 and T1 vertebrae
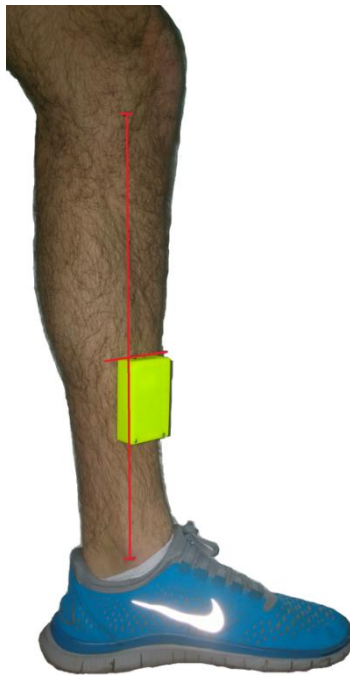
Figure 3.1.2. Placement of tibial accelerometer on the mid-tibia.

Each bowler completed a 'self-prescribed' warm up until they felt ready to bowl. Bowlers were then instrumented with sensors as previously described. Instructions to bowl six balls (one over) with maximal effort were given to enable the participants to familiarise themselves with bowling whilst instrumented. Following this, participants bowled with maximal effort for one over whilst data were recorded. One over was deemed sufficient due to previous literature and this thesis' reliability study highlighting little variability between overs and deliveries (Burnett et al. 1995; Portus et al. 2000; Schaefer, O'dwyer, Ferdinands & Edwards, 2018). All bowlers bowled at a right-handed batsman in a standard 'nets' setup as part of a typical training session on grass wickets.

3.1.6 Data Processing

3.1.6.1 Tibial acceleration

All data were collected in Sensor Suite (Version 504) and transferred to Matlab (Ed. R2012a). Acceleration data were filtered using a bidirectional second-order, zero lag low-pass Butterworth filter with a cut-off frequency of 50 Hz to remove high frequency noise. Residual analysis of tibial acceleration can be seen below, highlighting 50Hz as an appropriate cut-off using the method outlined by Winter (1990).
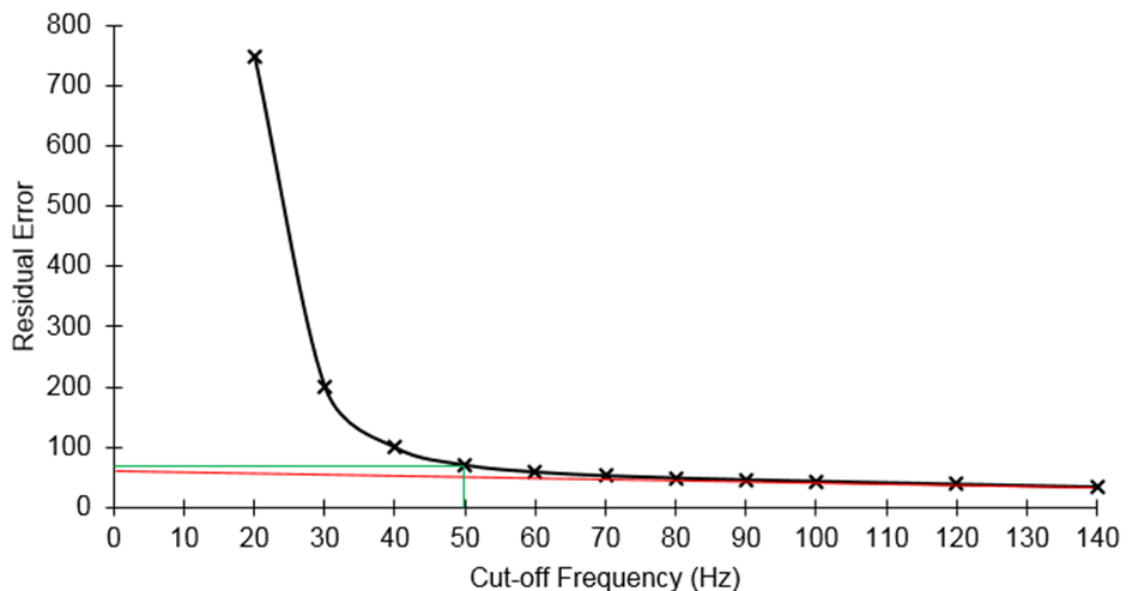
Figure 3.1.3. Residual analysis of tibial acceleration at front-foot impact during fast bowling using the method outlined in Winter (1990).

Peak tibial acceleration (*x, y* and *z*) were calculated relative to the orientation of the tibia, as the absence of an integrated gyroscope did not allow for correction of sensor tilt. Peak acceleration, peak resultant acceleration, time–to-peak acceleration and time-to-peak resultant acceleration data were identified during the back and front-foot impact phases of fast bowling (BFI and FFI). BFI and FFI were defined as the instance of peak acceleration on the corresponding tibia (see figure 3.1.4).
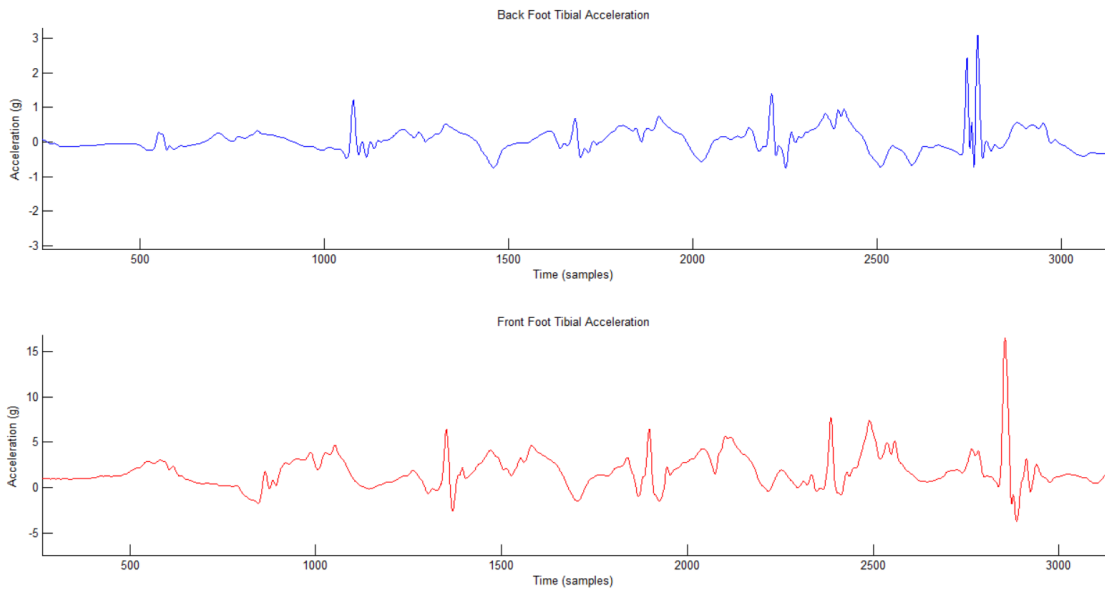
Figure 3.1.4. Synchronised back and front-foot tibial acceleration during fast bowling.

Time-to-peak acceleration was defined as the time taken for acceleration to reach its peak from the point of initial increase on the impact peak (figure 3.1.5).
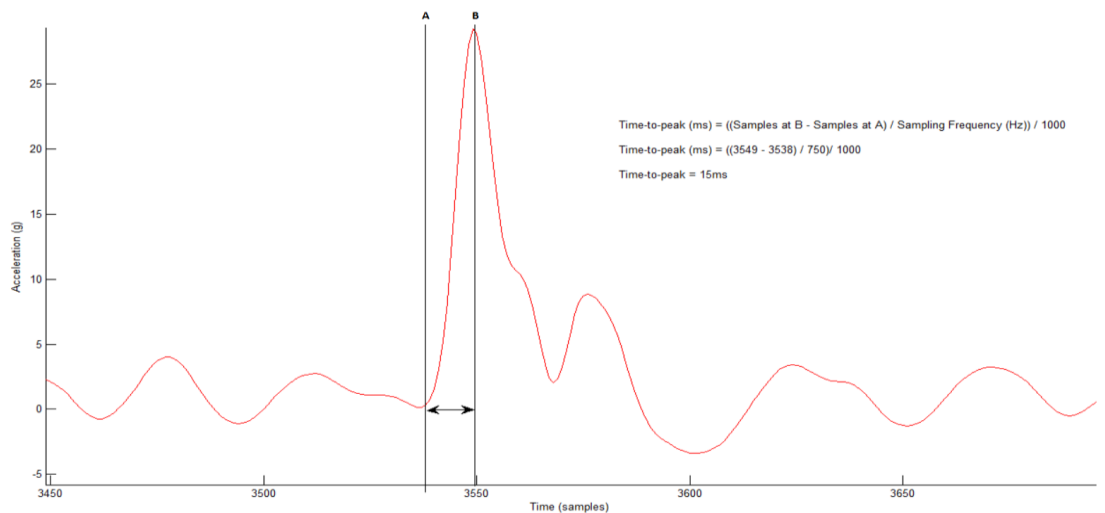
Figure 3.1.5. Illustration of the calculation of time-to-peak acceleration

Peak resultant acceleration was defined as the square root of the sum of squared accelerations along all three axes. Peak acceleration along the x axis and peak resultant acceleration were also normalised to gravity per kilogram of body mass ($g.kg^{-1}$) to allow a standardised measure of impact. Average loading rate was calculated by dividing peak tibial acceleration by time-to-peak acceleration (Stiles and Dixon, 2007).
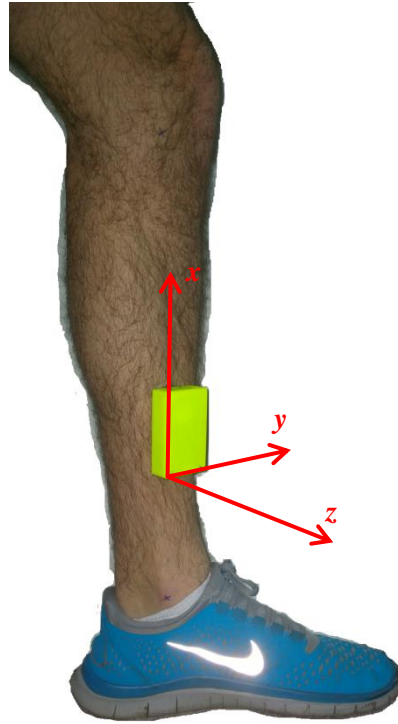
Figure 3.1.6. Tibial accelerometer axes. x = along tibial axis, y = perpendicular to x along to second edge of the sensor casing and, z = perpendicular to x along the short edge of the sensor casing.

3.1.6.2 Sacral Accelerations

All data were collected in Sensor Suite (Version 504) and transferred to Matlab (Ed. R2012a). The fused sensor elements (accelerometer, gyroscope and magnetometer) provide drift free orientation calculation allowing accelerations to be corrected for sensor tilt, thus unlike tibial acceleration, sacral accelerations were able to be expressed in terms of vertical, mediolateral and anterior-posterior acceleration. Correction for tilt was calculated as in equation 3.1.1. The same axes were assigned as the tibial accelerometer, thus with no sensor tilt: Vertical = x, mediolateral = z, anterior-posterior = y. In the below equations α = sensor pitch, β = sensor roll.

Correction for sensor tilt to produce vertical sacral acceleration

$$Pitch\ Corrected\ x = \frac{Raw\ x\ acceleration}{cos\alpha}$$

Once *x* acceleration has been corrected for sensor pitch, it was then corrected for sensor roll to produce true vertical sacral acceleration.

$$Vertical\ Sacral\ Acceleration = \frac{Pitch\ Corrected\ x}{cos\beta}$$

1 gravity ($9.81ms^{-2}$) was then subtracted from the signal to remove acceleration due to gravity.

Correction for sensor tilt to produce anterior-posterior sacral acceleration

$$Anterior - posterior\ Sacral\ Acceleration = Raw\ y\ acceleration \cdot cos\alpha$$

Correction for sensor tilt to produce mediolateral sacral acceleration

$$Mediolateral\ Sacral\ Acceleration = Raw\ z\ acceleration \cdot cos\beta$$

Equation 3.1.1. Calculations for the correction of sensor tilt for vertical, anterior-posterior and mediolateral sacral acceleration.

Peak resultant acceleration and time-to-peak vertical and resultant acceleration were also identified at the sacrum at BFI and FFI. Time-to-peak and resultant acceleration were calculated using the same method as described for tibial accelerations. Peak vertical acceleration and resultant acceleration were normalised to gravity per kilogram in the same way as tibial acceleration. In addition, attenuation of peak resultant acceleration between the tibia and sacrum was calculated using the equation 3.1.2. Thus, this metric is able to express how much peak resultant acceleration has decreased between the tibia and sacrum.

$$Acceleration\ Attenuation\ = 100 - \left(\left(\frac{Peak\ Resultant\ Sacral\ Acceleration}{Peak\ Resultant\ Tibial\ Acceleration} \right) * 100 \right)$$

Equation 3.1.2. Calculation for the expression of attenuation of resultant acceleration from the tibia to the sacrum.

3.1.6.3 Spinal Kinematics

All raw data collected in Sensor Suite (Version 504) and transferred to Matlab (Ed. R2012a) for processing. Absolute orientations provided by each inertial sensor were used to derive relative angles between two sensors from their direction cosine matrices (Burnett et al. 1998). This enabled the spine to be divided into lumbar, thoracic and thoracolumbar regions. The natural standing posture at the back of the bowler's run-up facing the direction of delivery (towards the wickets) was taken as the initial frame of reference from which all movements were determined. Resultant movement-time graphs were filtered using a bidirectional second-order, zero-lag low-pass Butterworth filter with a cut off frequency of 5Hz (Burnett et al. 1998). Residual analysis of lumbar rotation can be seen below, highlighting 5Hz as an appropriate cut-off using the method outlined by Winter (1990).
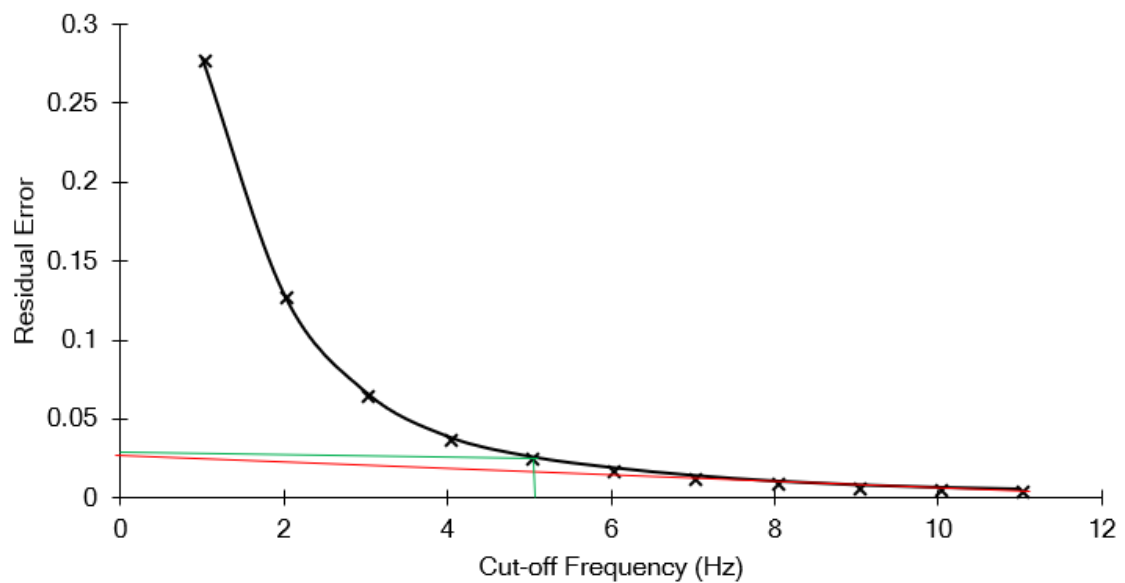
Figure 3.1.7. Residual analysis of lumbar rotation during the fast bowling delivery stride using the method outlined in Winter (1990).

Back-foot impact (BFI) was determined from the synchronised tibial mounted accelerometer (peak along-tibial acceleration). Spinal inertial sensors and the tibial accelerometer were synchronised using linear interpolation and trimmed to the point of neutral posture before the beginning of the run-up using a manual data mark in the software (as seen in figure 3.1.8).
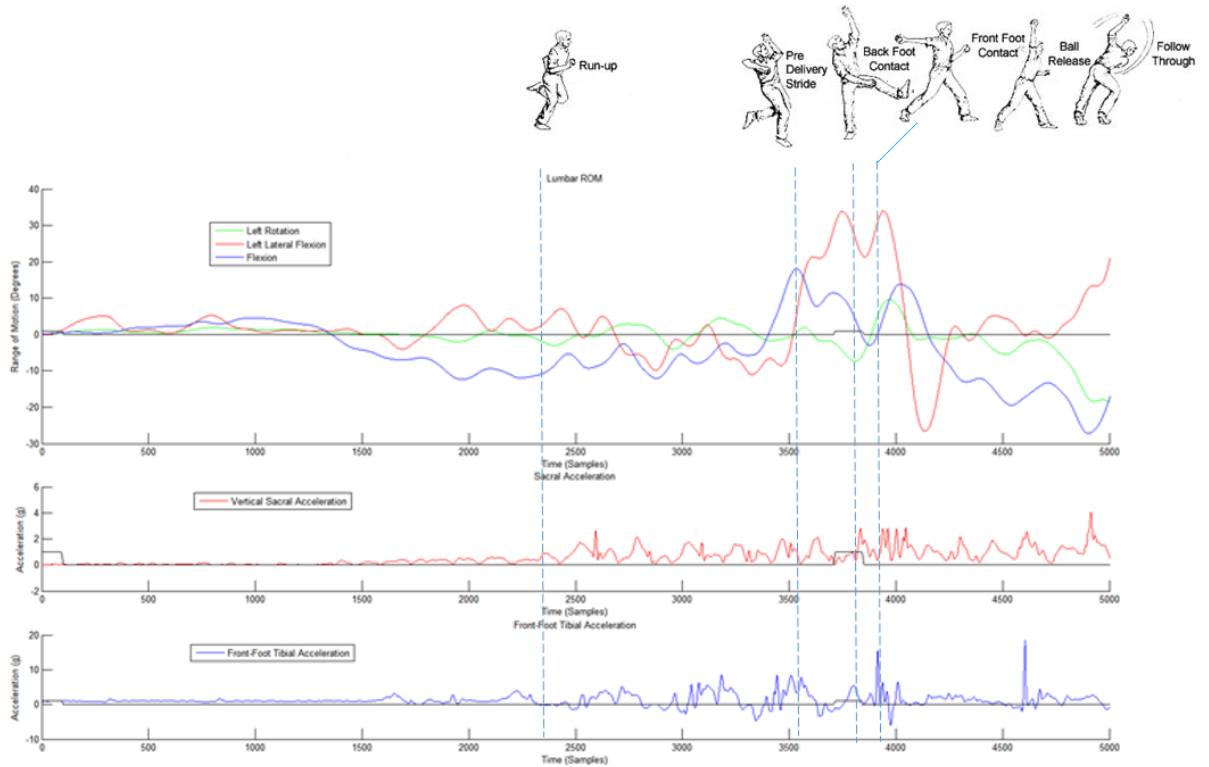
Figure 3.1.8. Lumbar spinal kinematics synchronised with vertical sacral and along-tibial acceleration during fast bowling. Indents in black line along the 0g mark represent manual data marking at neutral posture (0 time) and visual identification of back foot impact (approx. 3800 samples). Pictures correspond to typical phases within a delivery (ball release and follow through not aligned as this cannot be identified from the data presented).

Front-foot impact (FFI) was determined from the acceleration signal of the sacral mounted accelerometer. Therefore, kinematics between BFI and FFI were determined to enable comparison of the delivery stride as is common in fast bowling literature. Analysis of a random sample of deliveries (n=60) highlighted mean delay from peak tibial acceleration to peak sacral acceleration at FFI was 82±57ms (figure 3.1.9).
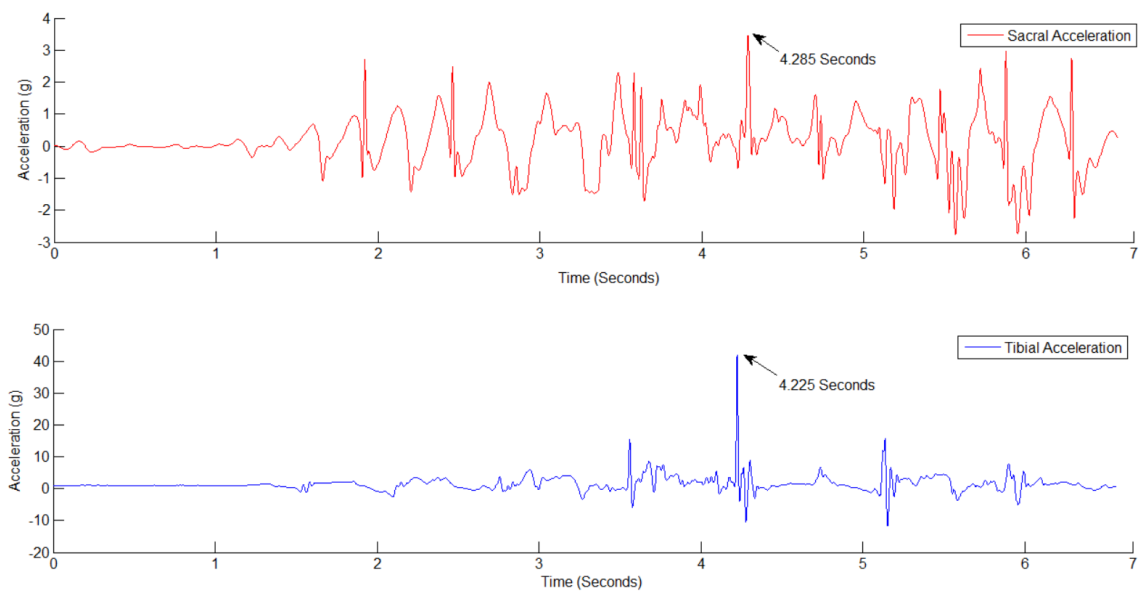
Figure 3.1.9. Delay between peak tibial and sacral acceleration at front-foot impact (60ms).

In addition to spinal range of motion, shoulder counter-rotation (SCR) and hip-shoulder separation angle (HSS) were calculated by subtracting T1 orientation at BFI from T1 maximum right rotation (SCR) and taking the maximum difference in hip and shoulder orientation about the longitudinal axis following BFI (HSS)(Portus et al. 2004).
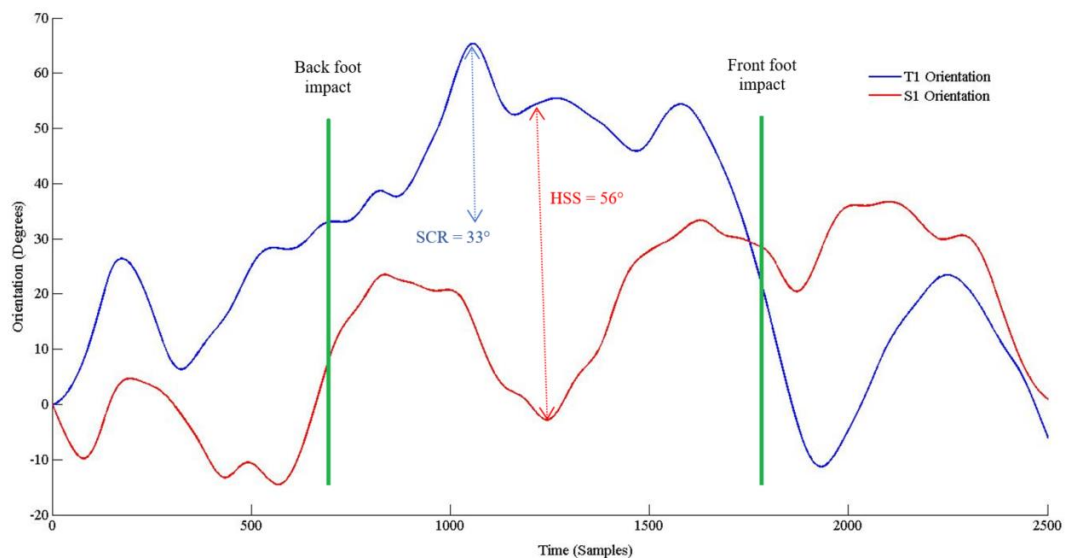
Figure 3.1.10. Shoulder and hip orientation relative to the wickets, used to calculate shoulder counter-rotation (SCR) and hip-shoulder separation (HSS).

Spinal kinematics were described as the resultant spinal orientations at BFI and FFI with ROM during the delivery stride representing the difference between FFI and BFI and reported as flexion, extension, lateral flexion and rotation for each spinal region. In addition to this, SCR and HSS were also reported. All data for left-handed bowlers were converted to read as data for right-handed bowlers. Therefore, flexion, left lateral flexion and left rotation were defined as positive.

3.2 Additional Methodologies

3.2.1 Study 4.1 – Validity

Validity Instrumentation

Bowling kinematics were recorded using a 14 camera Vicon Motion Analysis System (Oxford, UK) operating at 200Hz. Ground reaction forces at BFI and FFI were also recorded using two Kistler force plates (900x600mm) sampling at 1000Hz.

39 14mm retroflective markers were attached to each participant, positioned on landmarks dictated by the full body plug-in-gait model (Figure 3.2.1)(Vicon Nexus 2.7).

Inertial sensors were also attached as in the main methods section of this thesis.

Validity Procedure

Each participant completed a self-selected warm up and was then allowed as many bowls as was needed for them to familiarise themselves with the experimental set-up. The lab allowed for a full-length run up. The ball was then bowled into a net 5m away from the point of ball release. Following familiarisation one over (6 balls) were bowled maximally and recorded for analysis. If clean contact with force plates at BFI and FFI were not achieved the trial was repeated.
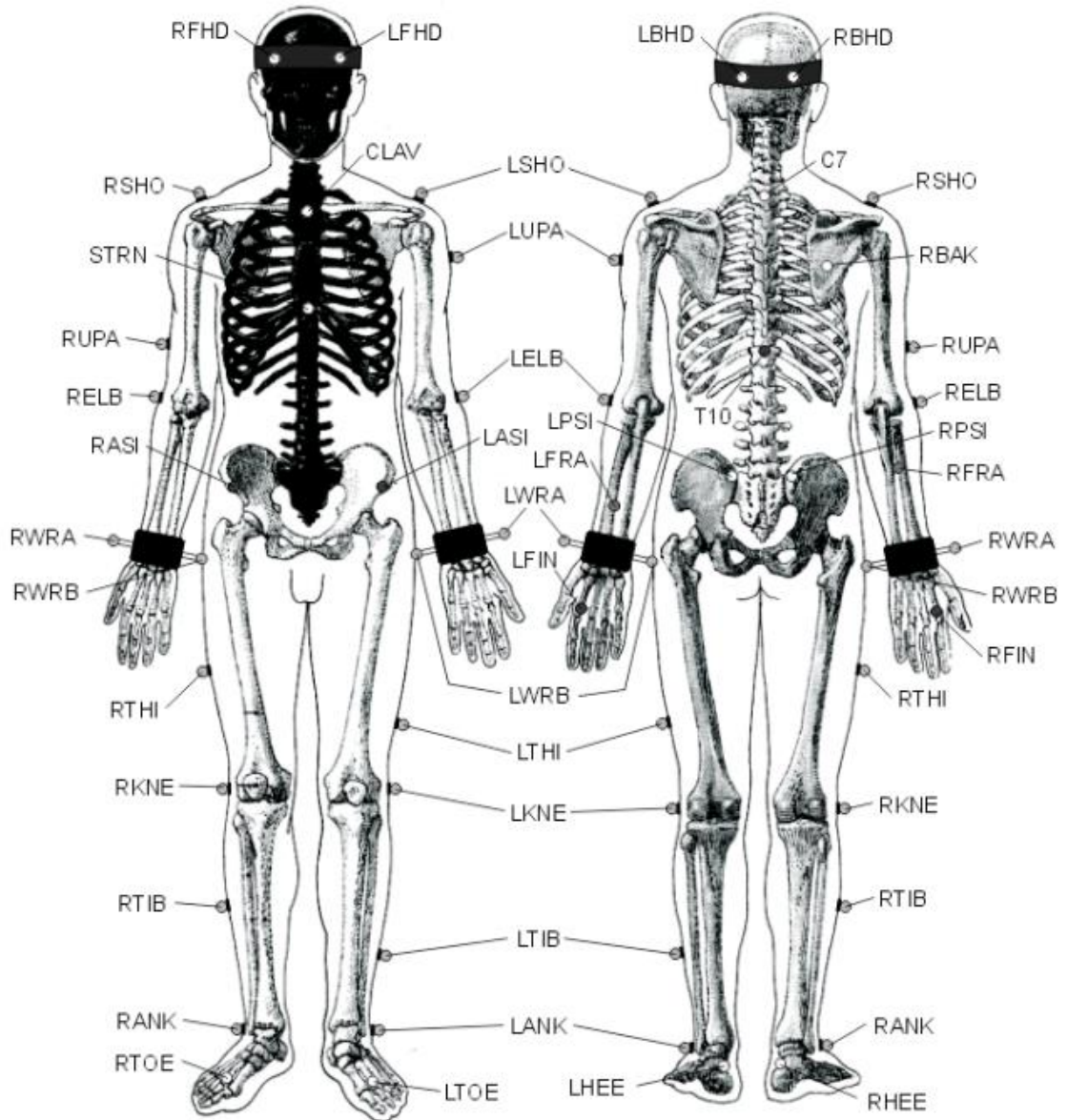
Figure 3.2.1. Marker placement for full body plug-in-gait model (Vicon Nexus 2.7).

Validity Data Processing

For the purpose of this study only orientation of the thorax and pelvis segments were needed from the plug-in gait kinematic model. The thorax segment was constructed using clavicle (CLAV), sternum (STRN), C7 and T10 markers (as labelled in figure 3.2.1). The pelvis segment was constructed using left and right anterior and posterior iliac spines (LASI, RASI, LPSI, RPSI) (figure 3.2.1).

Back and front foot impact were defined as the point at which a force greater than a 5 Newton threshold was observed on the corresponding force plate. Raw force data were

exported and processed in Matlab (R2012a). All force data were filtered using a bidirectional second-order, low-pass Butterworth filter with a cut-off frequency of 50Hz. Peak vertical, anterior-posterior, mediolateral and resultant GRF at BFI and FFI were recorded. Resultant GRF was defined as the square root of the sum of squares of vertical GRF, anterior-posterior GRF and mediolateral GRF (same method as resultant accelerations in the main methodology). Time to peak vertical and resultant GRF were defined as the time between initial contact with the force plate and peak vertical and resultant GRF. These values were also recorded at both BFI and FFI.

Shoulder counter-rotation was defined as the orientation of the thorax at BFI subtracted from max rotation away from the direction of delivery (same as with T1 inertial sensor in main method). Three-dimensional lumbar kinematics were obtained via the relative orientations of the thorax segment relative to the pelvis. Lumbar flexion, lateral flexion and rotation were recorded at BFI and FFI to enable comparisons to inertial sensor data.

Processing of inertial sensor data was identical to that described in the main method section in this thesis.

3.2.2 Study 4.2 - Playing surfaces and lower limb impacts

Impact Testing Procedure

These methods of data collection were adapted in order to quantify surface firmness as was the aim of study 4.2. In order to quantify the surface properties of different cricket playing surfaces, a custom-built impactor was developed. The same ±200 g tri-axial accelerometer (THETAmetrix, Waterlooville, UK, ADXL377), sampling at 750 Hz was utilised, aligned vertically with the centre of mass of an impact weight 63 mm in diameter and 2.5 kg in weight. The impact weight was suspended in a guidance tube to standardise drop height to 200mm (Figure 3.2.2). This testing rig system was based on similar impactor devices (Baker et al. 2001). An additional 20 mm of Adiprene polyurethane foam and 3 mm of rubber (taken from the heel of a typical sports training shoe) was attached to the bottom of the impact weight to more accurately simulate impact conditions during cricket bowling.

Acceleration data were collected across four different cricket playing surfaces: Grass wicket, artificial outdoor wicket, indoor wood and indoor rubber composite (Uniturf) (as seen in figure 3.2.4a, b, c, d,). Impact data were sampled at 12 locations of the popping crease, as this is where front-foot impact occurs during bowling. A customised

planning frame, consisting of twelve 400x440mm squares was used to identify sampling locations (seen in figure 3.2.3). Each square was tested 6 times in random locations within the square.
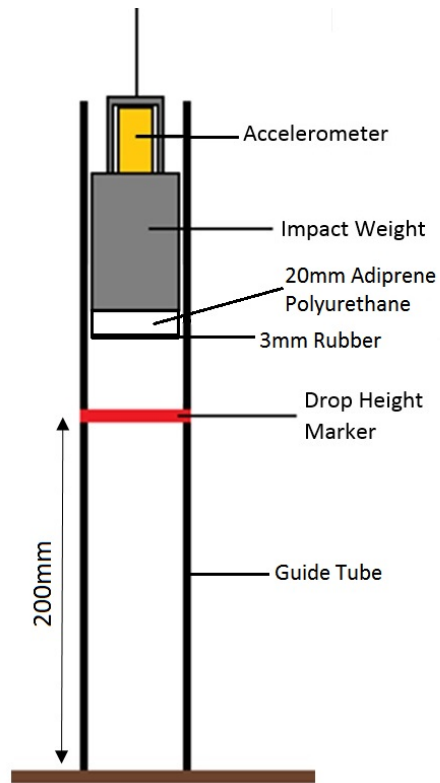
Figure 3.2.2. Set-up of a custom built impactor
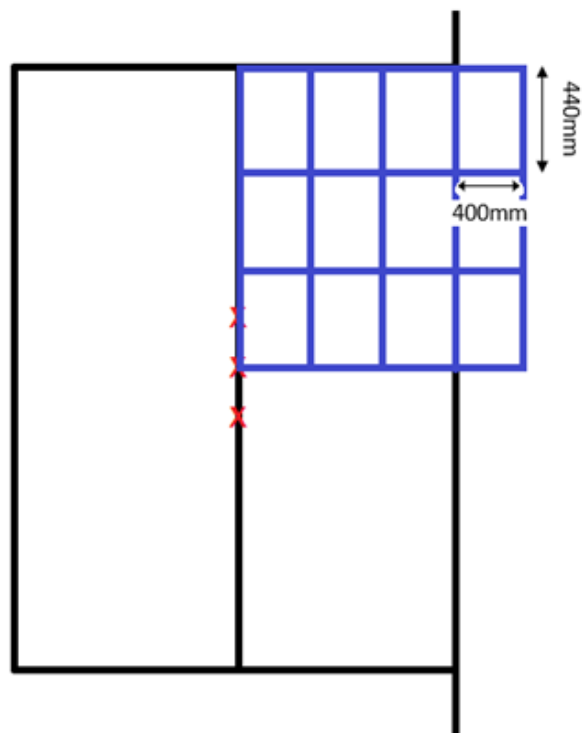
Figure 3.2.3. Segmentation of popping crease for impactor testing locations

Figure 3.2.4. a) grass wicket (top left), b) artificial outdoor wicket (top right), c) indoor rubber composite wicket (bottom left), d) indoor wooden wicket (bottom right).

Data Processing

All data were collected in Sensor Suite (v504) and transferred to Matlab (Ed. R2012a) where peak and time-to-peak acceleration data were identified for the initial impact of the weight with the playing surface. Acceleration data were filtered using a bidirectional second-order, low-pass Butterworth filter with a cut-off frequency of 50Hz. Peak acceleration was identified manually, and time-to-peak acceleration defined as the time taken for acceleration to reach its peak from the point of initial increase on the impact peak.

For surface impactor values a mean of the 6 tests at each square was taken for all variables, thus giving 12 values used to describe the characteristics of each surface.

Average loading rate was calculated by dividing peak tibial acceleration by time-to-peak acceleration (Stiles and Dixon, 2007).

### 3.2.3 Study 5.2 – Fast bowling biomechanics and lower back pain risk

Injury Surveillance

Before the start of the 2015 season, history of low back pain or injury was explored using a specifically created questionnaire, which included playing history (See Appendix 3). The questionnaire was administered with the guidance of the researcher and verified by the club physiotherapist where possible. The questionnaire sought to determine if a previous history of low back pain or injury was present, enabling a sub-grouping of bowlers based on LBP history. In addition to previous history of back pain, pain experienced in the following season was also explored. Bowlers were instructed to keep a record of any LBP or injury during the 2015 season if a physiotherapist was not able to do this for them. This study defined LBP as any pain affecting the area of the back inferior to the lower ribs, superior to the inferior gluteal folds and medial to the mid-axillary line that impacted on their ability to bowl for a minimum of 3 days. Junior and senior fast bowlers were grouped separately in order to avoid age becoming a confounding variable. Therefore, bowlers were able to be sub-classified based on whether they had a history of LBP as well as whether they went on to develop LBP in the following season.

Ball Release Speed Analysis

In addition to the fast bowling technique kinematic and kinetic variables highlighted above, study 6.1 aimed to provide a comparison with ball release speed.

Instrumentation

One high-speed video camera (Sony FX1000) sampling at 200Hz was used to record ball release speed. This camera was positioned as shown in figure 3.2.5.
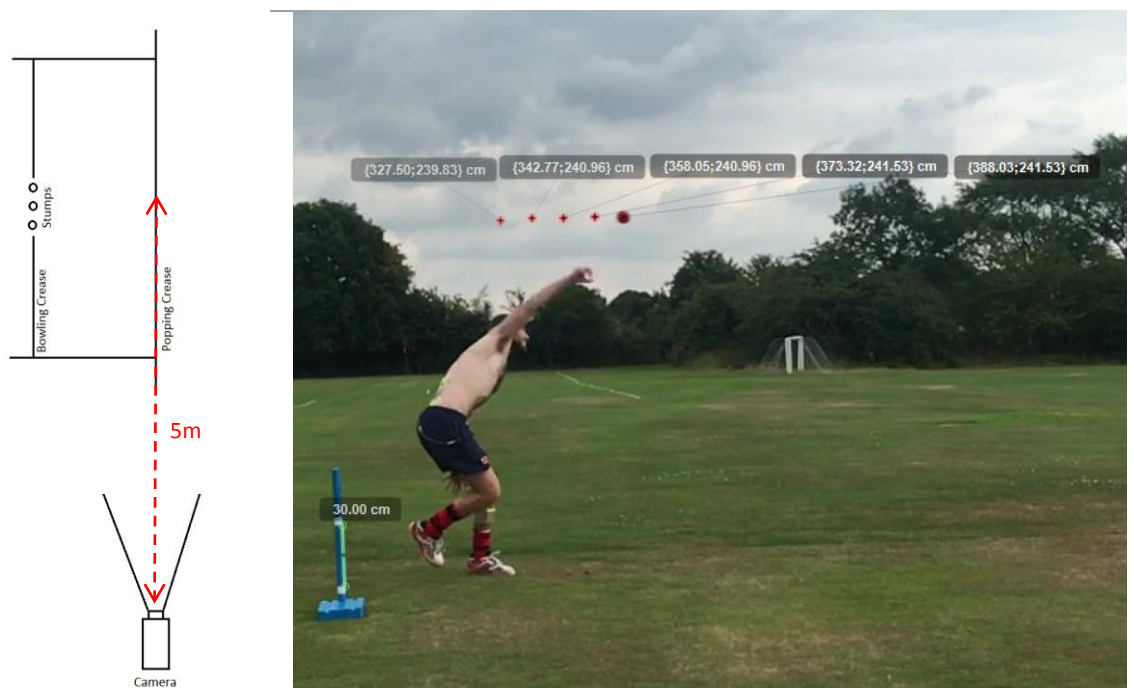
Figure 3.2.5. Camera position and example of the digitising process used to calculate ball release speed.
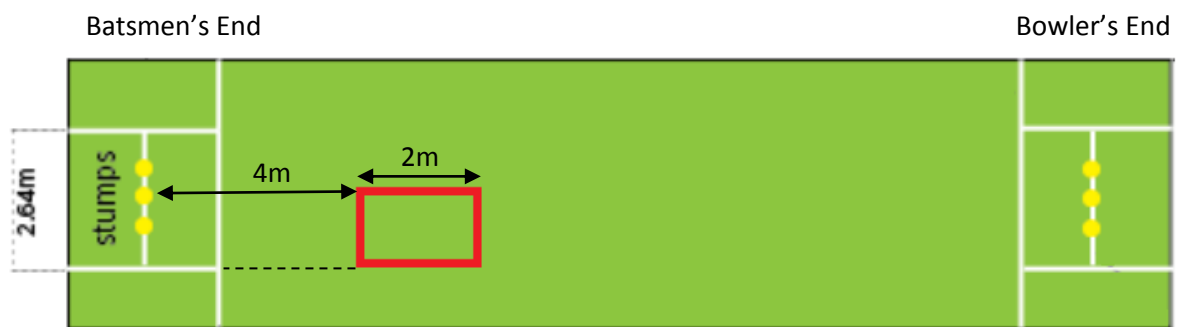
Figure 3.2.6. Target area (red box) used to define a 'successful' bowl.

The camera was positioned to record in the sagittal plane 5m from the middle stump and aligned with the popping crease. Markers were placed on the camera-facing stump 30cm apart to allow distance calibration. A target area was also placed on the pitch using cones. This area denotes a 'good' line and length and as such only trial landing inside this area were analysed. Figure 3.2.5 provides this study's definition of a 'good' line and length.

Data Processing

All video data were processed in Kinovea (v0.8.15). As shown in figure 3.2.5, the ball was tracked manually for the first 5 frames following ball release and *x*(horizontal) and *y* (vertical) coordinates were recorded. Ball speed was calculated between each of these points (as detailed in the Equation 3.2.1) with the average value recorded as the participant's ball release speed.

$$Ball\ Release\ Speed_t = \frac{\sqrt{\left(x_t - x_{(t-\Delta t)}\right)^2 + \left(y_t - y_{(t-\Delta t)}\right)^2}}{0.005}$$

X = x position, Y= y position, t = time

Equation 3.2.1. Calculation of ball release speed from digitised two-dimensional co-ordinates of ball position of two consecutive frames at 200Hz sampling frequency following ball release.

3.3 Study Methodologies

The above method for measurement of tibial and sacral impacts and three-dimensional spinal kinematics allows this thesis to address some novel questions pertaining to fast bowling biomechanics and lower back pain risk as well as strengthen the current body of literature. As such this method was adapted to address the below questions:

Study 4.1- The reliability and validity of accelerometry and inertial sensors for the measurement impacts and three-dimensional spinal kinematics during fast bowling.

Study 4.2 – How does playing surface affect front-foot tibial impact force during fast bowling?

Study 4.3 – The relationship between shoulder counter-rotation, hip-shoulder separation and three-dimensional spinal kinematics during fast bowling.

Study 5.1 – The comparison of fast bowling impacts and three-dimensional spinal kinematics in elite junior and senior fast bowlers.

Study 5.2 – The effect of fast bowling impacts and three-dimensional spinal kinematics on risk of lower back pain in elite junior and senior fast bowlers.

Study 6.1 – The relationship between ball release speed and spinal kinematics and tibial and sacral accelerations during fast bowling.

<u>3.3.1 Participants</u>

Four different cohorts of fast bowlers were used to address the above questions. Figure 3.3.1 provides an overview of how these bowlers were distributed between studies. Table 3.3.1 details a priori and post hoc sample size calculations for the above studies.
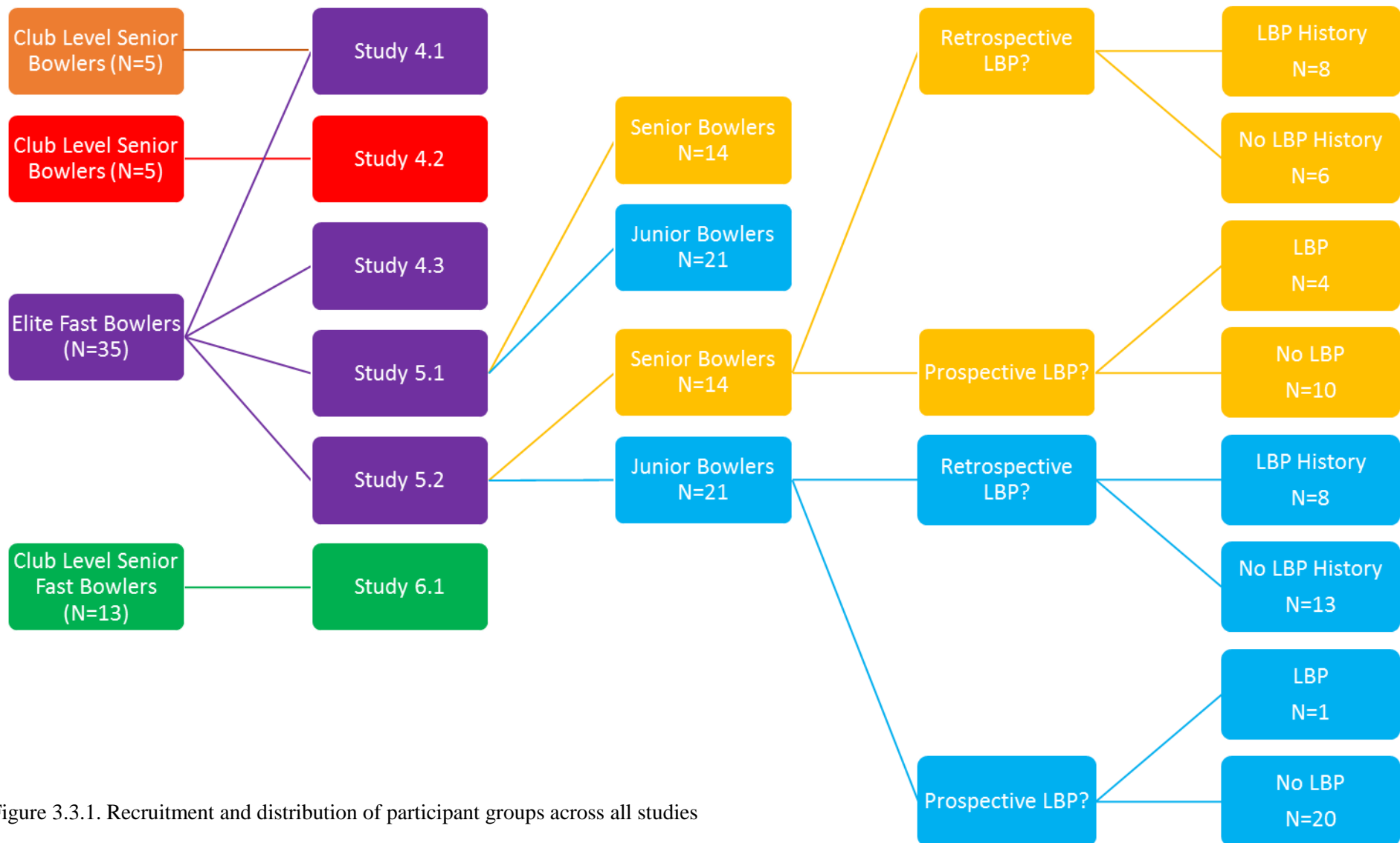
Figure 3.3.1. Recruitment and distribution of participant groups across all studies

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Table3.3.1. A priori sample size calculations and post hoc calculations of achieved power for all studies within this thesis.

Studies (Chronologically)	A Priori Sample Size	Actual Sample Size	Achieved Power	Study/Variable Used for A Priori	Primary Outcome Variable Used for Achieved Power
Study 4.1	4	5	0.68	Hurrion et al. 2000 Vertical GRF at BFI and FFI	Lumbar rotation at FFI
Study 4.2	23	5	0.637	Portus et al. 2004 Vertical GRF at FFI	Peak result acceleration at FFI
Study 4.3	16	35	0.815	Crewe et al. 2011 SCR and lumbar rotation	SCR and thoracolumbar lateral flexion
Study 5.1	44	35	0.771	Bayne et al. 2016 Thorax lateral flexion at FFI	Thoracolumbar rotation at BFI
Study 5.2 Juniors Retrospective	44	21	0.748	Bayne et al. 2016 Thorax lateral flexion at FFI	Thoracolumbar rotation at BFI
Study 5.2 Seniors Retrospective	44	14	0.412	Bayne et al. 2016 Thorax lateral flexion at FFI	Thoracolumbar flexion at BFI
Study 5.2 Seniors Prospective LBP	44	14	0.814	Bayne et al. 2016 Thorax lateral flexion at FFI	Lumbar Flexion at BFI
Study 6.1	12	13	0.794	Lumbar flexion at BFI from senior prospective LBP study	Thoracic lateral flexion at FFI

A priori sample size calculations were for an expected power of 0.8.

Study 4.1 – Reliability and validity

35 elite male fast bowlers were used for the reliability section of this study. Mean (± SD) age was 20.13 (4.62) years, height 1.84 (0.07) m and mass 80.32 (11.02) kg. Participants were recruited through coaches and physiotherapists at professional county cricket clubs. As testing was conducted in the field and a laboratory environment was needed for the validity analysis four club level senior bowlers were also recruited. Mean (± SD) age was 19.33 (1.15) years, height 1.80 (0.12) m and mass 78.67 (22.30) kg.

These participants consisted of 14 adult bowlers (mean (±SD) age 24.12(4.31) years, height 1.89(0.05) m and weight 89.20(4.63) kg) and 21 junior bowlers (mean (±SD) age 16.94(0.70) years, height 1.81(0.05) m and weight 73.00(9.21) kg). Participants were grouped differently depending on the specific research question (as seen in figure 3.3.1), however the same inclusion criteria were applied for inclusion in these studies throughout.

As reliability testing was conducted in the field and a laboratory environment was needed for the validity analysis, five club level senior bowlers were also recruited for the validity analysis. Mean (± SD) age was 19.33 (1.15) years, height 1.80 (0.12) m and mass 78.67 (22.30) kg.

Study 4.2 – Playing surfaces and lower limb impacts

Data for this study were collected from five trained male fast bowlers, mean (±SD) age was 23.70 (0.60) years, height 1.81 (0.02) m and weight 81.90 (10.10) kg. All participants were right handed and classified as fast bowlers by their club coach. Participants were recruited through amateur cricket clubs in the county league structure and were regularly training and playing at the time of testing. Participants were instructed to wear normal training shoes and to wear the same footwear on all surfaces.

Study 4.3 – Shoulder counter-rotation and three-dimensional spinal kinematics comparison 35 elite male fast bowlers were used for the reliability section of this study. Mean (± SD) age was 20.13 (4.62) years, height 1.84 (0.07) m and mass 80.32 (11.02) kg. Participants were recruited through coaches and physiotherapists at professional county cricket clubs.

Study 5.1- Junior and senior fast bowlers comparison

35 elite male fast bowlers were used for the reliability section of this study. Mean (± SD) age was 20.13 (4.62) years, height 1.84 (0.07) m and mass 80.32 (11.02) kg. Participants were recruited through coaches and physiotherapists at professional county cricket clubs. Participants were grouped into junior and senior bowlers. Therefore, groups consisted of 14 adult bowlers (mean (±SD) age 24.12(4.31) years, height 1.89(0.05) m and weight 89.20(4.63) kg) and 21 junior bowlers (mean (±SD) age 16.94(0.70) years, height 1.81(0.05) m and weight 73.00(9.21) kg).

Study 5.2 – Fast bowling biomechanics and lower back pain risk

35 elite male fast bowlers were used for the reliability section of this study. Mean (± SD) age was 20.13 (4.62) years, height 1.84 (0.07) m and mass 80.32 (11.02) kg. Participants were recruited through coaches and physiotherapists at professional county cricket clubs. Junior and senior bowlers were analysed separately in this study. Thus, participants consisted of 14 adult bowlers (mean (±SD) age 24.12(4.31) years, height 1.89(0.05) m and weight 89.20(4.63) kg) and 21 junior bowlers (mean (±SD) age 16.94(0.70) years, height 1.81(0.05) m and weight 73.00(9.21) kg). Junior and senior bowlers were grouped based on both retrospective and prospective LBP as seen in figure 3.3.1. Details on how this study defines LBP is detailed later in this methods section.

Study 6.1 – Fast bowling biomechanics and ball release speed

Data for this study were collected from 13 trained male fast bowlers, mean (±SD) age was 23.00 (5.00) years, height 1.81 (0.06) m and mass 79.0 (10.79) kg. Following correlation analysis, participants were grouped based on lower limb orientation at back-foot impact (detailed later in this section). 7 bowlers had a side-on lower limb orientation, mean (±SD) age was 22.80 (6.53) years, height 1.82 (0.07) m and mass 79.60 (15.27) kg. 6 bowlers had a front-on orientation, mean (±SD) age was 23.17 (3.97) years, height 1.80 (0.06) m and mass 78.50 (6.77) kg. All participants were right handed and classified as fast bowlers by their club coach. Participants were recruited through amateur cricket clubs in the county league structure.

Inclusion Criteria for all studies

All participants needed to be classified as a fast bowler by a qualified cricket coach and aged between 18-40 years for senior fast bowlers. Adolescent fast bowlers needed to be between 11-18 years old, as this encompasses all adolescent fast bowlers able to participant in 'hard-ball' cricket. All fast bowlers must have at least 3 years playing experience (regular training and match schedule) in order to be classified as a trained bowler. Adolescent fast bowlers may class non hard-ball cricket as training and playing experience. All participants were free of injury at the time of testing and gave informed written consent to take part in the study. Ethical approval for the study was gained through Bournemouth University.

3.3.2 Statistical Analysis

Study 4.1 - Reliability and validity

Reliability

Accelerations

Average measures intra-class correlation coefficients (ICC), standard error of measurement (SEM) and minimum detectable change (MDC$_{95}$) calculations were carried out for a repetition of 6 deliveries per participant for all measures at BFI and FFI for both tibial and sacral accelerations. SEM and MDCs were calculated as in Equation 3.3.1 (Eliasziw et al. 1994):

$$SEM = SD\sqrt{(1\text{-}ICC)}$$

$$MDC = 1.96 * \sqrt{2}*SEM$$

Equation 3.3.1. Calculations of standard error of measurement (SEM) and minimum detectable change (MDC) taken from Eliasziw et al. (1994).

Spinal Kinematics

Average measures intraclass correlation coefficients (ICC), standard error of measurement (SEM) and minimum detectable change (MDC) calculations were carried out for lumbar, thoracic and thoracolumbar variables (Eliasziw et al. 1994). This provides a measure of consistency and variability for the peak ROM values only. Therefore, in addition the coefficient of multiple correlation (CMC) and root mean square error (RMSE) were also calculated for lumbar, thoracic and thoracolumbar kinematics between BFI and FFI to provide a measure of consistency and variability of the movement behaviour across time for the whole delivery stride (Ferrari et al. 2010).

Validity

All tibial accelerations and ground reaction force data were normally distributed. Thus, the relationship between tibial accelerations and GRF was assessed via Pearson's correlations. A Bonferroni correction for multiple comparisons was applied, resulting in an alpha of $p<.003$. Correlations were assessed on a ball-by-ball basis, therefore 30 balls were compared. Correlations were run on comparative measures. For example,

peak vertical GRF was compared to peak along-tibial acceleration. All peaks and time to peak variables were compared.

As spinal kinematics between Vicon and inertial sensors are expressed in the same metric, further comparisons were able to be carried out on this data. As with the impact data, Pearson's correlations were conducted. Additionally, a one-way ANOVA was used to look for differences between the measurements of both devices. An alpha level of $p<0.05$ was used for all tests. Additionally, mean bias (mean difference between measurements) and root mean square error of prediction (RMSEP) were calculated as in Equation 3.3.2 (Wundersitz et al. 2015).

$$RMSEP = \sqrt{\frac{(Inertial\ Sensor\ Measurement - Vicon\ Measurement)^2}{Number\ of\ observations}}$$

Equation 3.3.2. Calculation of root mean square error of prediction as in Wundersitz et al. (2015).

Study 4.2 - Playing surfaces and lower limb impacts

Impactor Comparisons

After checks for normality (Shapiro-Wilk) and sphericity (Mauchly's), a one-way ANOVA with a Tukey's post hoc test was used to determine if any significant differences existed between group mean peak and time-to-peak acceleration of the impactor across the four playing surfaces. An ANOVA was also used to determine any differences in peak and time-to-peak acceleration between the 12 popping crease locations. Furthermore, absolute agreements for peak and time-to-peak accelerations between all impactor tests were calculated using average measures intra-class correlation coefficients (ICC) with standard error of measurement (SEM) and minimal detectable change (MDC) also calculated for peak and time-to-peak acceleration (Eliasziw et al. 1994).

Tibial Accelerations

A one-way ANOVA with a Tukey's post hoc test was used to determine if any significant differences existed between group mean peak ('along-tibial' and resultant) acceleration and time-to-peak tibial acceleration across the four playing surfaces, as well as any differences in normalised peak acceleration and average loading rate.

Average measures ICC, SEM and MDC calculations were also carried out for peak and time-to-peak acceleration, as well as normalised peak acceleration and average loading rate (Eliasziw et al. 1994). An alpha level of 0.05 was accepted for all statistical tests. Effect sizes were also calculated comparing impact characteristics on different surfaces. An effect size ≥0.8 was classed as large, with *d* ≥0.5 being classed as moderate and *d* ≥0.2 as small. Anything less than 0.2 is classed as trivial (Sullivan and Feinn, 2012).

Study 4.3 - Shoulder counter-rotation and three-dimensional spinal kinematics comparison

Data were not normally distributed, therefore a series of Spearman's pairwise correlations were performed to explore the relationship between mean SCR, HSS and spinal kinematics for each bowler (n=35). A Bonferroni correction for multiple significance testing was applied resulting in an alpha of $p<0.003$.

Study 5.1 - Junior and senior fast bowlers comparison

Not all variables resulted in normally distributed data.

Multiple log or reflected log (depending on skewness) transformations were applied to data that were not normally distributed, however some variables remained non-normally distributed. Independent t-tests were used to compare tibial and sacral accelerations and lumbar, thoracic and thoracolumbar kinematics between junior and senior groups or a Mann-Whitney U test where data were not normally distributed following transformation. A Bonferroni correction for multiple comparisons was applied, resulting in an alpha of $p<.001$. Furthermore, effect sizes (Cohen's *d*) were calculated for differences in impact characteristics and spinal kinematics between senior and junior fast bowler groups at back and front-foot impact. The same effect size constructs were used as in previous methods in this thesis (Sullivan and Feinn, 2012).

Study 5.2 - Fast bowling biomechanics and lower back pain risk

Not all variables resulted in normally distributed data.

Multiple log or reflected log (depending on skewness) transformations were applied to data that were not normally distributed, however some variables remained non-normally distributed. Independent t-tests were used to compare tibial and sacral accelerations and lumbar, thoracic and thoracolumbar kinematics between groups with and without a

history of LBP or a Mann-Whitney U test where data were not normally distributed following transformation. A Bonferroni correction for multiple t-tests was applied, resulting in an alpha of p<.001. Furthermore, effect sizes (Cohen's *d*) were calculated, as in previous methods in this thesis, for differences in impact characteristics and spinal kinematics between 'LBP history' and 'no LBP history' groups for the retrospective analysis (Sullivan and Feinn, 2012). This same statistical approach was conducted after bowlers were re-grouped based on whether they developed LBP in the 2015 season, the prospective analysis.

Study 6.1 - Fast bowling biomechanics and ball release speed

All data were normally distributed, therefore a stepwise multiple regression was performed to explore the relationship between mean tibial and sacral accelerations, spinal kinematics and ball release speed for each bowler (n=13). An alpha of p<0.05 was set. Furthermore, a one-way ANOVA was performed to analyse differences in fast bowling impacts and spinal kinematics between groups with either a 'front-on' or 'side-on' back leg technique.

Chapter 4

Reliability, Validity and Supporting Studies

4.1 The reliability and validity of accelerometry and inertial sensors for measurement of impacts and three-dimensional spinal kinematics during fast bowling in cricket. A pilot study.

4.1.1 Introduction

Previously in this thesis an in-depth analysis of current literature pertaining to fast bowling lower limb kinetics, as well as methodologies used for their analyses was presented. Current analysis of fast bowling kinetics has been limited to force plate analysis (Elliott and Foster, 1984; Foster and Elliott, 1985; Foster et al. 1989; Mason et al. 1989; Elliott et al. 1992; Hurrion et al. 2000; Portus et al. 2004; Crewe et al. 2013; Worthington et al. 2013). Whilst this enables accurate analysis of the ground reaction force (GRF), it is limited to the laboratory environment and is inherently expensive, therefore a more portable and cost effective method of kinetic analysis is needed if routine take-up at club level is desired (Hurrion et al. 2000; Portus et al. 2004; Stuelcken and Sinclair, 2009; Crewe et al. 2013; Worthington et al. 2013; Spratford and Hicks, 2014; Bayne et al. 2016; King et al. 2016; Middleton et al. 2016). This would enable coaches to provide real-time feedback in representative bowling environments.

Accelerometers at the tibia and sacrum have previously been validated as a representative measure of kinetic variables, such as peak and time-to-peak acceleration during high impact movements including running, jumping and falling (Crowell et al. 2010; Theobald et al. 2010; Tran et al. 2010; Sell et al. 2014; Henriksen et al. 2004). Tran and colleagues (2010) report that accelerometer data resulted in acceptable measurement error and showed moderate correlations when compared with GRF data in jumping and landing tasks, suggesting accelerometers may be a valid method for measuring impacts in the field (Tran et al. 2010). Furthermore, running literature has highlighted a very strong relationship between tibial accelerations recorded by tibial mounted accelerometers and GRF obtained using force plates ($r^2 = 0.95$) (Hennig et al. 1993). Furthermore, the use of both tibial and sacral accelerometry has been utilised to describe impact attenuation during running (Mizrahi et al. 2000). Whilst accelerometers are becoming a more commonly utilised method, they have not yet been used for the analysis of lower limb impact during fast bowling. Prior to the uptake of any new technology, a reliability and validity analysis is warranted. Variables such as peak and time-to-peak acceleration at the tibia and sacrum may provide important information in

relation to impact during fast bowling, as well as knowledge regarding impact attenuation, providing coaches and practitioners with insights into the accelerations experienced during fast bowling.

Three-dimensional spinal kinematics during fast bowling have been reported in a number of studies with reference to performance and injury (Burnett et al. 1998; Ranson et al. 2008; Ferdinands et al. 2009; Ranson et al. 2009; Stuelcken et al. 2010; Crewe et al. 2013; Bayne et al. 2016). Whilst studies have drawn individual conclusions based on their findings, limitations and heterogeneity of current methodologies have made collective synthesis difficult. Studies reporting three-dimensional spinal kinematics in fast bowling typically use multi-camera optoelectronic motion analysis systems (Burnett et al. 1998; Ranson et al. 2008; Ferdinands et al. 2009; Ranson et al. 2009; Stuelcken et al. 2010; Crewe et al. 2013; Bayne et al. 2016). These systems allow the collection of a wealth of data, to a high degree of accuracy (Windolf et al. 2008). However, the fact that these systems are expensive and typically limited to a laboratory environment prevent the routine live analysis of bowling kinematics. Thus, alternative technologies that overcome previous limitations are desirable.

Burnett and colleagues (1998) reported on the use of an electromagnetic tracking device for the analysis of three-dimensional spinal kinematics during fast bowling. Whilst this overcomes some of the limitations of optoelectronic systems such as line of sight and cost, the operating volume for electromagnetic systems are small unless the electromagnetic source is attached to the individual (as in this study). Using the source as a sensor yields a comparatively large and heavy sensor resulting in significant inertial properties during such a ballistic task like fast bowling and may limit or alter the bowler's natural movement.

Inertial sensor technology, consisting of gyroscopes, accelerometers and magnetometers, have been validated for the use in clinical analysis of three-dimensional spinal kinematics and more dynamic sporting movements (Charry et al. 2011; van den Noort et al. 2009; Hu et al. 2014; Williams et al. 2013; Williams et al. 2014; Swaminathan et al. 2016). As inertial sensors are not dependant on cameras or line of sight, they offer the potential for 'in-field' data collection, whilst being a smaller and lighter option to electromagnetic systems. Previous studies reporting spinal kinematics from inertial sensors show very strong correlation with values reported from

electromagnetic systems in clinical settings (as high as $R^2 = .999$) (Ha et al. 2013). Furthermore, a review analysing the validity of using inertial sensors for human movement highlighted good validity and reliability but also acknowledged that this is task specific (Cuesta-Vargas et al. 2010). The validity and reliability of inertial sensors for the analysis of three-dimensional spinal kinematics and impacts during fast bowling has not been previously investigated. Therefore, before this technology can be recommended for analysis of fast bowling spinal kinematics a reliability and validity analysis is warranted.

4.1.2 Aim of the Study

This study aimed to assess the reliability and validity of using accelerometry and inertial sensors to measure fast bowling impacts and three-dimensional spinal kinematics during fast bowling in cricket.

4.1.3 Results

Validity

Pearson's correlations highlighted significant correlations ($p<0.003$) in 79% of all compared acceleration and GRF variables at both BFI and FFI (See tables 4.1.1, 4.1.2 and 4.1.3). Strong to very strong correlations ($r>0.7$ and $r>0.9$ respectively) were observed in all variables except time to peak resultant acceleration and GRF which highlighted a significant moderate correlation ($r=.640$).

Table 4.1.1. Comparison and correlation of mean tibial acceleration and ground reaction force at back-foot impact of n=30 deliveries.

GRF Variable at BFI	Mean (±SD)	Accelerometer Variable at BFI	Mean (±SD)	r
Vertical peak GRF (N)	1738.4 (391.2)	Along-tibial peak acceleration (g)	14.1 (6.6)	.974*
Anterior-posterior peak GRF (N)	845.8 (138.1)	Anterior-posterior peak acceleration (g)	11.7 (6.4)	.977*
Mediolateral peak GRF (N)	254.2 (150.8)	Mediolateral peak acceleration (g)	3.5 (3.2)	.966*
Resultant peak GRF (N)	1875.5 (379.8)	Resultant peak acceleration (g)	20.4 (9.4)	.968*
Time to peak vertical GRF (ms)	30.4 (16.8)	Time to peak along-tibial acceleration (ms)	25.8 (8.5)	.979*
Time to peak resultant GRF (ms)	34.5 (15.4)	Time to peak resultant acceleration (ms)	22.4 (9.4)	.767

*Denotes p<0.003

Table 4.1.2. Comparison and correlation of mean tibial acceleration and ground reaction force at front-foot impact of n=30 deliveries.

GRF Variable at FFI	Mean (±SD)	Accelerometer Variable at FFI	Mean (±SD)	r
Vertical peak GRF (N)	3072 (921.9)	Along-tibial peak acceleration (g)	30.9 (14.4)	.871*
Anterior-posterior peak GRF (N)	604.6 (587.3)	Anterior-posterior peak acceleration (g)	23.5 (8.4)	.860*
Mediolateral peak GRF (N)	405.2 (388.0)	Mediolateral peak acceleration (g)	16.7 (8.4)	.878*
Resultant peak GRF (N)	3206.7 (965.1)	Resultant peak acceleration (g)	46.4 (20.8)	.946*
Time to peak vertical GRF (ms)	15.7 (10.1)	Time to peak along-tibial acceleration (ms)	18.2 (3.2)	.772
Time to peak resultant GRF (ms)	15.8 (10.1)	Time to peak resultant acceleration (ms)	16.6 (2.8)	.640

*Denotes p<0.003

Pearson's correlations, highlight strong to very strong correlations across all lumbar kinematics at both BFI and FFI. As metrics for lumbar kinematics were identical between devices a one-way ANOVA was also conducted to investigate if any differences in measurements are present. Lumbar rotation was significantly larger at FFI when using Vicon (p = 0.029). Mean bias highlighted inertial sensor data (IMU) overestimated kinematics between 1.9-4° (Table 4.1.3). The largest difference was seen in lumbar rotation at FFI which displayed mean bias of -5° (negative values denote higher values in the Vicon data compared with IMU). Consequently, root mean square error of prediction (RMSEP) ranged from 0.3-1.5°.

Table 4.1.3. Comparison and correlation of mean spinal kinematics, mean bias and RMSEP between inertial sensors and optoelectronic motion analysis at back and front-foot impact of n=30 deliveries.

Variable	Vicon (°±SD)	IMU (°±SD)	r	Mean Bias (°)	RMSEP(°)
Shoulder counter-rotation	24.9 (7.7)	24.0 (7.7)	.948*	-0.9	0.3
Lumbar Flexion at BFI	5.7 (5.6)	7.5 (4.7)	.986*	1.9	0.5
Lumbar Lateral Flexion at BFI	5.8 (2.1)	9.8 (6.6)	.949*	4.0	1.2
Lumbar Rotation at BFI	10.3 (6.4)	12.1 (9.9)	.612	1.8	0.5
Lumbar Flexion at FFI	13.6 (8.8)	17.3 (5.0)	.958*	3.6	1.1
Lumbar Lateral Flexion at FFI	10.8 (10.9)	13.9 (7.2)	.954*	3.2	0.9
Lumbar Rotation at FFI	21.2 (7.5)	16.1 (7.3)	.846*	-5.1	1.5

*Denotes p<0.003

Reliability

4.1.3.1 Tibial acceleration

Tibial acceleration characteristics at BFI and FFI can be seen in table 4.1.4. ICCs for peak tibial acceleration in all three planes and resultant tibial acceleration demonstrate very high agreement for repeated trials at BFI and FFI (table 4.1.5). ICCs for time-to-peak resultant acceleration at BFI also displayed very high agreement. Time-to-peak acceleration along the *x*-axis at BFI and FFI displayed a strong agreement, whilst time-to-peak resultant acceleration at FFI displayed a moderate agreement. As ICCs for all measures demonstrate moderate to very strong agreement, corresponding SEM and MDC measures were low (as seen in table 4.1.5) suggesting that a change greater than 3.4g for along tibial acceleration represents a change greater than the natural variation observed during repeated bowling. Likewise, a change of more than 16.0ms for along tibial time-to-peak impact representing a change greater than natural variation. Furthermore, a change greater than 5.4g or 11.4ms for the same variables at front-foot impact represents true change.

4.1.3.2 Sacral acceleration

Sacral acceleration characteristics at BFI and FFI can be seen in table 4.1.4. ICCs for all variables at back-foot impact display strong to very strong agreement. ICCs for peak sacral acceleration in all three planes at FFI display moderate to strong agreement, whilst peak resultant acceleration and time-to-peak vertical acceleration displaying low to moderate agreement. While some ICCs were not as high for sacral acceleration compared with tibial acceleration, all SEM and MDC values were low (table 4.1.5) suggesting a change of 0.9g or 31ms in peak or time-to-peak vertical or resultant acceleration, representing true change at back-foot impact. Additionally, a change in excess of 1.5g or 32.0ms in the same variables represent true change at FFI.

Table 4.1.4. Mean (±SD) tibial and sacral accelerations during back and front-foot impact of n=35 fast bowlers.

Tibial Acceleration	Back-Foot Impact	Front-Foot Impact
Peak Tibial Acc x (g)	12.42 (5.57)	25.91 (11.31)
Peak Tibial Acc y (g)	4.29 (3.7)	12.42 (8.21)
Peak Tibial Acc z (g)	15.85 (8.76)	20.31 (11.91)
Resultant Tibial Acc (g)	20.11 (7.80)	35.17 (15.26)
Time-to-peak Tibial Acc x (ms)	25.47 (11.10)	20.92 (10.39)
Time-to-peak Resultant Tibial Acc (ms)	54.59 (21.80)	58.29 (13.48)
Mean Tibial Loading Rate x (g.s^{-1})	619.17 (309.45)	1597.59 (852.30)
Mean Resultant Tibial Loading Rate (g.s^{-1})	438.64 (210.46)	754.55 (527.42)
Normalised Peak Tibial Acc x (g.kg^{-1})	0.16 (0.07)	0.33 (0.13)
Normalised Resultant Tibial Acc (g.kg^{-1})	0.25 (0.10)	0.44 (0.16)
Sacral Acceleration		
Peak Vertical Sacral Acc (g)	2.73 (0.73)	3.22 (0.57)
Peak Mediolateral Sacral Acc (g)	0.91 (0.67)	0.98 (0.61)
Peak Anterior-Posterior Acc (g)	1.45 (0.77)	1.48 (0.78)
Resultant Sacral Acc (g)	3.23 (0.83)	3.88 (0.73)
Time-to-peak Vertical Sacral Acc (ms)	70.71 (18.58)	64.83 (13.33)
Time-to-peak Resultant Sacral Acc (ms)	73.64 (20.71)	68.13 (15.15)
Mean Sacral Vertical Loading Rate (g.s^{-1})	45.22 (17.74)	60.47 (17.21)
Normalised Peak Vertical Sacral Acc (g.kg^{-1})	0.04 (0.01)	0.04 (0.01)
Resultant Attenuation (%)	81.62 (7.45)	86.29 (6.32)

Acc, acceleration; g, gravity; ms, milliseconds; s, seconds; kg, kilograms.

Table 4.1.5. Repeated measures reliability of tibial and sacral acceleration at back and front-foot impact during one over (6 balls) of fast bowling (n=35)

	Back-Foot Impact			Front-Foot Impact		
	ICC	SEM	MDC	ICC	SEM	MDC
Tibial Acceleration						
Peak Tibial Acc x	0.95	1.22g	3.38g	0.97	1.96g	5.43g
Peak Tibial Acc y	0.93	1.01g	2.80g	0.93	2.19g	6.07g
Peak Tibial Acc z	0.96	1.80g	4.99g	0.95	2.79g	7.73g
Resultant Tibial Acc	0.97	1.42g	3.94g	0.95	3.31g	9.17g
Time-to-peak Tibial Acc x	0.73	5.78ms	16.02ms	0.84	4.10ms	11.36ms
Time-to-peak Resultant Tibial Acc	0.90	7.30ms	20.23ms	0.53	9.24ms	25.61ms
Sacral Acceleration						
Peak Vertical Sacral Acc	0.81	0.32g	0.89g	0.68	0.32g	0.89g
Peak Mediolateral Sacral Acc	0.92	0.20g	0.55g	0.77	0.30g	0.83g
Peak Anterior-Posterior Acc	0.82	0.33g	0.91g	0.57	0.51g	1.41g
Resultant Sacral Acc	0.83	0.34g	0.94g	0.43	0.55g	1.52g
Time-to-peak Vertical Sacral Acc	0.73	9.67ms	26.80ms	0.26	11.48ms	31.82ms
Time-to-peak Resultant Sacral Acc	0.79	11.26ms	31.21ms	0.52	9.84ms	27.28ms

ICC, intraclass correlation coefficient; SEM, standard error of measurement; MDC, minimum detectable change; acc, acceleration; g, gravity; ms, milliseconds.

4.1.3.2 Spinal Kinematics

Mean (SD) spinal orientations and resultant ROM for the delivery stride can be seen in table 4.1.6. Reliability statistics for all spinal kinematic variables can be seen in table 4.1.7.

Table 4.1.6. Mean (±SD) spinal orientations and range of motion during the fast bowling delivery stride (n=35).

	Lumbar			Thoracic			Thoracolumbar		
	BFI	FFI	Range	BFI	FFI	Range	BFI	FFI	Range
Flexion (°)	-14.2 (14.5)	21.3 (8.2)	35.1 (21.3)	-28.1 (18.4)	31.1 (15.1)	57.9 (28.7)	-37.3 (18.4)	42.7 (16.5)	79.2 (32.5)
Lateral Flexion (°)	-11.7 (10.0)	19.9 (8.2)	31.0 (14.1)	-15.3 (16.0)	26.7 (11.8)	41.4 (21.9)	-22.4 (14.5)	30.7 (12.0)	53.1 (20.4)
Rotation (°)	-2.6 (8.5)	14.3 (7.3)	16.9 (11.9)	-9.9 (11.1)	19.3 (12.2)	29.2 (16.9)	-9.2 (16.7)	26.8 (12.3)	36.0 (19.7)
SCR									36.9 (9.6)
HSS									34.9 (18.0)

BFI, back-foot impact; FFI, front-foot impact; SCR, shoulder counter-rotation; HSS, hip-shoulder separation; °, degrees. Flexion, left lateral flexion and left rotation were defined as positive.

ICCs for all kinematic variables demonstrated good (>0.6) to excellent (>0.75) agreement for repeated trials. Consequently, corresponding SEM and MDC measures were low illustrating minimal intra-individual movement variation across the six bowls. Thoracolumbar lateral flexion displayed a higher SEM and MDC, potentially due to greater intra-individual variability and ICCs were also lowest for this variable.

All CMCs demonstrated good agreement and RMSEs were low suggesting the delivery stride curves were of a similar shape for the repeated movement cycles. Examples of CMC analysis can be seen in the figures 4.1.1-4.1.9.

Table 4.1.7. Repeated measures reliability of spinal range of motion between back and front-foot impact for one over (6 balls) of fast bowling (n=35).

	ICC	SEM (°)	MDC (°)	CMC	RMSE (°)
Shoulder counter-rotation	0.72	2.66	7.37		
Hip-shoulder separation	0.77	7.88	21.84		
Lumbar flexion	0.93	4.02	11.14	0.63	3.93
Lumbar lateral flexion	0.64	4.46	12.36	0.71	2.92
Lumbar rotation	0.67	4.82	13.36	0.70	4.32
Thoracic flexion	0.69	8.57	23.75	0.62	4.40
Thoracic lateral flexion	0.83	7.88	21.84	0.69	3.31
Thoracic rotation	0.90	7.13	19.76	0.67	4.81
Thoracolumbar flexion	0.75	8.51	23.59	0.65	4.73
Thoracolumbar lateral flexion	0.63	20.68	57.32	0.70	2.64
Thoracolumbar rotation	0.96	6.63	18.38	0.71	5.47

ICC, intraclass correlation coefficient; SEM, standard error of measurement; MDC, minimum detectable change; °, degrees; CMC, coefficient of multiple correlation; RSME, root mean square error.
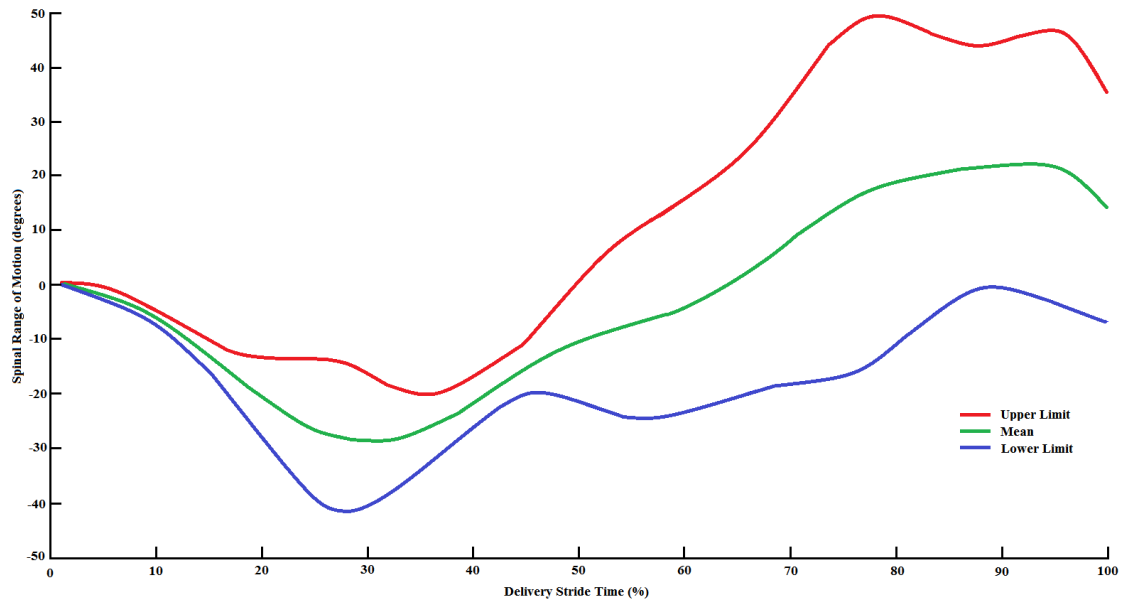
Figure 4.1.1. Mean lumbar flexion during the fast bowling delivery stride and 95% upper and lower confidence intervals (Flexion = Positive).
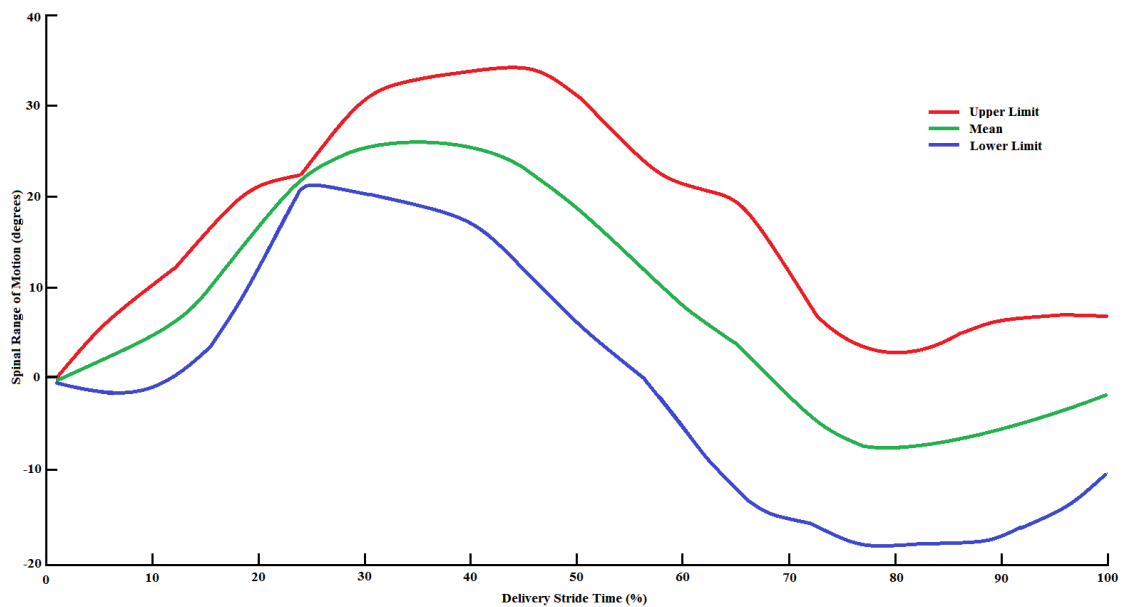
Figure 4.1.2. Mean thoracic flexion during the fast bowling delivery stride and 95% upper and lower confidence intervals (Flexion = Positive).
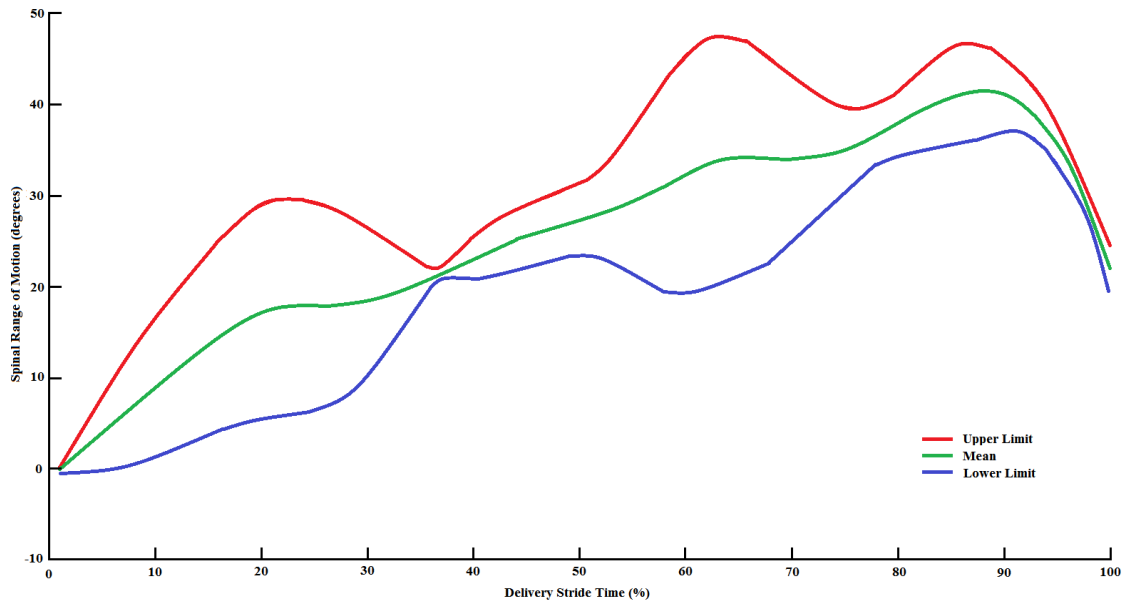
Figure 4.1.3. Mean thoracolumbar flexion during the fast bowling delivery stride and 95% upper and lower confidence intervals (Flexion = Positive).
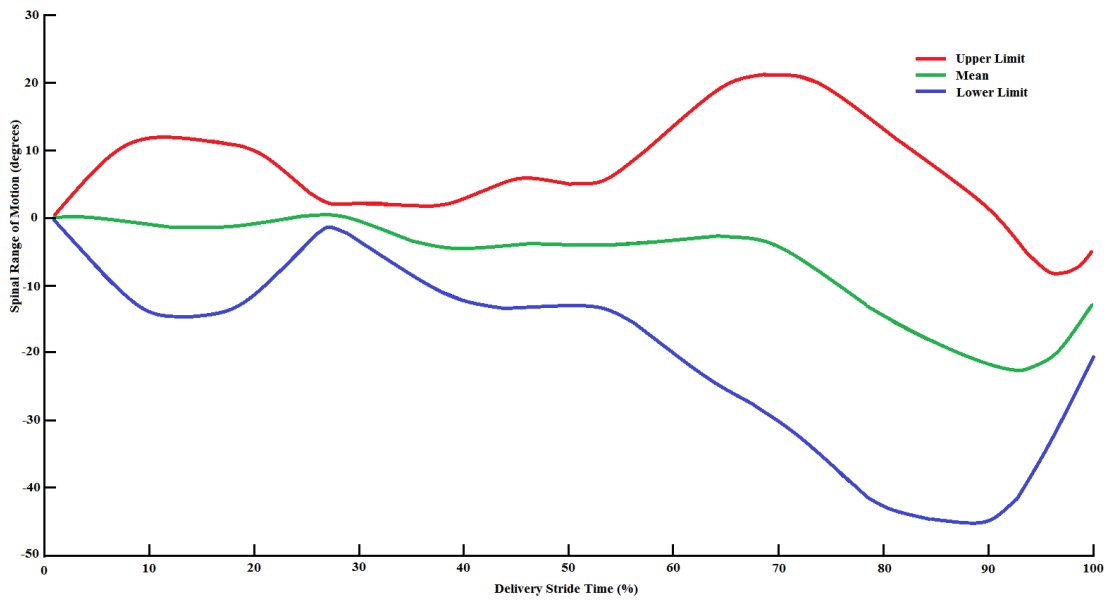
Figure 4.1.4. Mean lumbar lateral flexion during the fast bowling delivery stride and 95% upper and lower confidence intervals (Right lateral flexion = Positive).
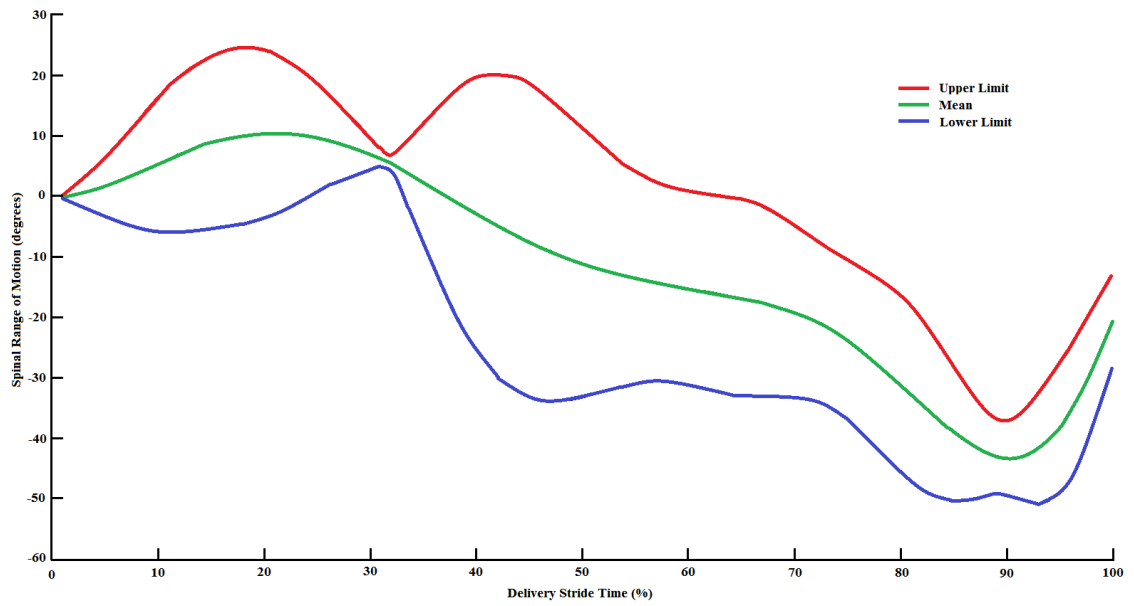
Figure 4.1.5. Mean thoracic lateral flexion during the fast bowling delivery stride and 95% upper and lower confidence intervals (Right lateral flexion = Positive).
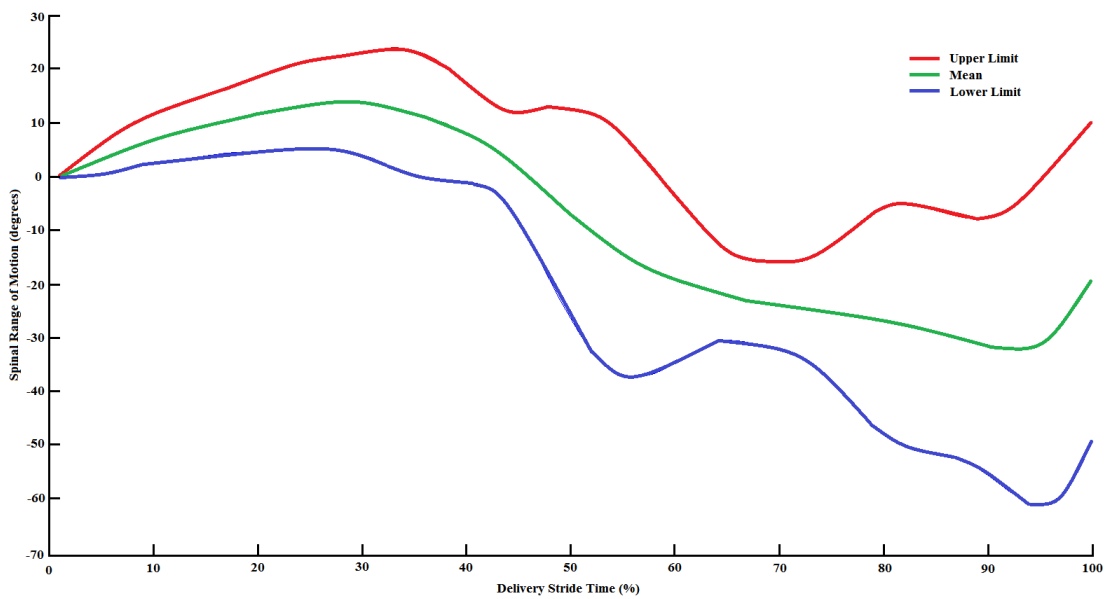
Figure 4.1.6. Mean thoracolumbar lateral flexion during the fast bowling delivery stride and 95% upper and lower confidence intervals (Right lateral flexion = Positive).
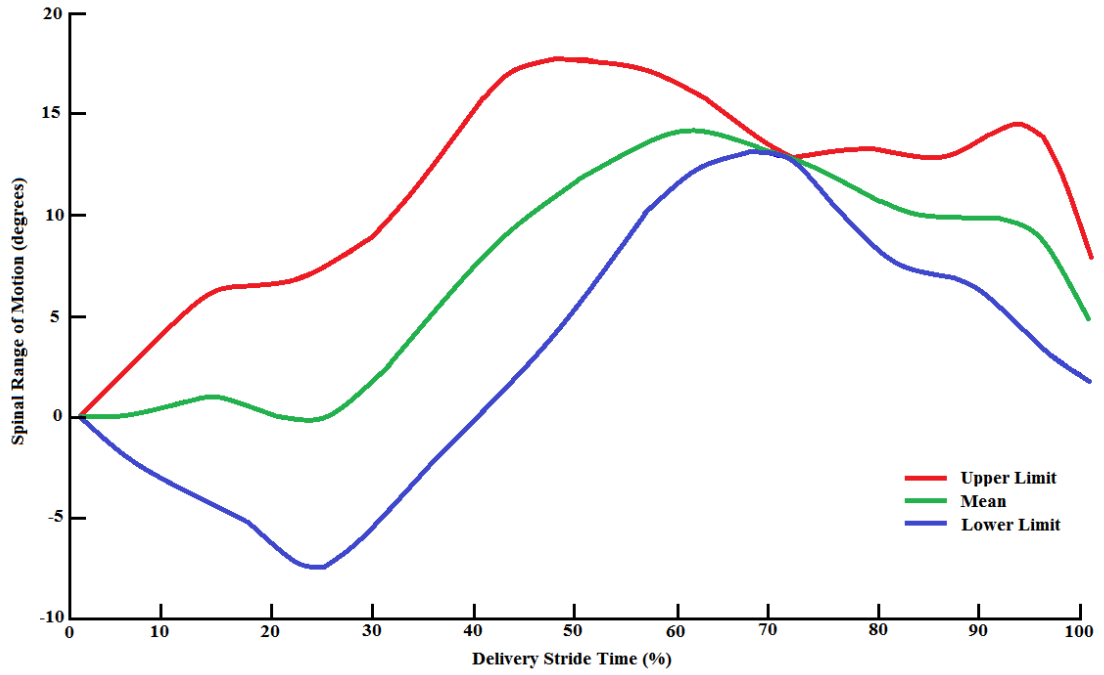
Figure 4.1.7 Mean lumbar rotation during the fast bowling delivery stride and 95% upper and lower confidence intervals (Right Rotation = Positive).
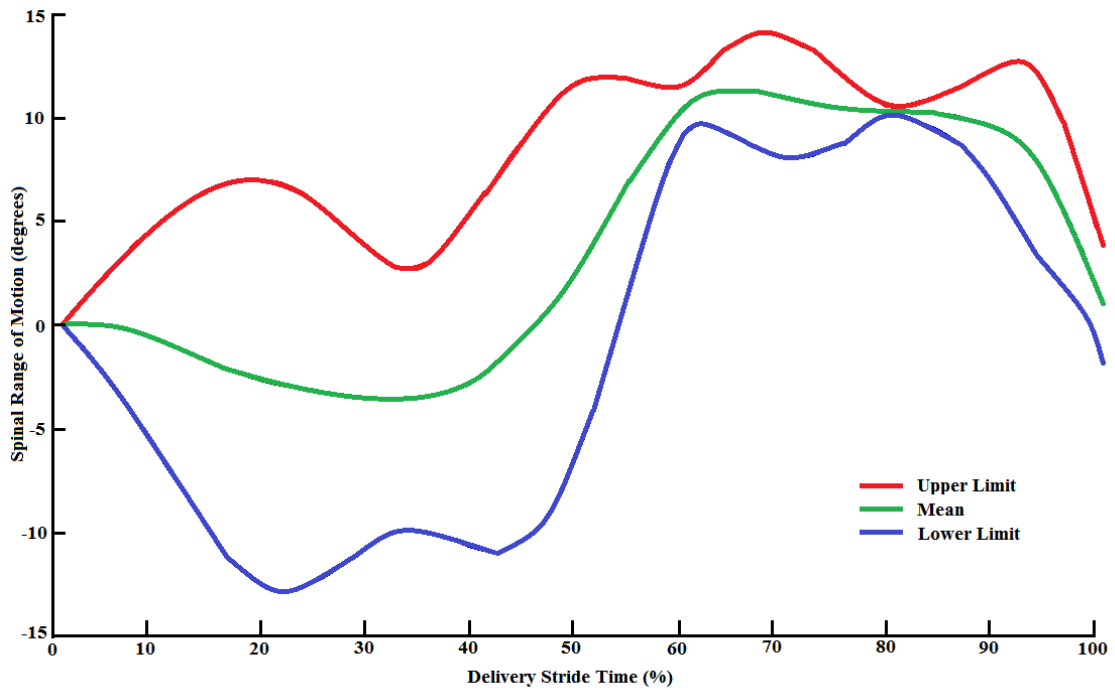
Figure 4.1.8. Mean thoracic rotation during the fast bowling delivery stride and 95% upper and lower confidence intervals (Right Rotation = Positive).
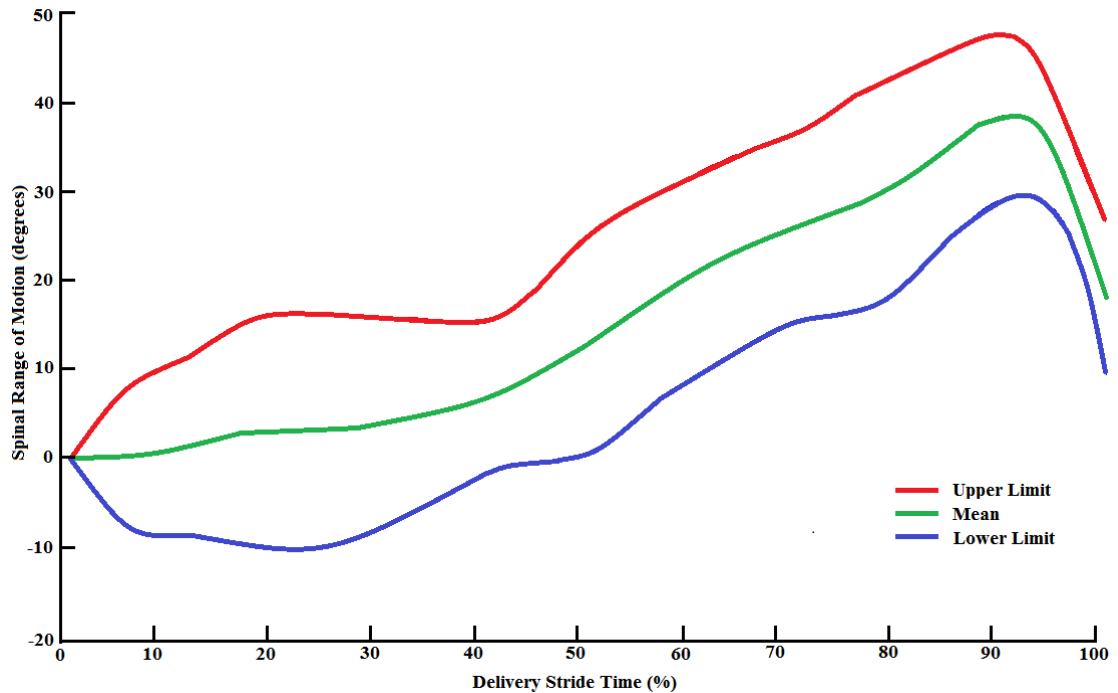
Figure 4.1.9. Mean thoracolumbar rotation during the fast bowling delivery stride and 95% upper and lower confidence intervals (Right Rotation = Positive).

4.1.4 Discussion

This study aimed to assess the reliability and validity of accelerometry and inertial sensors for analysis of impact variables and three-dimensional spinal kinematics between BFI and FFI during cricket fast bowling. Previous studies have reported kinematic and kinetic variables using force plate and three-dimensional optoelectronic motion analysis technology (Hurrion et al. 2000; Portus et al. 2004; Stuelcken and Sinclair, 2009; Crewe et al. 2013; Worthington et al. 2013; Spratford and Hicks, 2014; Bayne et al. 2016; King et al. 2016; Middleton et al. 2016). However, due to methodological restrictions of this technology, most studies have analysed bowling in a laboratory environment (Elliott and Foster, 1984; Foster and Elliott, 1985; Foster et al. 1989; Mason et al. 1989; Elliott et al. 1992; Portus et al. 2004; Crewe et al. 2013; Worthington et al. 2013). Whilst this may not necessarily provide space restrictions the expense and location of such set-ups may provide barriers to many within the target population. To the author's knowledge, this is the first study to use accelerometers and

inertial sensors to analyse fast bowling impacts and spinal kinematics, and therefore the first study to measure fast bowling impacts with fewer environmental constraints.

<u>4.1.4.1 Tibial acceleration</u>

<u>*Validity*</u>

Lower limb impacts have been well reported in previous literature (Crewe et al. 2013; Worthington et al. 2013; Stuelcken et al. 2009). However, all impacts have been described using force plates; whilst this is considered the gold standard, it does typically limit testing to a laboratory environment and are expensive. Tibial accelerometry addresses these limitations and as such, may provide an alternative method of describing lower limb impacts for in-field fast bowling. However, as no previous studies have utilised this method a validity analysis is warranted. The results of this study demonstrate a strong relationship between tibial accelerations at BFI and FFI and the corresponding GRF variables. As force plates and accelerometers produce different metrics a comparison of results via a Pearson's correlation was considered the most appropriate approach.

Comparison of peak GRF and peak accelerations at BFI produced correlations of r=.966-.977, display very strong relationships between metrics. Time to peak vertical force and acceleration also displayed a very strong relationship with r=.979. Time to peak resultant force and acceleration was slightly weaker at r=.767, however this is still a strong correlation. These values are in line with those reported in previous literature comparing accelerometers and force plates in running and stronger than some reported in jumping tasks (Simons and Bradshaw, 2016; Raper et al. 2018).

Similar relationships are observed in the comparison of GRF and tibial accelerations at FFI. Peak GRF and accelerations displayed strong to very strong correlations with r=.860-.946. Time to peak vertical GRF and acceleration displayed r=.772 and time to peak resultant GRF and acceleration r=.640. These relationship are similar, yet slightly weaker than those displayed at BFI, this may be due to the increased magnitudes seen at FFI. Higher magnitudes of impacts are likely to elicit greater movement variability in order to find a mechanism to dissipate the additional load experienced (Bartlett, Wheat & Robins, 2007). Consequently, this additional variability may also result in measurement variability between methods. However, even with slightly weaker correlations seen at FFI, they are still in line with previously reported values.

The above findings suggest accelerometry is a valid alternative to force plates for the measurement of impacts in fast bowling. Nevertheless, measurement differences that may produce discrepancies between methods must be acknowledged. Due to the absence of a gyroscope in the tibial accelerometer, corrections for tilt cannot be applied. As a result force plate and accelerometer axes may not directly correspond. This may produce more discrepancies in movements more multidirectional than fast bowling. However, the use of resultant acceleration and force is able to overcome this issue and still give a valid measure of impact. The slightly lower values seen in time to peak values between devices may be due to the lack of mass element in the accelerometer readings, this may result in a slight phase shift or altered pattern when compared to force data if weight of the bowler is distributed differently. Due to the inability to time synchronised the two devices this relationship was not able to be explored further. Additionally, due to the uncertainty in mass distribution throughout the bowl, no attempt was made to calculate predicted force from accelerometer readings. This was due to the likelihood of inducing errors that may be deemed unnecessary due to the strong correlation between force and accelerations in this study.

Reliability

The results of this study demonstrate that accelerometry is a reliable method of measuring tibial accelerations during real-time, in-field cricket fast bowling. All variables investigated demonstrated good to excellent reliability, with the weakest reliability reported for time-to-peak resultant tibial acceleration at FFI (0.53). This is likely to be a result of the increased variability at FFI discussed above. Previous studies investigating reliability of tibial accelerometry report ICC values of 0.64-0.97 for walking and 0.82 for running, consequently tibial accelerometry for the analysis of fast bowling impacts are as reliable as other previously reported tasks (Turcot et al. 2008; Raper et al. 2018). In addition, ICCs in this study are consistent with those reported for landing tasks using force plates, which are typically reported as >0.8 (Walsh et al. 2006). However, no previous research has reported reliability of impact variables during fast bowling. It may be hypothesised that more variation in impacts (as a result of technique variations) will be observed during bowling compared with a less complex skill such as a drop landing. Thus, the fact that ICCs remain in line with values seen in less complex skills is encouraging. ICCs alone can only explore the relationship between repeated measurements, thus further metrics are needed to assess whether

accelerometers are usable for in-field measurement of fast bowling impacts. Consequently, SEM and MDC values were calculated to assess measurement variability and sensitivity of repeated measurements, in their specific units, to evaluate what may be considered a real change outside of natural variation in performance or measurement error.

SEM values for all measures were small (≤1.8g or 7.3ms at BFI and ≤3.3g or 9.2ms at FFI) demonstrating there was little variation in repeated testing for all tibial acceleration variables measured. MDC values of less than 5.0g for all peak accelerations at BFI and less than 9.2g at FFI highlight that accelerometers can detect small changes in tibial accelerations during fast bowling and consequently provide a reliable measure of in-field analysis of fast bowling impacts. Furthermore, MDC values suggest that tibial accelerometry is appropriate to identify actual change in performance, outside of biological variability and measurement error; providing researchers and coaches with valuable data when implementing technique or injury prevention interventions. MDCs for time-to-peak values were reported at 16.0ms and 11.4ms for the x axis at BFI and FFI respectively, with resultant acceleration time-to-peak at 20.2ms and 25.6ms respectively, slightly less sensitive than peak values but still able to detect changes of a fraction of a second.

The present study reports mean (±SD) time-to-peak tibial acceleration along the x axis at 20.92 (±10.39) ms, slightly faster than previous studies that report values between 26 to 90ms (Hurrion et al. 2000; Portus et al. 2004; Crewe et al. 2013; Worthington et al. 2013). However, as correction for tilt was not able to be carried out, the x axis for tibial acceleration in this study is aligned with the long axis of the tibia, which is unlikely to be aligned vertically at FFI. Consequently, these values may not be directly comparable. However, mean (±SD) time-to-peak resultant acceleration is in line with previous studies at 58.29 (±13.48) ms (Hurrion et al. 2000; Portus et al. 2004; Crewe et al. 2013; Worthington et al. 2013). To the author's knowledge, no previous studies have reported time-to-peak GRF at BFI, therefore, values reported in this study may be considered novel findings. The addition of time-to-peak acceleration data in this study may provide additional insight into the relationship between fast bowling impacts and musculoskeletal injury, which have been described as 'rate dependant' (Courtney et al. 1994; Tran et al. 1995).

Reliability

The results of this study demonstrate accelerometry is a reliable method of measuring sacral accelerations during fast bowling. All variables at BFI demonstrate excellent reliability, with the majority of variables at FFI demonstrating good to excellent reliability. Resultant sacral acceleration and time-to-peak vertical sacral acceleration demonstrated the weakest reliability with ICCs of 0.43 and 0.26 respectively. Overall, the majority of values are lower than those reported for tibial accelerations in this study, demonstrating less consistency between repeated trials. However, these values were still in line with those reported in jump landing studies of 0.74-0.94 (Picerno et al. 2011). This may be explained by the fact that sacral accelerations are recorded further up the body, with a capacity for greater variability through the knee and hip joints offering greater degrees of freedom in terms of impact attenuation. However, this may also be a result of measurement errors, such as skin artefact, as the sacral accelerometer attaches to an area with more soft tissue compared to the medial aspect of the mid-tibia. Despite the variations in ICCs, SEM values remained low for all variables at less than 0.6g or 11.5ms for all measures; demonstrating that lower repeated measures agreements highlighted by lower ICCs were most likely a result of natural variability in technique, as opposed to measurement error. This suggests that accelerometers are reliable for the measurement of sacral accelerations, but impacts experienced at the sacrum are less consistent than tibial impacts. MDC values of less than 1.5g show that accelerometers are able to detect small changes in sacral accelerations during fast bowling and consequently may provide a sensitive measure of detecting change during in-field analysis of fast bowling impacts. This may allow coaches and researchers to effectively monitor true changes in fast bowling impacts brought about by manipulation of technique or environmental factors such as footwear or playing surface. Similar to tibial acceleration, time-to-peak MDC values were slightly less sensitive to change, with values of 26.8ms at BFI and 31.8ms at FFI.

Forces experienced at the sacrum or lower back have not been reported as frequently in fast bowling literature (Crewe et al. 2013; Bayne et al. 2016). These studies have reported joint moments in the lumbar spine using three-dimensional optoelectronic motion analysis systems synchronised with force plates (Crewe et al. 2013; Bayne et al.

2016). Whilst these systems are able to provide a wealth of information, as with force plate technology used to report GRF, analysis is limited to a laboratory environment. Furthermore, no studies have reported impact characteristics of the lower back at BFI, therefore findings in the present study may be considered novel. In addition to this, studies have not analysed the reliability of these measures in fast bowling and therefore the accuracy of these findings in relation to direct measures such as the inertial sensors used in this study or in vivo measures are unknown.

4.1.4.3 Spinal Kinematics

Validity

The validity of measurement of three-dimensional spinal kinematics using inertial sensors has previously been reported in such movements as walking and running (Von Marcard et al. 2016). Strong correlations with optoelectronic systems have been reported for these movements, however, this is the first study to compare spinal kinematics in fast bowling using inertial sensors and the gold standard optoelectronic motion analysis. The inertial sensors were able to measure lumbar, thoracic and thoracolumbar kinematics as well as SCR. However, the previously validated marker set used for the optoelectronic system (full body plug-in gait) was only able to record lumbar kinematics and SCR, due to the definition of the torso segments (Guitierrez, Bartonek, Haglund-Akerlind & Saraste, 2003; Attias et al. 2015). Consequently, this study was able to directly compare three-dimensional lumbar kinematics and SCR between BFI and FFI.

Pearson's correlations between the corresponding measurements produced by the inertial sensors and the optoelectronic system showed very strong correlations in all but one variable (r=.846-.986) with lumbar rotation at BFI production an r value of .612. However, despite this slightly lower correlation a RMSEP between the two methods was only 1.2°.

In addition to a correlation analysis a one-way ANOVA highlighted only one significant difference between the two measurement methods (Lumbar Rotation at FFI). These findings highlight that typically there is a very strong agreement between the two measurement methods, however there may be some variation in rotation. Table 4.1.3 also highlights that typically the inertial sensors overestimated spinal kinematics with mean bias ranging from 1.8-4.0°, with only SCR and lumbar rotation at FFI reporting smaller values than the optoelectronic system. These differences may be attributed to a number of factors: firstly, this may be a result of different definitions of BFI and FFI and thus kinematics being taken at slightly different time points (as discussed previously regarding tibial accelerometry validity). There is also a slightly different definition of the lumbar segment between the inertial sensors and optoelectronic model and as such, range of motion may vary between the two. However, despite these

methodological differences the high level of agreement between the two measures is encouraging for the application of inertial sensors in fast bowling analysis.

Reliability

Values for three-dimensional lumbar and thoracolumbar range of motion in this study are comparable to those reported by previous research (seen in table 4.1.8) (Burnett et al. 1998; Ranson et al. 2008; Ferdinands et al. 2009; Ranson et al. 2009; Stuelcken et al. 2010; Crewe et al. 2013; Bayne et al. 2016). This study also reports thoracic range of motion not previously reported in the published literature providing novel insight into kinematics of the thoracic spine during fast bowling. The similarities in values reported suggests that inertial sensors produce similar measurement values to other established technologies and may be a viable option for measurement of three-dimensional spinal kinematics during fast bowling. The small differences in values may be explained by differences in measurement technique, as studies have used different definitions as to the spinal boundaries, different definitions of bowling phase and likely different technique employed by individual bowlers. The findings from this study demonstrate that inertial sensors are a reliable method of measuring spinal kinematics during fast bowling in cricket. Previous studies have reported moderate to good reliability using a range of differing methods. The most common method employed to study spinal kinematics during fast bowling is optoelectronic motion analysis with reported ICCs of 0.74-0.98 and SEM 1-17° (Ranson et al. 2008; Ranson et al. 2009). The results of this study show that inertial sensors offer similar levels of repeated measures reliability as those seen in this gold standard. Moreover, electromagnetic systems have also demonstrated high repeated measures reliability (CMCs >0.89) (Burnett et al. 1998) for fast bowling, slightly higher than those observed in this study. This may be due to the number of repeated bowls used, which was three compared to six for the current study, or greater movement variability demonstrated by the participants in the current study.

Table 4.1.8. Three-dimensional spinal kinematics (±SD) reported in previous research and this study.

Authors	Subjects (n)	Spinal Segment Analysed	Bowling Phase Analysed	Flexion (°)	Extension (°)	Left Lateral Flexion (°)	Left Rotation (°)
Current Study	35	S1-L1	BFI-FFI	21± 8	14± 14	20± 8	14± 7
Current Study	35	L1-T1	BFI-FFI	31± 15	28± 18	27± 12	19± 12
Bayne et al. 2016	13	L5-L1	FFC-BR	20 ± 4		11 ± 4	4± 2
	12			21 ± 5		12± 3	5± 2
Crewe et al. 2013	13	S1-L1	FFI - BR			10± 4	
	18					12± 3	
	8					11± 3	
Stuelcken et al. 2010	14	S1-T1	BFI- BR	27± 12	14± 9	42± 6	26± 6
	12			29± 10	13± 9	38± 6	27± 6
Ferdinands et al. 2009	21	S2-T10	BFI-FFI	38± 8	6± 2	16± 11	19± 2
Ranson et al. 2009	14	S1- T10	BFI-FFI		0± 7	34± 7	29± 9
Ranson et al. 2008	50	S1- T10	BFI-BR		9± 6	34± 7	32± 8
Burnett et al. 1998	20	S2- L1	BFI-FFI	48	10	30	11

°, degrees; LBP, lower back pain; BFI, back-foot impact; FFI, front-foot impact; BR, ball release.

The standard error of measurement for the lumbar spine was less than 5 degrees, and less than 9 degrees for the thoracic spine highlighting small errors associated with repeated maximal measurements. The value is constructed of error associated with the sensor combined with the human-sensor interaction as well as the natural variability of this particular task (biological variability). SEM recorded in this study are in line with

those reported by optoelectronic systems during bowling (Ranson et al. 2008; Ranson et al. 2009) further suggesting the utility of inertial sensors for fast bowling analysis.

MDC values in this study highlight that, on average, any deviation in range of motion greater than 13 degrees for the lumbar spine and 24 degrees for the thoracic spine can be interpreted as a difference attributed to actual changes in kinematics as opposed to between-trial variability or measurement error. These values are of importance to coaches and health practitioners when implementing technique interventions or monitoring kinematics that may have been altered as a result of pain (Williams et al. 2010). The above values are encouraging when looking at inertial sensors as an alternative to current methodologies. However, it is important to note that they represent peak values, at a single point in time. Whilst this is commonplace in fast bowling literature, it does not provide an understanding of the similarities in movement behaviour across the bowling stride. It must also be noted that MDC values obtained for thoracolumbar lateral flexion are high, this is a result of the larger SEM value obtained for this movement and may be a consequence of bowlers adopting different strategies of movement in bowling.

To achieve this CMC values are necessary and those calculated in the current study demonstrated good agreement for all spinal kinematics suggesting the movement patterns were highly consistent. These values demonstrate that fast bowlers in this study were able to reproduce the same movement patterns, with small degrees of variation. These findings agree with previous studies looking at fast bowling over a longer bowling spell, finding no differences in kinematics between overs; suggesting that the motion of fast bowling is one associated with high levels of internal consistency (Burnett et al. 1995). Furthermore, these findings suggest that inertial sensors are able to reliably portray the movement behaviour across time during the delivery stride. Whilst this study has aimed to align its data analysis to previously used methods, the heterogeneity of previous methods has made direct comparisons difficult. As a result, this study recommends that future studies adopt an accepted protocol for generic analysis of spinal kinematics. This may include how spinal segments are defined, as well as at which time point's kinematics will typically be reported between. Due to the small numbers of elite fast bowlers available to researchers, studies tend to suffer from small sample sizes. Standardised methodologies may allow pooling of data and consequently produce stronger conclusions.

This study has highlighted that inertial sensors may be an acceptable alternative to current systems, however there are limitations to these sensors that need to be taken into account when considering their use for this form of analysis. Some of the above studies report spinal kinematics up to ball release (Ranson et al. 2008; Stuelcken et al. 2010; Crewe et al. 2013; Bayne et al. 2016). However, due to the absence of a method to define ball release, this was not possible in the current study. Future studies may be able to synchronise camera systems with these devices or provide alternate methods to define this characteristic. Conversely, the lack of cameras may also be an advantage, as minimal set up of the testing environment is required, allowing testing to be carried out in a realistic environment with little preparation. Furthermore, the ease of use of these sensors means that minimal training is required beyond being able to identify key bony landmarks and operate basic software. However, if data involving a large number of joints is required, this process may become more logistically challenging in regards to sensor attachment. It may be at this point that optoelectronic systems may be desirable where smaller, lighter markers would make this task easier. However, in comparison to electromagnetic systems, inertial sensors offer a more portable, less environmentally constrained and cheaper alternative.

4.1.5 Conclusion

To the author's knowledge, this study is the first of its kind to use accelerometers and inertial sensors to analyse impacts at the tibia and sacrum and three-dimensional spinal kinematics during 'in-field' fast bowling. Results demonstrated that inertial sensors are a valid alternative to the current gold standards of force plate and optoelectronic motion analysis. Reliability analysis displayed that both tibial and sacral accelerometers demonstrate high reliability, comparable to those reported in GRF studies for jumping and landing tasks. Furthermore, this study demonstrates that accelerometers are able to repeatedly highlight points of interest within the delivery stride, that have been previously reported using force plate technology, to a high degree of reliability. Additionally, spinal kinematics displayed similar values to those reported in previous research. Reliability statistics are in line with previously reported values for fast bowling analysis using an optoelectronic motion analysis system (the current gold standard). As such, accelerometers and inertial sensors may be an appropriate alternatives to force plate technologies, providing a cost effective, 'in-field' measure of

impacts and spinal kinematics during fast bowling with 'real world' applications for coaches and researchers.

4.2 How does playing surface affect front-foot tibial acceleration during cricket fast bowling? A pilot study.

4.2.1 Introduction

It has been previously suggested that repetitive exposure to high magnitudes of vertical ground reaction force (GRF) may be associated with the abnormally high prevalence of lumbar stress fractures, knee and ankle injuries seen in the elite junior and senior fast bowling population (Portus et al. 2000; Johnson et al. 2012; Spratford and Hicks, 2014). Knee injuries have been reported as having the highest occurrence in first class fast bowlers, accounting for 11% of all injuries sustained, while injuries to the lower back resulted in the most games missed at 247 games over 6 seasons (Orchard et al. 2002).

Material properties of playing surfaces have been shown to significantly affect the resultant GRF (Nigg, 1983). Cricket pitches display very high firmness values in order to facilitate the required ball-surface interaction (bounce) and therefore display significantly higher surface firmness than other playing surfaces (Carre et al. 1999). Research reports mean peak football pitch firmness of 42.6 g using a clegg hammer classification system; this compares to cricket pitch values ranging between 176-388 g dependant on time of year (Canaway et al. 1990; Baker et al. 1998). Thus, if no other factors are taken into account, it may be assumed that fast bowlers experience signficantly higher GRF than a footballer during a similar movement. However, it has been demonstrated in running that despite varying surface firmness, individuals may display altered kinematics in order to accommodate the surface properties and thus no differences in GRF are observed (Hardin et al. 2004). Consequently, exploring the surface properties in isolation may be unable to provide the necessary information regarding the effects on the actual individual. To this end, studying the human-surface interaction is critical. No fast bowling research has investigated this interaction on actual cricket playing surfaces and thus, the question of how surface properties affect front-foot impact during fast bowling remains unanswered.

Peak vertical GRF has been reported between 3.8-9 times body weight (BW) at front-foot impact during fast bowling (Foster and Elliott, 1985; Mason et al. 1989; Worthington et al. 2013). It has been commonplace to investigate GRF in fast bowling using a laboratory mounted force plate (Elliott and Foster, 1984; Foster and Elliott, 1985; Elliott et al. 1986; Foster et al. 1989; Mason et al. 1989; Elliott et al. 1992;

Hurrion et al. 2000; Crewe et al. 2013; Mason et al. 2004; Portus et al. 2004; Worthington et al. 2013). Whilst force plates are a reliable and valid data collection method that enables a weath of impact characteristics to be recorded, it is limited to describing the foot-force plate interaction without the addition of other technologies. This limits environmental factors that can be analysed such as the effects of different playing surface interactions. Studies have attempted to place material over the force plate (polyflex surface and artificial grass), however no additional calculations to recalibrate the force plate were reported for GRF values to compensate for the added force plate-surface interface (Hurrion et al. 2000; Preuss and Fung, 2004; Stuelcken and Sinclair, 2009; Worthington et al. 2013). Furthermore, although it has been attempted, no simple method of effectively recording GRF on complex surfaces outside a laboratory environment has been found (Hurrion et al. 2000). Therefore, 'in-field' fast bowling cannot be effectively analysed using current methods. In order to overcome the limitations associated with force plate technology, new technologies and their application to cricket fast bowling should be explored.

The previous study in this chapter has highlighted that inertial sensors are a reliable method for measuring fast bowling impacts. Furthermore, peak resultant acceleration data measured by miniature accelerometers have been validated for measurement of tibial impacts and shown to strongly correlate with peak resultant GRF during jumping (Tran et al. 2010; Sell et al. 2014). In addition to this, studies investigating impact attenuation during activities such as running and falling have utilised accelerometry, producing reliable data with methodologies that can be implemented outside of a laboratory environment, with no 'landing area' restrictions (Crowell et al. 2010; Theobald et al. 2010). Such methodologies may be useful in a fast bowling environment, however this has not yet been explored (Bali et al. 2011). Research has highlighted bone failure risk as 'rate dependant', with exposure to increased loads over a shorter time period increasing risk of injury (Courtney et al. 1994; Tran et al. 1995). New technologies, such as accelerometers, may enable the portable measurement of key impact variables, such as peak acceleration, time-to-peak acceleration and average loading rate, which may provide researchers and coaches with key impact characteristics to identify risk of injury or variations in performance.

4.2.2 Aim of the study

This study has two phases. The aim of phase one was to classify different cricket playing surface firmness using accelerometry. The aim of phase two was to quantify impact characteristics such as peak and time-to-peak acceleration and average loading rate of the front tibia during the delivery stride of live fast bowling during bowling on different surfaces.

4.2.3 Results

4.2.3.1 Phase One: Surface Impact Testing

ICC values demonstrate high reliability for both peak acceleration and time-to-peak acceleration. Mean SEM and MDC for peak and time-to-peak acceleration across all surfaces were low. These results (seen in table 4.2.1) show a high degree of repeated measures reliability for the impactor.

Impactor results can be seen in table 4.2.2. Peak vertical accelerations were significantly larger on the rubber surface compared with astroturf, grass and wood ($F(3,284)=$ 30.347, p=.001, p<.001 and p<.001). Peak acceleration was significantly higher on the astroturf surface compared with grass and wood ($F(3,284)=$ 30.347, p<.001 and p<.001). No other surfaces were significantly different. Time-to-peak vertical acceleration was significantly faster on the rubber surface compared with astroturf, grass and wood ($F(3,284)=$ 19.999, p=.012, p<.001 and p<.001). Time-to-peak vertical acceleration was also significantly faster on the astroturf surface than grass and wood ($F(3,284)=19.999$, p<.001 and p=.025).

4.2.3.2 Phase Two – Tibial Acceleration Testing

Tibial acceleration ICC results demonstrated very high reliability for peak 'along tibial' and resultant acceleration, time-to-peak acceleration, normalised peak acceleration and average loading rate. Mean SEM was low for all measures; therefore, mean MDCs were also low. These results (seen in table 4.2.1) demonstrate a high degree of consistency for tibial acceleration during fast bowling.

Peak and time-to-peak tibial acceleration during bowling can be seen in figure 3.2.5. Results showed that peak 'along tibial' acceleration was significantly greater on astroturf compared with grass, rubber and wood ($F(3,236)=33.972$, p=.006, p<.001 and

p<.001). Tibial acceleration on the wooden surface was also significantly smaller than grass and rubber (F(3,236)= 33.972, p<.001 and p=.001). The grass and rubber surfaces were not significantly different to each other (p>0.05).

Peak resultant tibial acceleration reported greater peak acceleration on astroturf compared with grass, rubber and wood (F(3,236)=23.000, p=.034, p<.001, p<.001). Peak resultant tibial acceleration was also significantly smaller than grass and rubber (F(3,236)=23.000, p<.001, p=.024).

Peak tibial acceleration normalised for body weight (seen in table 4.2.2) mirrored the above findings. Peak tibial acceleration per kilogram of bodyweight was significantly greater on astroturf compared with grass, rubber and wood (F(3,236)= 32.981, p=.004, p<.001 and p<.001). Wood also displayed a significantly smaller peak acceleration per kilogram of body weight than grass and rubber (F(3,236)=32.981, p<.001 and p=.001). There was no significant difference between grass and rubber surfaces (p>0.05).

In addition, time–to-peak tibial acceleration was significantly longer on grass than astroturf (F(3,236)=5.231, p=.001). Average loading rate (seen in table 4.2.2) was significantly faster on astroturf compared with grass, rubber and wood (F(3,236)=8.818, p=.002, p=.001 and p<.001). No other significant differences were present (p> 0.05). Effect sizes for significant differences were calculated and can be seen in figure 4.2.1.

Table 4.2.1. Repeated measures reliability for front tibial acceleration (n=5x6 balls) during bowling and a custom-built surface impactor (n=4x12 impacts).

| | **Bowling Tibial Acceleration** | | | | | | | | | | | | | | |
| | **Peak Along-Tibial Acceleration** | | | **Peak Resultant Acceleration** | | | **Time-to-peak Acceleration** | | | **Normalised Peak Acceleration** | | | **Loading Rate** | | |
Surface	ICC	SEM (g)	MDC (g)	ICC	SEM (g)	MDC (g)	ICC	SEM (ms)	MDC (ms)	ICC	SEM (g.kg^{-1})	MDC (g.kg^{-1})	ICC	SEM (g.s^{-1})	MDC (g.s^{-1})
Astroturf	.932	1.17	3.24	.939	1.46	4.05	.919	1.99	5.52	.934	0.01	0.03	.879	230.00	637.52
Grass	.932	1.54	4.27	.939	1.58	4.38	.919	3.68	10.20	.934	0.02	0.06	.879	228.71	633.95
Rubber	.932	0.86	2.38	.939	1.14	3.16	.919	1.67	4.63	.934	0.01	0.03	.879	106.76	295.92
Wood	.932	1.04	2.88	.939	1.33	3.69	.919	2.60	7.21	.934	0.03	0.08	.879	144.85	401.50
Mean (SD)		**1.15**	**3.19**		**1.38**	**3.82**		**2.49**	**6.89**		**0.02**	**0.05**		**177.58**	**492.22**
	Peak Vertical Acceleration						**Impactor**								
Astroturf	.820	1.55	4.30				.716	0.46	1.28						
Grass	.820	1.12	3.10				.716	0.44	1.22						
Rubber	.820	1.19	3.30				.716	0.50	1.39						
Wood	.820	2.93	8.12				.716	0.54	1.50						
Mean (SD)		**1.70**	**4.71**					**0.49**	**1.35**						

g, gravity; SD, standard deviation; ms, milliseconds; ICC, intra-class correlation coefficient; SEM, standard error of measurement; MDC, minimal detectable change; kg, kilograms; s, seconds.

Table 4.2.2. Mean (±SD) acceleration data from the front tibia during bowling (n=5) and a custom-built surface impactor (n=12).

Surface	Peak Vertical Acceleration Impactor (g)	Time-to-peak Acceleration Impactor (ms)	Peak Along-Tibial Acceleration (g)	Peak Resultant Tibial Acceleration (g)	Peak Along-Tibial Acceleration (g.kg⁻¹)	Time-to-peak Acceleration (ms)	Peak Tibial Loading Rate (g.s⁻¹)
Astroturf	54.61 (2.52)	7.69 (0.76)	26.57 (5.92)	28.14 (5.79)	0.35 (0.07)	26.93 (13.23)	1432.84 (1136.23)
Grass	51.39 (1.88)	8.31 (0.74)	24.68 (6.64)	29.38 (6.41)	0.30 (0.07)	36.36 (16.11)	901.53 (815.49)
Rubber	57.60 (2.67)	7.22 (0.98)	21.97 (4.89)	24.14 (4.62)	0.27 (0.07)	30.55 (13.42)	867.78 (490.87)
Wood	51.38 (5.71)	8.13 (1.02)	17.84 (5.69)	21.21 (5.38)	0.22 (0.07)	30.00 (9.76)	732.76 (622.60)
Mean (SD)	**53.75 (3.20)**	**7.84 (0.88)**	**22.77 (5.79)**	**25.72 (5.55)**	**0.29 (0.07)**	**30.96 (13.13)**	**983.73 (842.35)**

g, gravity; SD, standard deviation; ms, milliseconds; kg, kilograms; s, seconds.
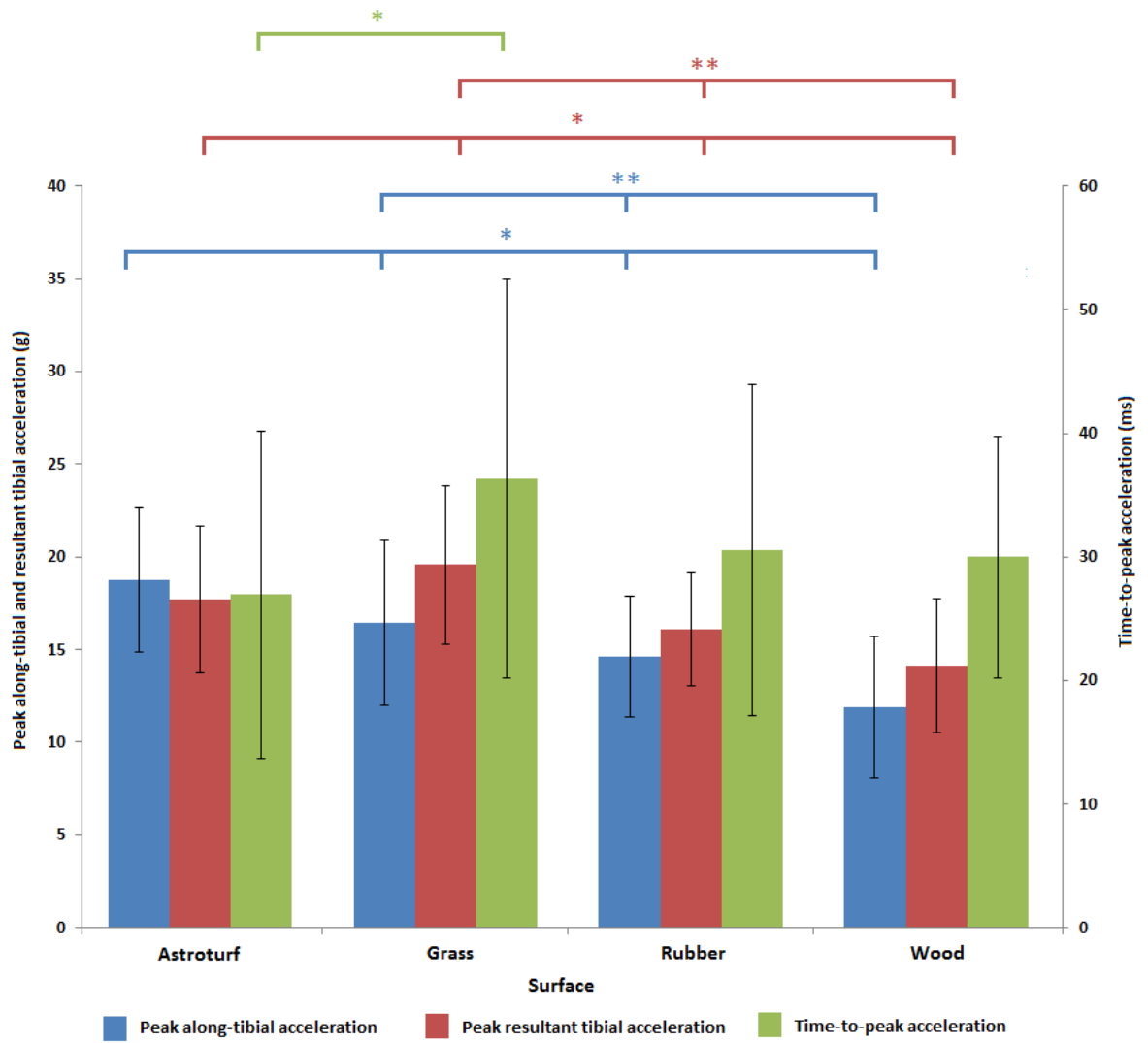
Figure 4.2.1.Mean peak along-tibial, resultant and time-to-peak tibial acceleration during bowling on four different playing surfaces (n=5,*p<0.05*).
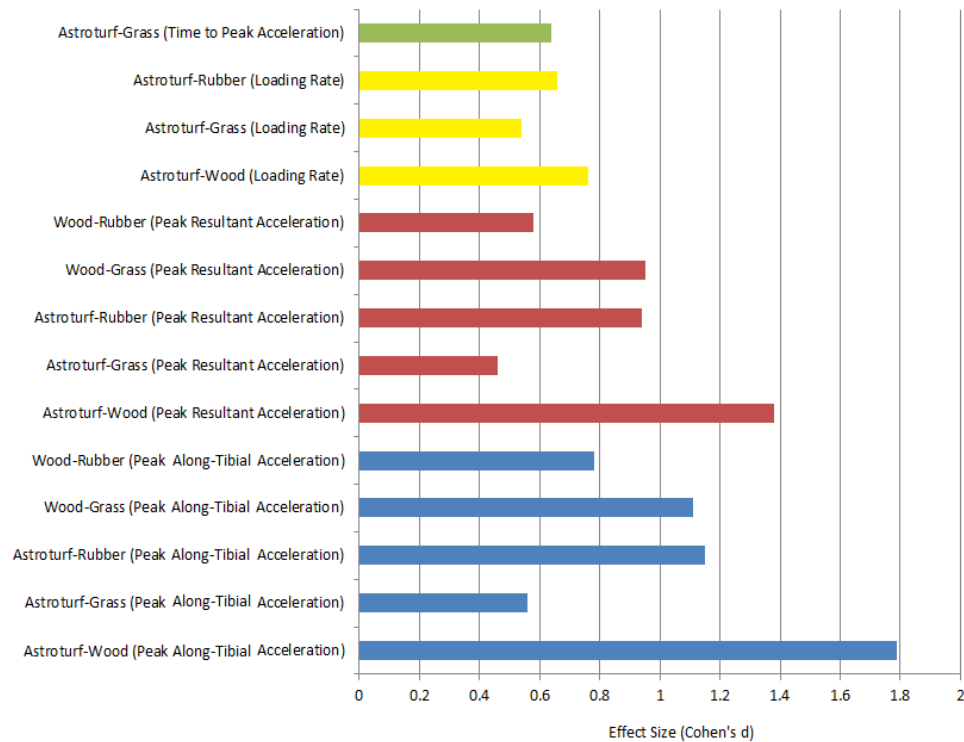
Figure 4.2.2. Effect sizes for significant differences in tibial acceleration characteristics between playing surfaces.

4.2.4 Discussion

4.2.4.1 Reliability of measurement

This study has used a novel method for both surface firmness classification and determination of front-foot impact characteristics during fast bowling. Consequently, reliability analysis was warranted. ICCs for both the impactor and tibial acceleration demonstrated good to excellent reliability. SEM values for all measurements were small and consequently, there was little variation in repeated tests for both impactor and tibial accelerations. Mean MDC values for peak and time-to-peak tibial acceleration shows that the accelerometer is able to detect small changes in tibial accelerations during fast bowling. These values are useful in analysing which changes can be attributed to meaningful differences in performance of a skill as opposed to natural variation of the individual. As all significant differences for tibial acceleration are greater than the MDC values produced, it can be concluded (with 95% confidence) that these differences are a result of factors other than natural variance, such as change in bowling surface properties. The ICCs reported in this study are similar to those reported for other surface

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impactor devices (0.77-0.87) and tibial accelerations during running (0.75-0.95) (Turcot et al. 2008; Twomey et al. 2011). The high reliability of these methods suggest that the testing apparatus used in this study offers a viable solution for surface analysis in other sports, such as running, where surface firmness may significantly affect performance or risk of injury (Dixon et al. 2000).

4.2.4.2 Phase One - Impactor results

Previous studies have investigated the difference in playing surfaces across different sports using a similar impactor device (Carre and Haake, 2004; Bartlett et al. 2009). No studies have investigated the surface firmness of different cricket playing surfaces therefore no previous impact results were available for direct comparison. Impactor results implied that the rubber surface was the firmest with the fastest time-to-peak suggesting that loading rate was significantly greater than the other surfaces. Therefore, these results suggest that high workloads should be avoided on the rubber surface. Conversely, grass and wooden surfaces, which demonstrated the lowest peak and slowest time-to-peak, may be more desirable for impact reduction. However, running studies have reported that individuals subconsciously adapt running kinematics to compensate for different firmness characteristics (Squadrone and Gallozzi, 2009). Therefore, impactor results in isolation may not simulate the stress being experienced by bowlers on different surfaces during front-foot impact, and thus can only be used to report differences in surface firmness. Analysis of tibial acceleration during bowling on different surfaces may be a more representative measure when making training recommendations.

4.2.4.3 Phase Two - Tibial Acceleration results

The results in this study suggest tibial acceleration does not corroborate those of the impactor in isolation confirming the importance of considering the human-surface interaction. Peak tibial acceleration was greatest on the astroturf surface as opposed to the rubber surface as determined by the impactor in isolation. Furthermore, no significant difference in peak acceleration was observed between the grass and rubber surfaces, in opposition to the impactor results. This suggests that variables other than surface firmness may affect peak tibial acceleration. It may be postulated that this difference is as a result of altered bowling kinematics. This hypothesis is in line with previous research measuring GRF in running which suggests participants adjust their

kinematics to compensate for harder or softer impact conditions (Squadrone and Gallozzi, 2009). However, as no previous fast bowling research has analysed the effect of surface on front-foot impact conditions and the resulting kinematics, this effect cannot yet be confirmed in fast bowling. Results of normalised peak tibial acceleration mimicked the above results, suggesting that these effects are also present regardless of bodyweight. However, it can be concluded that as astroturf produced the highest peak and fastest time-to-peak acceleration, it may be the least desirable surface to bowl high workloads on. In order to make conclusions based on both peak tibial acceleration and time-to-peak acceleration the analysis of average loading rate data has been recommended, as risk of bone failure has been reported to be affected by the rate at which force is applied (Courtney et al. 1994; Tran et al. 1995). Acceleration has been shown to correlate with GRF during landing tasks; thus, acceleration variables reported in this study may give insight into injury risk related to peak force and loading rate dependant injuries such as tibial and lumbar stress fractures (Courtney et al. 1994; Tran et al. 1995; Sell et al. 2014).

Average loading rate of the four surfaces did produce differing results to the peak tibial acceleration data. These results demonstrate that astroturf displayed a significantly greater average loading rate than all other surfaces, but wood was not significantly lower than grass or rubber as demonstrated in peak tibial accelerations. Although no significant difference in average loading rate was observed, a lower peak acceleration on the wooden surface may still be advantageous in reducing risk of injury. These results demonstrate that while the amalgamation of peak and time-to-peak acceleration into average loading rate may be a useful statistic when looking at pathomechanics; reporting average loading rate alongside peak acceleration may give a more comprehensive analysis of these characteristics.

The comparison between tibial acceleration on the rubber and grass surfaces is important in regards to future recommendations, as these are the two main surfaces most commonly utilised by professional cricket clubs. Impactor results report a significant difference between these two surfaces, but no significant difference in tibial accelerations were observed. This suggests individuals may be altering bowling kinematics to compensate and that impact to the human is equivocal across the two surfaces. The effect this may have with regards to injury from transfer of regular bowling on rubber surfaces (in the winter off season) to regular bowling on grass

(during the season) has yet to be established. Although this has not been reported in cricket, research on running has highlighted that a change in impact conditions (i.e. surface or shoes) may increase the risk of stress injuries (Van Mechelen, 1992). This has been attributed to the corresponding change in running kinematics (Hardin et al. 2004). As the effect of surface on bowling kinematics has not been addressed, no inferences can be made at this stage. Additionally, as stated above, footwear is likely to alter impact conditions. Studded footwear is likely to be worn on grass wickets due to decreased friction and may therefore change the lower limb's interaction. Whilst footwear was controlled for the purpose of comparisons in this study, an investigation looking at the effect of studded footwear may be warranted.

4.2.4.4 Implications of Results

This study has assessed the feasibility of the use of a custom-built impactor for surface firmness analysis, as well as the use of accelerometry for classification of front-foot tibial acceleration during bowling. The ability of this method to detect small changes in accelerations may be utilised to advise coaches and therapists in the implementation of technique interventions during 'live' bowling scenarios. This real-time feedback may aid in performance analysis or in rehabilitation settings where enforcing an incremental increase in impact may be advantageous as a return to play protocol. The highly portable and relatively inexpensive nature of this method, may prove an effective alternative to current force plate methodologies across a variety of sports.

This study has highlighted that there were no significant differences in the tibial impact characteristics during fast bowling on the two most commonly used surfaces in professional cricket (grass and rubber) even though the surfaces themselves display different firmness. This may indicate altered bowling patterns between the two surfaces, however the implications this may have on injury or performance are unknown. All tests showed the astroturf surface to be the firmest, therefore excessive bowling workloads on this surface may increase risk of injury. Conversely, wooden surfaces may be beneficial in instances such as return to play and high bowling workloads, where impact reduction may be desired.

4.2.5 Conclusion

This study has attempted to quantify the effect of different cricket playing surfaces on fast bowlers and has highlighted significant differences in peak, time-to-peak and

average loading rate between surfaces. However, it is still unclear how these surfaces affect bowling kinematics. Therefore, future studies should aim to investigate the effect of different bowling surfaces on fast bowling kinematics. The effect of high bowling workloads on injury has been well documented, but the effect of which surface this bowling is carried out on is yet to be reported, and may go some way to further the understanding of the relationship between impact characteristics and injury in the fast bowling population.

4.3 Shoulder counter-rotation and hip-shoulder separation angle in cricket fast bowling: What are they really measuring?

4.3.1 Introduction

Fast bowlers in cricket have been identified as having a significantly higher risk of musculoskeletal injury compared with the rest of the team (Johnson et al. 2012; Orchard et al. 2006). Spinal injury in the fast bowling population contribute to more than twice the number of games missed compared with any other injury (Orchard et al. 2002). Studies synthesising the literature have shown the prevalence of spondylolysis to be 27% for fast bowlers; significantly higher than the general and athletic populations at 6% and 12% respectively (Rossi and Dragoni, 1990; Kalichman et al. 2009). Missed playing time has been reported at 247 games over six seasons as a result of injuries to the lumbar spine (Orchard et al. 2002). Consequently, researchers have focused on attempting to identify the spinal kinematics of fast bowling and their link with spinal pathology (Johnson et al. 2012).

Previous systematic reviews have concluded that shoulder counter-rotation (SCR) in excess of 30 degrees during bowling is associated with a higher risk of developing lower back pathology, such as spondylolysis (Elliott et al. 1993; Portus et al. 2004; Morton et al. 2013). Consequently, many previous studies have focussed on reporting SCR (Ranson et al. 2008; Ranson et al. 2009; Stuelcken et al. 2010; Crewe et al. 2013). SCR is determined by subtracting the minimum shoulder alignment angle relative to the stumps, from shoulder alignment at back-foot impact (BFI), as seen in figure 4.3.1 (Ranson et al. 2008). Reported SCR values range between 10-45 degrees, with higher SCR values typically seen in bowlers with a mixed bowling action (Foster et al. 1989; Elliott et al. 1992; Johnson et al. 2012). However, whilst these values may be a useful metric for coaches to quickly analyse technique, it only considers shoulder alignment. SCR does not include the pelvis as a frame of reference and therefore can be created by spinal rotation or whole body rotation or a combination of both. Therefore, the actual spinal kinematics which determine SCR are unclear and thus the mechanisms of how SCR contributes to an increased likelihood of lower back pathology remain unclear.
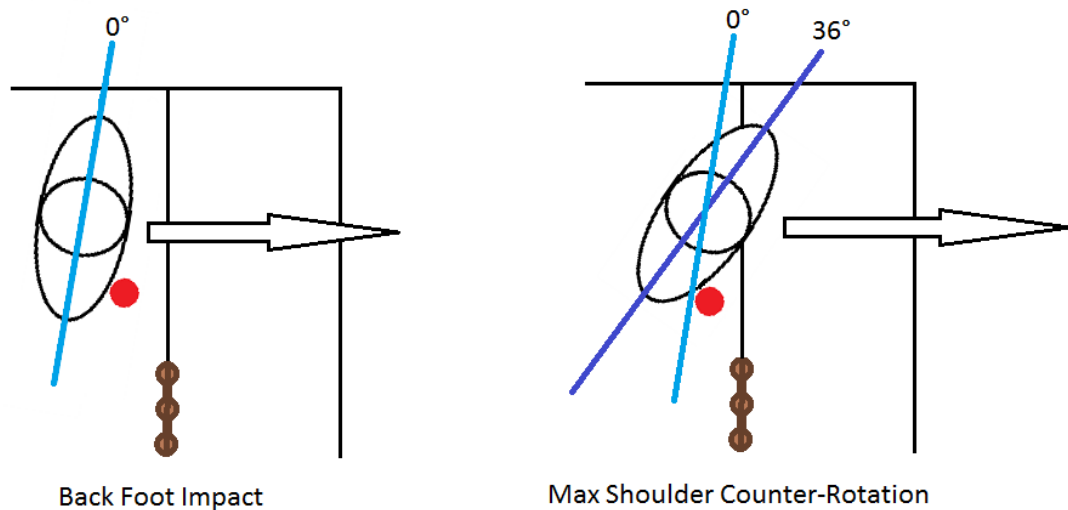
Figure 4.3.1. Calculation of shoulder counter-rotation using shoulder orientation.

In addition to SCR, hip-shoulder separation (HSS) angles have also been used by researchers and coaches to describe bowling kinematics (Burnett et al. 1995; Portus et al. 2004). HSS angle is typically analysed at back-foot impact and is calculated by subtracting hip orientation from shoulder orientation in the transverse plane (Burnett et al. 1995). As these values only analyse motion in a single plane they fail to describe three-dimensional spinal kinematics throughout the delivery stride. Therefore, it remains unclear as to how SCR and HSS relate to more traditional descriptions of three-dimensional kinematics of the lumbar and thoracic spine.

4.3.2 Aim of the study

The aim of this study was to investigate the relationship between shoulder counter-rotation, hip-shoulder separation and three-dimensional spinal kinematics during cricket fast bowling.

4.3.3 Results

Mean (SD) shoulder counter rotation and hip-shoulder separation values were 27.4° ($\pm$16.3°) and 33.0° ($\pm$21.6°) respectively.

In regards to SCR a total of eight correlations were deemed significant (table 4.3.1) however following Bonferroni correction for multiple significance testing two remained significant. These significant negative correlations were observed for thoracolumbar lateral flexion and thoracic lateral flexion between back-foot impact and maximum

contralateral T1 rotation. This suggests if a bowler displays a greater range of thoracic and thoracolumbar lateral flexion away from the direction of delivery at the beginning of the delivery stride SCR values will be larger.

In regards to HSS angle a total of four correlations were significant, with one remaining significant following Bonferroni correction (table 4.3.1). This negative correlation was evident between HSS and thoracolumbar lateral flexion between back-foot impact and maximum contralateral T1 rotation. This suggests that if a bowler displays greater thoracolumbar lateral flexion away from the direction of delivery at the beginning of the delivery stride HSS angle will be larger.

Table 4.3.1. The correlation between three-dimensional spinal kinematics, shoulder counter-rotation and hip-shoulder separation values (n=35).

| | Shoulder Counter Rotation | | | | | | Hip-Shoulder Separation | | | |
| | BFI-MCR | | | BFI-FFI | | | BFI-MCR | | BFI-FFI | |
	Mean (±SD)	r_s	P	Mean (±SD)	r_s	p	r_s	p	r_s	p
Lumbar Flexion	3.03 (15.85)	-.061	.538	35.85 (30.09)	-.141	.153	.024	.811	.031	.753
Lumbar Lateral Flexion	-20.18 (15.84)	-.254	.009	31.11 (26.06)	.095	.337	-.268	.006	.075	.450
Lumbar Rotation	-10.12 (12.88)	-.080	.421	18.12 (18.42)	-.050	.615	.077	.435	.084	.398
Thoracic Flexion	-17.49 (14.15)	-.118	.235	58.55 (32.18)	.225	.022	-.005	.956	.067	.502
Thoracic Lateral Flexion	-22.17 (18.37)	-.462	.000*	41.43 (30.68)	.106	.285	-.255	.009	-.075	.450
Thoracic Rotation	-11.48 (9.93)	-.227	.021	29.39 (25.74)	.196	.046	-.167	.090	.075	.447
Thoracolumbar Flexion	-13.62 (12.75)	-.121	.221	80.71 (43.62)	.159	.108	-.063	.524	.020	.844
Thoracolumbar Lateral Flexion	-28.96 (21.47)	-.460	.000*	52.46 (30.51)	.225	.022	-.552	.000*	.041	.676
Thoracolumbar Rotation	-13.15 (11.12)	-.260	.008	39.20 (41.12)	.170	.084	-.253	.010	.008	.937

BFI, back-foot impact; MCR, max contralateral T1 rotation; FFI, front-foot impact. * Denotes a significant correlation following Bonferroni correction (p<.003)

4.3.4 Discussion

Previous research has highlighted the importance of SCR during fast bowling as this has been linked to the presence of demonstrable spinal pathology (Elliott et al. 1993; Portus et al. 2004). Despite this previous research has highlighted SCR and spinal kinematics as separate variables in the analysis of either performance or injury surveillance (Ranson et al. 2008; Ranson et al. 2009; Stuelcken et al. 2010; Crewe et al. 2013). However, using SCR (attained from the shoulder alignment) to explain injuries obtained in the lumbar spine is challenging as it is not clear which three-dimensional kinematics are actually being measured. Although analysis of three-dimensional spinal kinematics may provide a more accurate understanding of injury mechanisms, it is difficult for coaches to monitor on a regular basis. Thus, understanding the relationship between three-dimensional kinematics and easy to measure metrics such as SCR and HSS is crucial for coaches when providing recommendations to players. Only one previous study has attempted to explore the association of these values with spinal kinematics and only lumbar kinematics were explored in relation to SCR (Crewe et al. 2011). Therefore, this study provides new insight into this relationship as well as novel findings relating to HSS and thoracic and thoracolumbar kinematics.

4.3.4.1 Back-foot impact to maximum T1 rotation

It has been suggested that increased SCR may be used as a conscious mechanism to generate pace on the ball when bowling (Portus et al. 2004). However, this may come at an injury cost. The results of this study illustrate that thoracic and thoracolumbar lateral flexion are the key spinal kinematic variables associated with SCR. The direction of the association is such that in those bowlers with high values of lateral flexion away from the direction of delivery (right lateral flexion for the right-handed bowler) are likely to display higher SCR values. It seems likely that these bowlers adopt a bowling strategy employing a wind-up phase utilising spinal lateral flexion and perhaps this element is the key preparation for driving SCR and ultimately generating pace on the ball. It has been hypothesised that SCR is predominately a surrogate measure of spinal rotation (Crewe et al. 2013; Glazier, 2010) however our results show no significant correlation with spinal rotation. The may be due to the strict Bonferroni correction applied as indeed without this a significant correlation was identified. This suggests that SCR may

be a complex interaction between lateral flexion and rotation as previously suspected (Crewe et al. 2013; Glazier, 2010).

HSS has been proposed as an additional method for coaches and researchers to describe the kinematics of fast bowling (Burnett et al. 1995; Portus et al. 2004). This is the first time the relationship between HSS and spinal three-dimensional kinematics has been explored. The results of this study demonstrate a significant relationship between thoracolumbar lateral flexion and HSS. The direction of the association suggests that bowlers who display greater values of lateral flexion ultimately display larger amounts of HSS. Therefore, it appears that these results mirror those for SCR.

It is important to acknowledge that despite the correlations being significant the actual magnitude of the correlations were moderate; suggesting a significant amount of SCR and HSS angles were not explained by three-dimensional spinal kinematics as described in this study. It is therefore likely that some of these measures of SCR and HSS are produced from whole body rotation and whole body lateral flexion. Such movements would not contribute to a change in the resultant angle between two sensors and therefore not recorded as resultant side flexion or rotation. The importance of which is that no resultant movement or minimal resultant movement is unlikely to pose the same injury risk. Therefore, it is imperative to differentiate between changes in whole spinal orientation and changes in resultant angle.

4.3.4.2 Back-foot impact to front-foot impact

No significant relationships were observed between SCR and HSS and spinal range of motion between BFI and FFI. This may be due to the fact that SCR describes shoulder orientation between BFI and MCR, and therefore conceptually may not be related to spinal kinematics of the whole delivery stride when movement towards the direction of delivery after MCR are considered. Furthermore, HSS does not take into account orientation relative to the wickets and is a static measure at one point in time (BFI) and therefore cannot differentiate between bowling actions (a completely front-on or side-on action would both produce HSS of 0°).

4.3.5 Conclusion

Results of this study have highlighted that SCR and HSS, as a means of describing spinal kinematics between BFI and FFI, differ from more traditional descriptions of

three-dimensional spinal kinematics. Therefore, bowling actions described according to SCR and HSS may not relate significantly to the underlying three-dimensional kinematics utilised. The significant relationship between SCR and thoracic and thoracolumbar lateral flexion may suggest techniques promoting increased lateral flexion during SCR may increase risk of injury. Further research should focus on the development of a more representative method of fast bowling technique analysis as well as if the proposed decrease in initial lateral flexion away from the direction of the delivery has any implications on ball release speed and injury risk.

4.4 Summary of Results and Key Findings

This chapter aimed to address some novel questions highlighted by the literature previously presented in this thesis. Whilst the main aim of this thesis is centred on the relationship between risk of lower back pain, impacts and spinal kinematics during 'in-field' fast bowling; the studies in this chapter offer methodological and theoretical support to this investigation.

Studies investigating fast bowling biomechanics share similar threats to validity. Most studies in fast bowling have used force plates and optoelectronic motion analysis systems, which are typically limited to a laboratory environment, thus a more portable system may be desirable for the analysis of 'in-field' fast bowling (Burnett et al. 1998; Ranson et al. 2008; Ferdinands et al. 2009; Ranson et al. 2009; Stuelcken et al. 2010; Crewe et al. 2013; Bayne et al. 2016). The results in this chapter demonstrate that inertial sensors are a valid and reliable method for the measurement of impact characteristics during fast bowling and three-dimensional spinal kinematics. Strong to very strong correlations were displayed in both impacts and spinal kinematics with gold standard comparators, with 79% of variables displaying $r > 0.8$. Furthermore, RMSEP displayed low values ranging from 0.3-1.5°. However, it must be acknowledged that no comparisons were able to be carried out for thoracic kinematics due to the optoelectronic model used.

Intra-class correlation coefficients (ICC) demonstrated high levels of agreement in repeated trials with all but one tibial measure (time-to-peak resultant acceleration at front-foot impact) displaying ICC's > 0.8. SEMs< 2.0g at back-foot impact and < 3.4g at front-foot impact for peak tibial acceleration along all axes highlights fast bowling as a repeatable skill and demonstrates small measurement error in the inertial sensors used. MDCs < 5.0g for back-foot tibial accelerations and <8.0g at front-foot impact in all three axes demonstrate inertial sensors can identify 'real change' (not attributed to natural variability or measurement error) to a high degree of sensitivity. ICCs, SEMs and MDCs for sacral impact characteristics displayed greater inter-trial variability than tibial impacts, however still demonstrated good reliability with low SEM and MDC values.

Intra-class correlation coefficients (ICC) for spinal kinematics demonstrated high levels of agreement for repeated trials, with all measures displaying good (>0.5) to excellent

(>0.8) ICCs. SEM typically ranged from 2.7-8.6°, with only thoracolumbar lateral flexion outside these values at 20.7°. With SEMs in previous fast bowling studies reported between 1-17° using optoelectronic motion analysis systems, the values reported in this study demonstrate good reliability. MDC values highlight any change in lumbar kinematics greater that 13 degrees can be attributed to 'real change'. Coefficients of multiple correlations (CMC) demonstrated good agreement (>0.5) for repeated trials across the whole delivery stride for all measures; highlighting that inertial sensors are able to detect consistent movement patterns demonstrated during the bowling action. Consequently, as inertial sensors offer a more portable and cost effective alternative to current force plate and optoelectronic methodologies and display good reliability for measurement of fast bowling impacts and spinal kinematics, they may be a desirable alternative for practitioners and coaches.

However, limitations to this technology must be considered in order to accurately interpret data provided. Whilst, they provide a portable measure of impacts and kinematics, the fact that they are larger and heavier than retroflective markers used in optoelectronic systems increases the likelihood of additional skin movement artefact. Furthermore, attachment to the skin on curved surfaces was problematic, especially fixation to the skin over the L1 spinous process in the paraspinal muscle gap, meaning that bridging this gap was necessary (see Appendix 4). In addition, some data processing issues were evident. Whilst, acceleration has been shown to closely correlate with ground reaction force, the absence of the mass element means that force cannot be reported and as such, comparisons with previous literature are difficult. However, previous literature using accelerometry has highlighted a strong relationship between acceleration and force, therefore peak acceleration may be used as an acceptable proxy for force for impact measurement (Tran et al. 2010). Lastly, due to the lack of line of sight needed, marking specific events that are not demarked by specific trends in the data are not able to be identified (for example, ball release).

The above limitations are acceptable in order to allow a reliable and more portable method of fast bowling analysis which is able to address limitations to current laboratory-based technologies. Consequently, this thesis was able to analyse the effect of different playing surfaces on fast bowling impacts; a protocol which is extremely difficult to carry out using force plate technology. The use of inertial sensors for the measurement of tibial impacts while bowling on different playing surfaces highlighted

that whilst playing surface did affect tibial impact characteristics, bowling on a 'harder' surface did not always result in larger or faster impacts. Outdoor astroturf was reported as the hardest surface and produced the highest tibial impacts. Consequently, high workloads should be avoided on this surface. Conversely, the indoor wooden sports hall floor produced the lowest impacts and thus may be advantageous in 'return to play' or high workload scenarios. Although, surface testing reported the indoor rubber surface as firmer than a grass wicket, no significant differences in tibial impacts were observed during bowling. This may suggest a regulatory method, such as altered kinematics, to cope with changes in impact conditions.

Inertial sensors were then used to quantify the relationship between three-dimensional spinal kinematics and shoulder counter-rotation (SCR) and hip-shoulder separation (HSS), two-dimensional measures that have been linked to risk of lower back injury. This study highlighted that whilst analysis of SCR and HSS are simple measures for coaches and clinicians, they showed few significant relationships with three-dimensional spinal kinematics during bowling, and thus must be used with caution when describing spinal kinematics. Significant relationships were found between SCR and thoracic and thoracolumbar lateral flexion during the early phase of the bowling action, however much of the variation in SCR and HSS values is unaccounted for by three-dimensional spinal kinematics.

This chapter has highlighted that inertial sensors can reliably measure fast bowling tibial and sacral impacts as well as three-dimensional spinal kinematics. They are also able to overcome some methodological limitations highlighted in previous studies. Consequently, they may be deemed appropriate to measure fast bowling impacts and kinematics during 'in-field' fast bowling with the aim at highlighting relationships between fast bowling biomechanics and LBP in junior and senior bowlers.

Summary of key findings

- Inertial sensors are a valid and reliable method of analysing fast bowling impacts at the tibia and sacrum and three-dimensional spinal kinematics.
- Outdoor artificial wickets produced the highest tibial accelerations; high workloads on this surface should be avoided.
- Wooden indoor surface produced the lowest tibial accelerations, which may be advantageous for return to play or high workload scenarios.

- Grass and indoor rubber wicket produced no differences in tibial accelerations.

- Shoulder counter-rotation and hip-shoulder separation may not be approriate for describing three-dimensional spinal kinematics during the delivery stride.

Chapter 5

Fast Bowling and Lower Back Pain in Junior and Senior Fast Bowlers

5.1 An investigation into the tibial and sacral accelerations and three-dimensional spinal kinematics of junior and senior fast bowlers. A pilot study.

5.1.1 Introduction

The previous chapter has produced new insights into fast bowling characteristics. However, more is needed in order to fully address the injury problem in the fast bowling population. High impact forces and excessive spinal range of motion during the delivery stride in cricket fast bowling have been hypothesised to increase risk of lower back injury, with junior bowlers at even greater risk (Elliott, 2000). Whilst variables such as immature structures in the spine and workload have been hypothesised to cause this increased risk, no direct comparison of fast bowling impacts of junior and senior bowlers has been reported in current literature (Johnson et al. 2012; Arora et al. 2014). Thus, it remains unclear as to whether differences in fast bowling impacts and spinal kinematics between junior and senior bowlers could contribute to junior bowlers' increased risk of injury.

The most commonly reported values are that of peak vertical and braking ground reaction force (GRF), with some studies also reporting time-to-peak GRF (Elliott et al. 1993; Hurrion et al. 2000; Stuelcken et al. 2009; Crewe et al. 2013; Worthington et al. 2013). Peak GRF at front-foot impact (FFI) has been reported at around six times body weight vertically and three times body weight anterior-posteriorly (Elliott et al. 1993; Hurrion et al. 2000; Stuelcken et al. 2009; Crewe et al. 2013; Worthington et al. 2013). These studies report time-to-peak GRF at FFI at around 59ms vertically and 55ms anterior-posteriorly. Fewer studies have reported GRF at back-foot impact (BFI) (Mason et al. 1989; Saunders and Coleman, 1991; Elliott et al. 1992; Hurrion et al. 2000). Peak vertical GRF at back-foot impact has been reported approximately 2-3 times body weight vertically and one times body weight anterior-posteriorly (Mason et al. 1989; Saunders and Coleman, 1991; Elliott et al. 1992; Hurrion et al. 2000).

Research has typically focused on kinetic analysis of either junior or senior fast bowlers, with little direct comparison between the two (Hurrion et al. 2000; Portus et al. 2004; Stuelcken and Sinclair, 2009; Crewe et al. 2013; Worthington et al. 2013; Spratford and Hicks, 2014; Bayne et al. 2016; King et al. 2016; Middleton et al. 2016). This comparison may enable further understanding of the pathomechanics of lower back injury and pain and why junior fast bowlers display increased injury risk. However,

heterogeneity in study methodologies has made between study comparisons difficult and therefore whether any differences in impact characteristics between junior and senior fast bowlers exist remains unknown.

Fast bowling kinematics have been well documented in the literature, with emphasis being placed on lower limb and spinal kinematics during the delivery stride and their association with performance and injury (Johnson et al. 2012; Morton et al. 2013). Whilst most studies report samples of either junior or senior bowlers, few studies have offered a direct comparison of the two (Portus et al. 2004; Elliott et al. 2005; Crewe et al. 2013).

SCR greater than 30 degrees has been linked to increased risk and has been suggested to be a strategy to increase bowling pace (Elliott et al. 1993; Portus et al. 2004; Elliott et al. 2005; Morton et al. 2013). Therefore, juniors looking to develop extra pace on their delivery may employ excessive SCR explaining the elevated risk. It may be possible that junior bowlers utilise a different kinematic strategy to bowling to offset their smaller anthropometrics, which may or may not further affect their risk of injury. Due to the lack of direct comparisons between junior and senior bowlers these theories remain speculative.

Current studies have reported three-dimensional kinematics of the lumbar and thoracolumbar spine at back-foot impact (BFI), front-foot impact (FFI) and ball release (Burnett et al. 1998; Ranson et al. 2008; Ferdinands et al. 2009; Ranson et al. 2009; Stuelcken et al. 2010; Crewe et al. 2013). However, studies have defined spinal segments differently, as well as reporting range of motion between different points of the delivery stride. Consequently, as junior and senior bowlers tend only to be reported in separate studies with varying methodologies, a direct comparison between the two would not be appropriate. Thus, a study that is able to compare three-dimensional spinal kinematics of junior and senior bowlers may provide valuable insight into whether junior fast bowlers exhibit different bowling kinematics to senior bowlers and whether they are likely to further increase their risk of lower back injury and pain.

5.1.2 Aim of the study

This study aims to compare impact characteristics of the tibia and sacrum and three-dimensional spinal kinematics between the front and back-foot impact phase of fast bowling in junior and senior fast bowlers.

5.1.3 Results

Impact characteristics for both groups can be found in table 5.1.1.

At BFI peak tibial acceleration along the y axis were significantly greater in seniors compared with juniors. Furthermore, time-to-peak resultant tibial acceleration was significantly longer in seniors compared with juniors at BFI.

No other significant differences were observed, however there were some large effect sizes. At BFI four variables demonstrated effect sizes >1. In addition to those mentioned above peak mediolateral sacral acceleration was greater in seniors but normalised peak vertical acceleration was greater in juniors. At FFI time-to-peak tibial acceleration in the along tibial axis was greater in seniors with an effect size of >1.2.

Moderate effect sizes (>0.5) were observed for resultant tibial acceleration, time-to-peak tibial acceleration along the x axis and resultant acceleration attenuation at BFI.

Only time-to-peak tibial acceleration along the x axis displayed a high effect size at FFI, with faster time-to-peak seen in junior bowlers. Moderate effect sizes were observed for peak tibial acceleration along the z axis, normalised peak tibial acceleration along the x axis and normalised peak vertical sacral acceleration at front-foot impact. Peak tibial acceleration along the z axis was greater in senior bowlers; however normalised accelerations for tibial acceleration along the x axis and vertical sacral acceleration were greater in juniors.

Table 5.1.1. Mean (±SD) accelerations at the tibia and sacrum during the back and front-foot impact phase of the delivery stride in junior (n=21) and senior (n=14) fast bowlers.

	Junior		Senior		p-value		Effect size (d)	
	Back-Foot Impact	**Front-Foot Impact**	**Back-Foot Impact**	**Front-Foot Impact**	BFI	FFI	BFI	FFI
Tibial Acceleration								
Peak Tibial Acc x (g)	11.3 (4.6)	26.2 (9.0)	14.11 (6.6)	25.5 (3.9)	.171	.866	-0.49	0.09
Peak Tibial Acc y (g)	2.7 (1.8)	13.3 (7.8)	6.71 (4.5)	11.1 (9.0)	.000*	.352	-1.18	0.27
Peak Tibial Acc z (g)	15.8 (9.2)	17.3 (7.3)	15.9 (8.5)	24.8 (15.9)	.955	.066	-0.01	-0.65
Resultant Tibial Acc (g)	18.1 (7.3)	33.1 (12.0)	23.1 (7.9)	38.2 (19.3)	.063	.455	-0.66	-0.33
Time-to-peak Tibial Acc x (ms)	22.8 (12.7)	16.6 (3.3)	29.5 (6.8)	27.5 (13.7)	.003	.004	-0.66	-1.21
Time-to-peak Resultant Tibial Acc (ms)	46.5 (5.2)	57.5 (13.0)	66.7 (9.9)	59.5 (14.6)	.000*	.670	-2.55	-0.15
Mean Tibial Loading Rate x (g.s^{-1})	664.1 (330.5)	1707.5 (648.6)	551.7 (272.4)	1432.7 (1097.2)	.299	.358	0.36	0.13
Mean Resultant Tibial Loading Rate (g.s^{-1})	475.1 (243.3)	741.0 (588.1)	384.0 (139.3)	775.0 (441.3)	.271	.743	0.44	-0.06
Normalised Peak Tibial Acc x (g.kg^{-1})	0.2 (0.1)	0.4 (0.1)	0.2 (0.1)	0.3 (0.2)	.654	.108	0.00	0.55
Normalised Resultant Tibial Acc (g.kg^{-1})	0.3 (0.1)	0.5 (0.1)	0.3 (0.1)	0.4 (0.2)	.782	.571	-0.10	0.19
Sacral Acceleration								
Peak Vertical Sacral Acc (g)	2.6 (0.8)	3.2 (0.7)	2.9 (0.7)	3.3 (0.4)	.411	.787	-0.28	-0.09
Peak Mediolateral Sacral Acc (g)	0.6 (0.3)	0.9 (0.6)	1.3 (0.9)	1.1 (0.6)	.003	.464	-1.13	-0.23
Peak Anterior-Posterior Acc (g)	1.4 (0.7)	1.5 (0.9)	1.6 (0.9)	1.5 (0.6)	.441	.944	-0.28	0.03
Resultant Sacral Acc (g)	3.1 (0.8)	3.9 (0.8)	3.4 (0.9)	3.9 (0.6)	.276	.973	-0.39	-0.01
Time-to-peak Vertical Sacral Acc (ms)	71.8 (4.0)	62.7 (13.0)	69.1 (19.6)	68.0 (13.7)	.671	.260	0.22	-0.39
Time-to-peak Resultant Sacral Acc (ms)	72.6 (6.2)	64.1 (9.6)	70.6 (20.3)	67.1 (15.7)	.754	.561	0.13	-0.23
Mean Sacral Vertical Loading Rate (g.s^{-1})	43.2 (14.4)	63.0 (18.5)	48.3 (22.0)	56.8 (15.0)	.504	.304	-0.29	0.36
Normalised Peak Vertical Sacral Acc (g.kg^{-1})	0.04 (0.01)	0.05 (0.02)	0.03 (0.01)	0.04 (0.00)	.120	.044	1.00	0.60
Resultant Attenuation (%)	80.1 (8.3)	86.7 (1.1)	83.9 (5.5)	85.7 (7.9)	.157	.684	-0.51	0.18

Acc, acceleration; g, gravity; ms, milliseconds; s, seconds; kg, kilograms; *, p<.001.

All spinal kinematics for junior and senior fast bowlers can be found in table 5.1.2.

5.1.3.1 ROM

Junior bowlers displayed significantly larger thoracolumbar rotation ROM during static range of motion trials as seen in figure 5.1.1 (p = .005). There were no other significant differences in ROM between the two groups.

5.1.3.2 Orientation at BFI

There were no significant differences in spinal orientation at BFI between junior and senior fast bowlers, however a large effect size was observed. Junior fast bowlers displayed more contralateral rotation of the thoracolumbar spine at BFI compared with senior bowlers (d=-1.02). In addition, a medium effect size was observed for hip-shoulder separation with greater values seen in senior bowlers. Both junior and senior bowlers bowled with 'front-on' bowling actions as highlighted by T1 and S1 orientation at BFI being larger than 240° (MCC, 1976; Elliott and Foster, 1984). However, shoulder counter-rotation and hip-shoulder separation angles were both larger than 30° in both groups which, according to the above guidelines, should only be the case in a 'mixed action'.

5.1.3.3 Orientation at FFI

No significant differences or notable effect sizes were observed for spinal orientation at FFI.

5.1.3.4 Spinal ROM during delivery stride

No significant differences or notable effect sizes were observed for spinal range of motion during the delivery stride (figure 5.1.2). However, when expressed as percentages of static ROM a large effect size was observed in lumbar right lateral flexion which was larger in junior fast bowlers (d=0.94). Moderate effect sizes also highlighted greater lumbar extension in junior bowlers (d=0.55). Senior fast bowlers displayed greater thoracic left lateral flexion and thoracolumbar right rotation (d= 0.75 and 0.69 respectively).
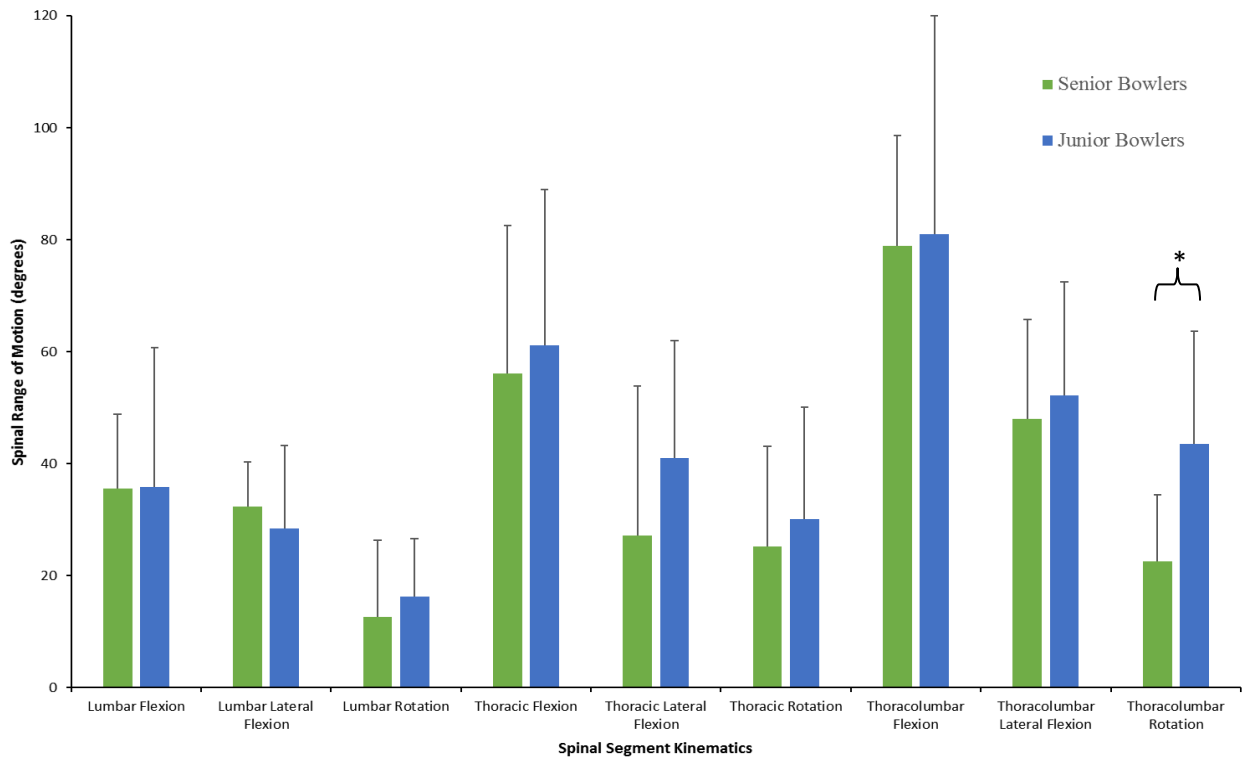
Figure 5.1.1. Spinal range of motion of junior (n=21) and senior (n=14) fast bowlers during static range of motion trials (* p< .006).
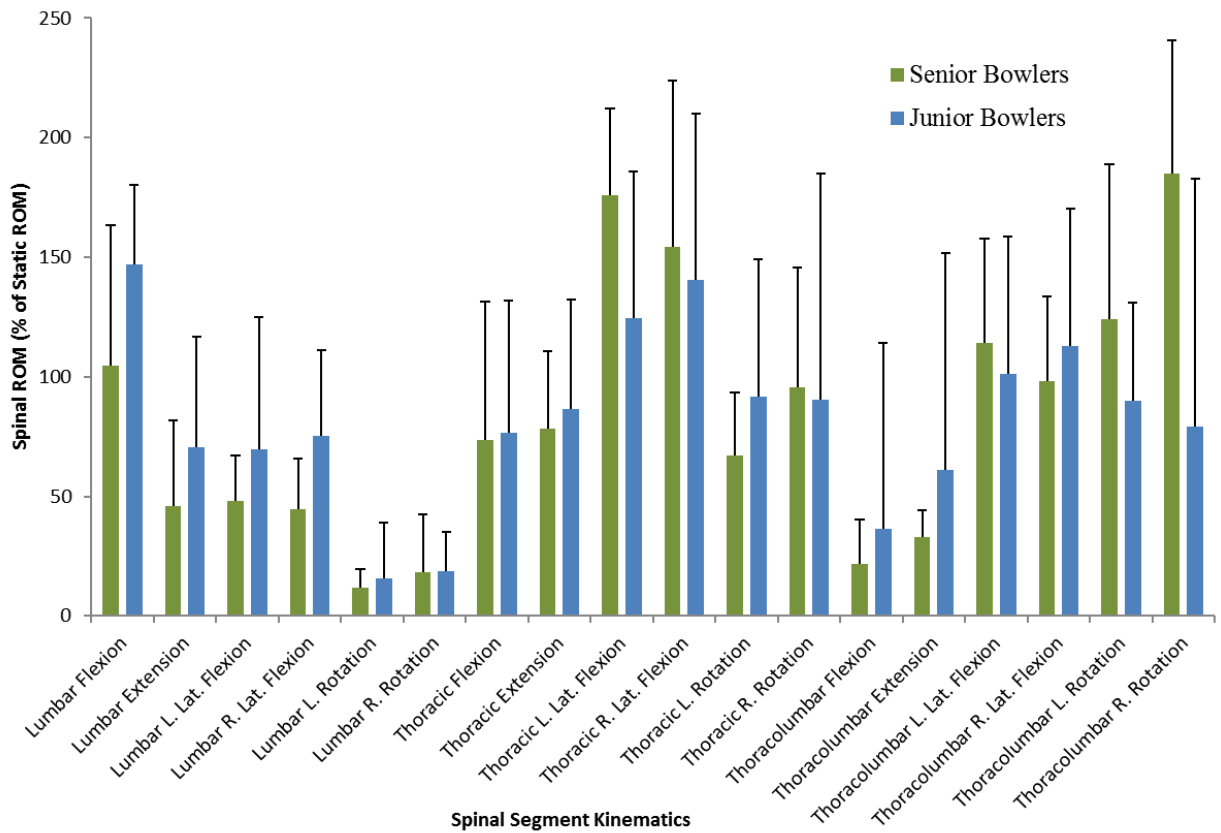
Figure 5.1.2. Junior (n=21) and senior (n=14) fast bowlers' maximum spinal range of motion (ROM) during bowling expressed as a percentage of static ROM. L, left; R, right.

Table 5.1.2. Mean (SD) spinal kinematics during cricket fast bowling in junior (n=21) and senior (n=14) fast bowlers.

Spinal Kinematics (°)	BFI				FFI				Range			
	Junior	Senior	P value	Effect Size (d)	Junior	Senior	P value	Effect Size (d)	Junior	Senior	P value	Effect Size (d)
SCR	37.8 (10.3)	35.6 (8.7)	.513	0.23								
HSS	30.5 (14.9)	41.5 (20.6)	.099	-0.63								
T1 orientation	270.9 (37.1)	268.6 (35.5)	.852	0.06								
S1 orientation	269.8 (38.9)	286.3 (25.4)	.117	-0.48								
Lumbar flexion	-13.7 (17.5)	-15.0 (9.0)	.778	0.09	20.5 (9.4)	22.4 (6.1)	.456	-0.23	34.2 (25.8)	36.5 (12.7)	.720	-0.11
Lumbar lateral flexion	-11.0 (10.6)	-12.7 (9.5)	.325	0.17	20.0 (9.4)	19.9 (6.2)	.972	0.01	31.0 (16.0)	31.0 (11.0)	.998	0.00
Lumbar rotation	-4.2 (7.6)	-0.1 (9.5)	.538	-0.49	14.1 (8.2)	14.6 (6.0)	.696	-0.07	18.3 (12.7)	14.7 (10.7)	.358	0.30
Thoracic flexion	-30.4 (15.4)	-24.5 (22.2)	.398	-0.32	28.4 (16.0)	35.2 (13.1)	.179	-0.46	58.8 (28.0)	56.6 (30.7)	.825	0.08
Thoracic lateral flexion	-18.1 (16.9)	-11.1 (14.0)	.191	-0.44	24.9 (10.5)	29.5 (13.5)	.293	-0.39	43.0 (20.3)	39.1 (24.7)	.628	0.18
Thoracic rotation	-11.1 (10.6)	-8.1 (11.8)	.455	-0.27	19.0 (14.5)	19.8 (8.3)	.546	-0.06	30.1 (18.5)	27.9 (14.8)	.738	0.13
Thoracolumbar flexion	-39.4 (21.8)	-34.2 (11.9)	.361	-0.28	40.1 (18.9)	46.4 (11.7)	.231	-0.38	79.6 (38.6)	78.6 (21.8)	.579	0.03
Thoracolumbar lateral flexion	-24.1 (10.6)	-19.9 (19.2)	.318	-0.29	29.1 (13.5)	33.2 (9.4)	.290	-0.34	53.2 (20.3)	53.1 (21.1)	.990	0.00
Thoracolumbar rotation	-15.4 (10.7)	0.0 (20.0)	.017	-1.02	25.7 (13.6)	28.5 (10.4)	.495	-0.23	41.0 (19.6)	28.5 (17.9)	.061	0.66

BFI, back-foot impact; FFI, front-foot impact; °, degrees; SCR, shoulder counter-rotation; HSS, hip-shoulder separation.

5.1.4 Discussion

This study aimed to investigate the differences between tibial and sacral accelerations and three-dimensional spinal kinematics in junior and senior fast bowlers in cricket. Junior fast bowlers have been identified as displaying increased risk of lower back injury when compared to senior fast bowlers (Johnson et al. 2012). Repetitive exposure to high impacts and excessive spinal ranges of motion during bowling have been hypothesised to contribute to this increased risk, yet research is still unclear on the exact mechanism of injury (Hurrion et al. 2000; Portus et al. 2004; Stuelcken et al. 2009; Crewe et al. 2013; Worthington et al. 2013). Direct comparisons between junior and senior fast bowlers in this study offer some insights into why junior bowlers may be at increased risk and the potential underlying mechanics.

5.1.4.1 Tibial accelerations

This is the first study to report tibial accelerations as a method of quantifying impact and therefore the results represent novel findings. Despite this some comparisons can be drawn to those studies using force-plates. This study demonstrated that peak tibial acceleration at BFI was significantly higher along the *y* axis in senior fast bowlers compared with junior fast bowlers. Previous studies typically report vertical impact or anterior-posterior impact characteristics, with only two recent studies investigating mediolateral impacts (Elliott et al. 1993; Hurrion et al. 2000; Stuelcken et al. 2009; Crewe et al. 2013; Worthington et al. 2013; Middleton et al. 2016). However, neither compared junior and senior fast bowling (Crewe et al. 2013; Middleton et al. 2016). It may be hypothesised that higher acceleration along the *y*- axis in seniors could be a result of a more front-on lower limb position at BFI (as, due to the elite level of participants, it is unlikely that there will be significantly higher momentum experienced perpendicular to the direction of delivery). However, as no lower limb kinematics were recorded simultaneously, this hypothesis requires further investigation. Furthermore, due to the tibial accelerometer not being able to be corrected for tilt, the y-axis will not truly be anatomically anterior-posterior or the z-axis truly mediolateral. Nonetheless, due to the orientation of the sensor on the tibia, it is likely that a front-on lower limb orientation will elicit larger magnitudes of acceleration in the y-axis, as a side-on lower limb position would elicit larger z-axis acceleration (both in the direction of delivery for

their respective orientations). The extent to which this difference may affect injury or performance needs further investigation.

A significantly longer time-to-peak resultant tibial acceleration at BFI was observed in senior fast bowlers compared with juniors. However, as peak resultant acceleration was greater in seniors no significant difference was observed in mean tibial loading rate. The longer time-to-peak displayed in seniors may be a mechanism to decrease stress placed on the musculoskeletal system during high impacts through damping. The hypothesis that senior bowlers displayed a more front-on lower limb orientation at BFI further supports this finding, as this would allow more effective knee flexion or ankle dorsiflexion to dissipate the impact over a longer period of time. Previous research on fast bowling and running supports this hypothesis reporting altered kinematics as a result of increased GRF or decreased impact attenuation capabilities, like that seen during barefoot running (Hardin et al. 2004; Worthington et al. 2013).

Furthermore, the fact that a moderate effect size highlighted, per kilogram of body mass, senior peak x axis acceleration at FFI is lower than that experienced by junior fast bowlers, may also be a result of regulatory mechanisms (such as increased knee flexion or ankle dorsiflexion) used by the more trained senior bowlers to help reduce these impacts (Worthington et al. 2013). It could be hypothesised that adults have greater eccentric quadriceps muscle strength which would serve as a brake to further knee flexion allowing the impact to be born on a 'soft' knee, whilst adolescents have lower quadriceps strength therefore adopting a strategy that utilises the inert tissues to absorb the load. Such hypotheses require further testing.

No significant differences were observed between tibial accelerations of senior and junior bowlers at FFI. However, high effect sizes were observed in time-to-peak tibial acceleration along the x axis (1.21) with a faster time-to-peak seen in junior fast bowlers. This is in agreement with findings seen at BFI and thus is likely a result of similar mechanisms discussed above. This shorter damping period would result in increases in 'stiffness' of viscoelastic tissues and thus may be important in the consideration of tissue stress.

5.1.4.2 Sacral Accelerations

Sacral accelerations have not previously been reported during cricket fast bowling, therefore these findings are novel. No significant differences were reported between sacral accelerations of junior and senior fast bowlers in this study. However, this may be due to the excessively cautious Bonferroni corrections applied minimising the risk of type 1 error but simultaneously increasing the risk of type II error (Streiner and Norman, 2011). Despite this, some large effect sizes were calculated. Peak mediolateral sacral acceleration and normalised peak vertical sacral acceleration at BFI displayed high effect sizes. These values suggest orientation of the lower limb does not dictate orientation of the pelvis in fast bowling, as greater acceleration anterior-posteriorly was seen at the tibia for seniors, compared to greater mediolateral acceleration at the sacrum. Additionally, junior fast bowlers encountering higher peak acceleration per kilogram may suggest that an analysis solely including magnitudes of impacts may not sufficiently describe the interaction between fast bowling impacts and injury risk. Lower impacts, experienced by junior fast bowlers may be comparatively more damaging than higher impacts experienced by senior bowlers once body mass and consequently musculoskeletal strength is factored in.

At FFI only one moderate effect size for normalised peak vertical sacral acceleration was observed, agreeing with the trend at BFI that, per kilogram of body mass, junior fast bowlers experience higher impacts than senior bowlers. However, it must be noted that whilst some trends have been highlighted by analysis of effect sizes, no statistically significant differences have been highlighted between junior and senior fast bowlers at FFI.

Whilst the magnitudes of impacts are similar between the two groups it is highly likely that the loading tolerance of the musculoskeletal structures is different. The immature skeletal structures and reduced muscle mass of adolescent bowlers would indeed place their musculoskeletal system at a higher risk of injury despite experiencing similar levels of tissue stresses. Therefore, these findings suggest that methods to either reduce rate and magnitude of impacts, or strengthen immature structures may be considered as methods to reduce injury. Alternatively, protection of the immature structures to such impacts allowing them time to strengthen may all be viable options to help in the quest of reducing injury risk in the high risk adolescent bowling population.

5.1.4.3 Spinal Kinematics

This study aimed to investigate differences in three-dimensional spinal kinematics of junior and senior fast bowlers during bowling. Previous research has highlighted that junior fast bowlers are at increased risk of lower back injury and pain compared with senior bowlers (Gray et al. 2000; Morton et al. 2013). Whilst this increased risk has partly been attributed to immature structures in junior fast bowlers' spines, a direct comparison of junior and senior three-dimensional spinal kinematics during bowling has not been investigated (Crewe et al. 2013). Previous studies have utilised different methodologies, and therefore comparisons between studies are difficult. Consequently, this study is the first to provide a direct comparison between spinal kinematics of junior and senior bowlers using identical methods and the first to highlight differences. This study highlights novel findings based on this comparison.

The findings of this study demonstrate that junior bowlers possess more thoracolumbar spinal rotation during standard ROM testing. Furthermore, all ranges of motion were larger for juniors with the exception of lumbar lateral flexion. Whilst higher ranges of motion are to be expected in younger people, this was not the case in junior fast bowlers with only one out of the nine movements analysed being significantly different (Intolo et al. 2009). The additional range displayed in thoracolumbar rotation of junior bowlers may be the result of immature spinal structures allowing extra ROM in rotation. However, it is possible that junior fast bowlers have less constraint to rotation ROM resulting in excess range and elevated risk. Additionally, the fact that other ranges of motion were not significantly different to senior bowlers may suggest that bowling at a young age develops increased range through task repetition and thus as a result of fast bowling at a young age greater range is developed as a senior bowler. Therefore, comparison of junior and senior ROM in this study conflicts results seen in the non-fast bowling population (Intolo et al. 2009).

This study analysed orientations of the lumbar, thoracic and thoracolumbar spine at back and front-foot impact, as well as ROM between these two time points. Previous research has only reported lumbar and thoracolumbar kinematics (Burnett et al. 1998; Ranson et al. 2008; Ferdinands et al. 2009; Ranson et al. 2009; Stuelcken et al. 2010; Crewe et al. 2013; Bayne et al. 2016). Thus, analysis of kinematics of the thoracic spine in this study provides new insights into spinal kinematics of junior and senior fast

bowling. No significant differences were found between junior and senior bowlers for any kinematic variables following Bonferroni correction, however a large effect size was observed for thoracolumbar rotation at BFI ($d=1.02$). Despite the lack of statistical significance, the actual thoracolumbar orientation at this point was 15 degrees more for the junior bowlers. This finding highlights larger spinal rotation away from the direction of bowling at BFI in junior bowlers. Previous findings have observed larger shoulder counter-rotation values in junior bowlers attempting to bowl faster and thus the difference observed in this study may be a result of a similar mechanism to compensate for any height, strength or power deficits in junior bowlers (Elliott et al. 2005). Moreover, the range of motion during the delivery stride was greater in the junior group by 44% (12.5 degrees) suggesting this group used more thoracolumbar rotation perhaps as a 'wind-up' to achieve bowling pace as has been previously suggested.

Whether this 'wind-up' has a similar effect of increasing risk of lower back injury and pain, as seen in excessive shoulder counter-rotation, is currently unknown. Whilst this finding may have been significant if Bonferroni corrections were not applied, the lack of differences throughout may suggest that spinal kinematics remain similar as fast bowlers mature. However, as a result of maturation, similar spinal kinematics may place different stresses on the spine for junior and senior bowlers. Furthermore, what may be considered 'excessive' range of motion for one bowler may be within normal range for another. Consequently, analysis of spinal kinematics in relation to bowlers' static range of motion may provide some individualised measure of magnitudes of spinal kinematics. Previous research has hypothesised that excessive spinal range of motion may increase bowlers' risk of lower back injury and pain (Glazier, 2010). However, it may be the case that a bowler with a larger range of motion in clinical tests may be able to bowl effectively within their normal range of motion without the need for 'excessive' ranges to generate pace on the ball. This study reports spinal kinematics during fast bowling as a percentage of range of motion obtained during pre-bowling static range of motion trials in order to investigate this hypothesis further. A large effect size highlighted larger range of lumbar right lateral flexion in junior bowlers ($d= 0.94$) and a moderate effect size showing larger lumbar extension in juniors ($d=0.55$). This may suggest that junior bowlers are extending and laterally flexing away from the direction of bowling (typically seen at BFI) as a mechanism of generating extra pace on the ball as suggested above. Whilst this may compensate for some of the strength deficits seen

during maturation, a combination of these two movements has been proposed as potentially injurious (Burnett et al. 2008; Glazier 2010).

In addition, a medium effect size also highlighted larger thoracolumbar right rotation and thoracic left lateral flexion in senior bowlers. Mean thoracolumbar right rotation during bowling in seniors was 185% of the thoracolumbar rotation recorded in their static range of motion trials. Potentially this is due to the momentum carried by the hips towards the direction of delivery whilst the shoulders are forced into rotating away from the direction of delivery as a mechanism of 'wind-up', thus eliciting a much larger ROM at BFI. This suggests that certainly all the available range was being used by this group to achieve fast bowling performance. Additionally, larger left lateral flexion is also likely a result of larger momentum seen in senior bowlers pushing past their active ROM. Loading at or moving tissues to the end of their available range results in significantly greater tissue loads and is associated with a lowering in load tolerance, thus potentially contributing to increased injury risk (Chosa et al. 2004; Smith et al. 2005; Swaminathan et al. 2016). However, these findings also suggest that junior and senior fast bowlers adopted different strategies for generating pace on the ball. Previous literature has reported that it may not be excessive uniplanar kinematics that increase risk of lower back injury but a combination of multiplanar movements that place the spine in a position of mechanical weakness (Glazier et al. 2010). As a result, the fact that junior bowlers displayed greater ranges of lumbar extension and right lateral flexion (movements utilised at the same point of the bowling action) may be an important finding when considering injury risk.

5.1.5 Conclusion

This study has highlighted that senior fast bowlers show greater y axis acceleration and time-to-peak acceleration during BFI at the tibia. This may be a result of a more 'front-on' technique and more effective regulation of loading rate seen in the more experienced senior bowlers. This hypothesis was supported by high effect sizes showing slower time-to-peak along tibia acceleration at FFI. Normalised results report comparatively higher loading rates in junior fast bowlers suggesting a potential mechanism for injury.

No significant differences were observed between spinal kinematics during the delivery stride in junior and senior bowlers. Thus, it may be assumed that differences in spinal

kinematics do not explain the junior bowlers' increased risk of lower back injury and pain compared to seniors. However, effect size analysis suggests juniors displayed more lumbar extension and rotation away from the direction of bowling at back-foot impact and through the delivery stride, possibly as a mechanism to generate more pace on the ball, whilst seniors utilise greater thoracic contributions to ROM. Current literature is unclear as to whether similar spinal kinematics have similar effects on vertebrae at different stages of maturation and thus whether junior and senior bowlers should receive similar recommendations is not known. Bowling data expressed as percentages of static ROM showed differing relationships in junior and senior bowlers: Junior bowlers utilising more ROM from their lumbar spine whilst senior bowlers increasing the contribution from the thoracic spine. Such findings may also provide clues to the higher prevalence of LBP in juniors, yet further research is needed to verify this. Consequently, this study recommends that further research focus on further analysis of the above findings alongside injury and pain data.

5.2 The effect of tibial and sacral accelerations and three dimensional spinal kinematics on risk of lower back pain and injury in junior and senior fast bowlers.

5.2.1 Introduction

Fast bowlers have been identified as having increased risk of injury compared with the rest of the team (Orchard et al. 2002; Orchard et al. 2006; Johnson et al. 2012). Lower limb and lower back injuries have the highest reported incidences among fast bowlers (Stretch, 2003). Junior fast bowlers have been reported to be at significantly higher risk of lower back injury and pain when compared with senior fast bowlers, however previously in this chapter few differences in impact characteristics were observed between junior and senior bowlers (Gray et al. 2000; Johnson et al. 2012). These findings agree with previous work suggesting that immature structures in the spine (such as the pars interarticularis) and weaknesses in the surrounding stabilising muscles place junior bowlers at increased risk of lower back pain (LBP) and injury, not differences in technique (Kippers et al. 1998). However, the analysis of spinal kinematics in this chapter suggests that there may be differences between junior and senior bowlers, however their link to pain is currently unknown. Whilst high workloads have been hypothesised to increase risk of injury, research is yet to fully explain the relationship of impact and kinematic characteristics with fast bowling injuries and pain (Dennis et al. 2005).

Current research analysing fast bowling impacts have used force plates, with which we can only gain insight into the interaction between the foot and the plate through analysis of ground reaction force (Elliott and Foster, 1984; Foster and Elliott, 1985; Elliott et al. 1986; Foster et al. 1989; Mason et al. 1989; Elliott et al. 1992; Hurrion et al. 2000; Portus et al. 2004; Crewe et al. 2013; Worthington et al., 2013). Though some studies have used inverse dynamics models to assess impacts further up the body, they have not analysed this data in relation to LBP, therefore limiting our understanding of the relationship to ground reaction force variables (Crewe et al. 2013; Bayne et al. 2016). This thesis has previously highlighted accelerometers as a valid and reliable method of analysing fast bowling impacts at the tibia and sacrum that may address some of these limitations.

To date only shoulder counter-rotation (SCR) in excess of 30° has been highlighted to increase risk of lower back injury (Elliott et al. 1993; Portus et al. 2004; Morton et al.

2013). SCR only describes shoulder orientation and thus does not describe true three-dimensional spinal kinematics (Ranson et al. 2008). Therefore, the relationship between three-dimensional kinematics and back pain remains unclear. Currently studies have failed to identify a link between three-dimensional spinal kinematics and back pain, however this may be due to methodological limitations of those studies (Morton et al. 2013). Due to the elite and very specific population required sample sizes tend to be low, however as studies adopt different methods of spinal segmentation, direct comparisons or data pooling of reported spinal kinematics cannot be carried out (Stuelcken et al. 2010; Crewe et al. 2013b; Bayne et al. 2016).

Whilst fast bowling impacts, spinal kinematics and lower back injury have been reported in numerous studies, their relationship with LBP has received less attention (Portus et al. 2000; Johnson et al. 2012; Spratford and Hicks, 2014). Consequently, even less is known in regards to this relationship. No research has looked at the relationship between injury or LBP from both a retrospective and prospective viewpoint simultaneously. Consequently, by making comparisons between studies reporting different variables with different populations, it is difficult to gain an understanding of the impact and relationship of each factor within a holistic model.

5.2.2 Aim of the study

This study aims to investigate the relationship between impact characteristics at the tibia and sacrum and three-dimensional spinal kinematics during fast bowling, history of LBP and risk of future LBP in junior and senior fast bowlers.

5.2.3 Results

5.2.3.1 Junior Fast Bowlers

5.2.3.1.1 Prevalence and incidence of lower back pain

Eight junior fast bowlers reported a retrospective history of LBP at some point in their playing career, representing 38% of the sample. However, only two of these had imaging to confirm one with a herniated disc and the other stress fractures located at the pars interarticularis on the contralateral side of L4 and L5. Most bowlers reported no cricket missed as a result of pain. An entire season was missed as a result of pain due to the lumbar stress fractures. Two bowlers reported time away from cricket of four and six weeks as a result of undiagnosed lower back pain. No cricket was missed as a result

of the herniated disc due to its occurrence at the end of the season. The mean (±SD) age of first occurrence of LBP was 15.4 ±1.1 years.

Injury surveillance of junior fast bowlers in the 2015 season only highlighted one instance of LBP which resulted in three weeks of cricket being missed. No formal diagnosis was given for this pain, and following the three weeks missed the bowler was able to continue bowling without pain. The bowler had reported LBP, again with no formal diagnosis, prior to the 2015 season with no previous repeated occurrences until the 2015 season.

5.2.3.1.2 Fast bowling impact characteristics

A comparison of impact characteristics between junior fast bowlers with and without a history of LBP can be found in table 5.2.1. Following Bonferroni correction for multiple significance tests, no variables were significantly different between the two groups. However, peak mediolateral sacral acceleration was higher at front-foot impact in the 'no history of LBP' group, displaying a high effect size (>0.8).

Moderate effect sizes (>0.5) were observed in six variables. Peak tibial acceleration along the z axis, peak resultant tibial acceleration and normalised peak resultant tibial acceleration at BFI was larger in the 'no history of LBP' group. In addition, time-to-peak tibial acceleration along the x axis was faster in the 'no history of LBP' group at BFI. Peak mediolateral sacral acceleration and mean sacral vertical loading rate were also greater in the 'no history of LBP' group. No moderate effect sizes were observed at FFI.

As only one bowler developed LBP in the 2015 season no statistical prospective analysis of junior fast bowling impact characteristics was possible.

5.2.3.1.3 Fast bowling spinal kinematics

After Bonferroni corrections, no significant differences in any spinal kinematics were observed between bowlers that had a history of LBP and bowlers that did not. However, some large effect sizes were observed.

The comparison of three-dimensional spinal kinematics in junior fast bowlers with and without history of LBP (retrospective) (table 5.2.2) highlighted that bowlers without a history of LBP displayed larger thoracic rotation away from the direction of delivery at

BFI than bowlers with LBP history as seen in figure 5.2.1 (d= 1.27). Consequently, range of thoracic rotation between BFI and FFI was larger in bowlers without a history of LBP (d=0.90). Moderate effect sizes highlighted bowlers with no LBP history bowled with a slightly more 'side-on' pelvis position (S1 orientation at BFI). Lumbar lateral flexion and rotation at FFI were also larger in bowlers with no LBP history and consequently moderate effect sizes were also observed for range of lumbar lateral flexion and rotation between BFI and FFI. When these kinematics were expressed as a percentage of static range of motion, no significant differences were highlighted, but some large effect sizes were calculated. Thoracolumbar flexion and extension were smaller in bowlers without a history of LBP (d= 1.31 and 1.79 respectively).

As only one junior bowler went on to develop LBP in the 2015 season no prospective analysis on spinal kinematics and LBP risk in junior bowlers was able to be carried out.

Table 5.2.1. Mean (±SD) tibial and sacral impact characteristics of junior fast bowlers with (n=8) and without (n=13) a history of lower back pain.

	LBP (n = 8)		**No LBP (n = 13)**		*p*-value		Effect size (*d*)	
Tibial Acceleration	**Back-Foot Impact**	**Front -Foot Impact**	**Back-Foot Impact**	**Front-Foot Impact**	BFI	FFI	BFI	FFI
Peak Tibial Acc *x* (g)	10.1 (4.6)	28.0 (13.2)	12.0 (4.6)	25.0 (5.4)	.283	.557	-0.41	0.33
Peak Tibial Acc *y* (g)	2.5 (1.7)	15.5 (9.2)	2.8 (1.9)	12.0 (6.8)	.676	.327	-0.18	0.44
Peak Tibial Acc *z* (g)	13.0 (5.3)	19.1 (8.9)	17.6 (10.7)	16.3 (6.3)	.209	.449	-0.50	0.38
Resultant Tibial Acc (g)	15.4 (5.2)	36.2 (17.5)	19.8 (8.0)	31.3 (7.2)	.141	.470	-0.63	0.41
Time-to-peak Tibial Acc *x* (ms)	28.1 (18.1)	16.8 (4.3)	19.6 (6.9)	16.4 (2.7)	.238	.982	0.69	0.09
Time-to-peak Resultant Tibial Acc (ms)	39.8 (11.6)	59.0 (18.8)	50.6 (28.8)	56.5 (8.5)	.305	.735	-0.45	0.19
Mean Tibial Loading Rate *x* (g.s^{-1})	581.1 (396.4)	1848.0 (959.1)	715.2 (288.1)	1621.1 (378.5)	.423	.540	-0.40	0.35
Mean Resultant Tibial Loading Rate (g.s^{-1})	465.8 (276.8)	897.5 (937.0)	480.8 (232.1)	644.6 (194.4)	.831	.668	-0.06	0.43
Normalised Peak Tibial Acc *x* (g.kg^{-1})	0.2 (0.1)	0.4 (0.2)	0.2 (0.1)	0.3 (0.1)	.416	.474	-0.35	0.40
Normalised Resultant Tibial Acc (g.kg^{-1})	0.2 (0.1)	0.5 (0.2)	0.3 (0.1)	0.4 (0.1)	.220	.479	-0.54	0.40
Sacral Acceleration								
Peak Vertical Sacral Acc (g)	2.5 (0.7)	3.1 (0.6)	2.7 (0.8)	3.2 (0.7)	.366	.697	-0.40	-0.17
Peak Mediolateral Sacral Acc (g)	0.5 (0.1)	0.6 (0.2)	0.7 (0.3)	1.1 (0.7)	.165	.028*	-0.65	-0.89
Peak Anterior-Posterior Acc (g)	1.5 (0.7)	1.6 (1.0)	1.3 (0.7)	1.4 (0.9)	.492	.582	0.32	0.23
Resultant Sacral Acc (g)	3.0 (0.8)	3.7 (0.8)	3.2 (0.8)	4.0 (0.8)	.691	.456	-0.18	-0.33
Time-to-peak Vertical Sacral Acc (ms)	70.1 (19.0)	60.6 (16.5)	72.9 (18.5)	64.1 (10.8)	.749	.185	-0.15	-0.26
Time-to-peak Resultant Sacral Acc (ms)	73.6 (22.2)	65.1 (12.8)	75.0 (20.9)	68.2 (13.6)	.891	.316	-0.06	-0.16
Mean Sacral Vertical Loading Rate (g.s^{-1})	38.7 (8.3)	61.2 (21.5)	46.0 (16.9)	63.7 (17.2)	.204	.837	-0.51	-0.10
Normalised Peak Vertical Sacral Acc (g.kg^{-1})	0.04 (0.01)	0.05 (0.01)	0.04 (0.01)	0.05 (0.02)	.537	.619	-0.28	-0.22
Resultant Attenuation (%)	22.1 (8.2)	13.0 (7.0)	18.5 (8.4)	13.5 (4.0)	.313	.856	0.42	-0.10

LBP, lower back pain; acc, acceleration; g, gravity; ms, milliseconds; s, seconds; kg, kilograms; *, p<.003.

Table 5.2.2. Mean (±SD) three-dimensional spinal kinematics of junior fast bowlers with (n=8) and without (n=13) a history of lower back pain.

Spinal Kinematics (°)	BFI				FFI				Range			
	LBP	No LBP	P value	Effect Size (*d*)	LBP	No LBP	P value	Effect Size (*d*)	LBP	No LBP	P value	Effect Size (*d*)
Shoulder counter-rotation	39.7 (8.9)	36.6 (11.3)	.498	0.29								
Hip-shoulder separation	27.0 (7.6)	32.6 (18.0)	.340	-0.37								
T1 orientation	271.2 (42.8)	270.8 (35.1)	.979	0.01								
S1 orientation	257.1 (35.9)	277.6 (40.0)	.140	-0.53								
Lumbar flexion	-13.1 (21.9)	-14.1 (15.2)	.919	0.05	19.8 (12.7)	20.9 (7.3)	.832	-0.11	32.9 (33.3)	34.9 (21.4)	.883	-0.08
Lumbar lateral flexion	-8.6 (6.4)	-12.5 (12.5)	.358	0.37	17.1 (7.9)	21.7 (10.2)	.253	-0.50	25.7 (9.1)	34.2 (18.7)	.177	-0.54
Lumbar rotation	-2.9 (3.6)	-5.0 (9.3)	.804	0.28	11.4 (5.1)	15.7 (9.4)	.193	-0.53	14.3 (6.3)	20.8 (15.2)	.195	-0.51
Thoracic flexion	-33.4 (12.1)	-28.6 (17.3)	.468	-0.30	27.3 (12.4)	29.1 (18.4)	.800	-0.11	60.7 (20.9)	57.7 (32.4)	.797	0.11
Thoracic lateral flexion	-13.1 (19.6)	-21.2 (15.0)	.537	0.48	26.2 (9.7)	24.1 (11.3)	.659	0.19	39.2 (24.1)	45.2 (18.2)	.554	-0.29
Thoracic rotation	-3.9 (9.8)	-15.5 (8.8)	.016	1.27	16.6 (7.6)	20.5 (17.6)	.801	-0.27	20.5 (13.3)	36.0 (19.1)	.041	-0.90
Thoracolumbar flexion	-43.0 (26.2)	-37.3 (19.3)	.606	-0.26	43.3 (17.6)	38.2 (20.1)	.553	0.26	86.2 (41.3)	75.5 (37.9)	.429	0.27
Thoracolumbar lateral flexion	-21.9 (8.7)	-25.4 (11.7)	.443	0.33	29.0 (10.2)	29.1 (15.6)	.988	-0.01	51.0 (17.2)	54.5 (22.6)	.686	-0.17
Thoracolumbar rotation	-17.6 (7.2)	-14.0 (12.4)	.413	-0.33	23.4 (10.1)	27.1 (15.5)	.519	-0.27	41.0 (16.6)	41.1 (22.0)	.992	0.00

BFI, back-foot impact; FFI, front-foot impact; °, degrees; SCR, shoulder counter-rotation; HSS, hip-shoulder separation; LBP, lower back pain.
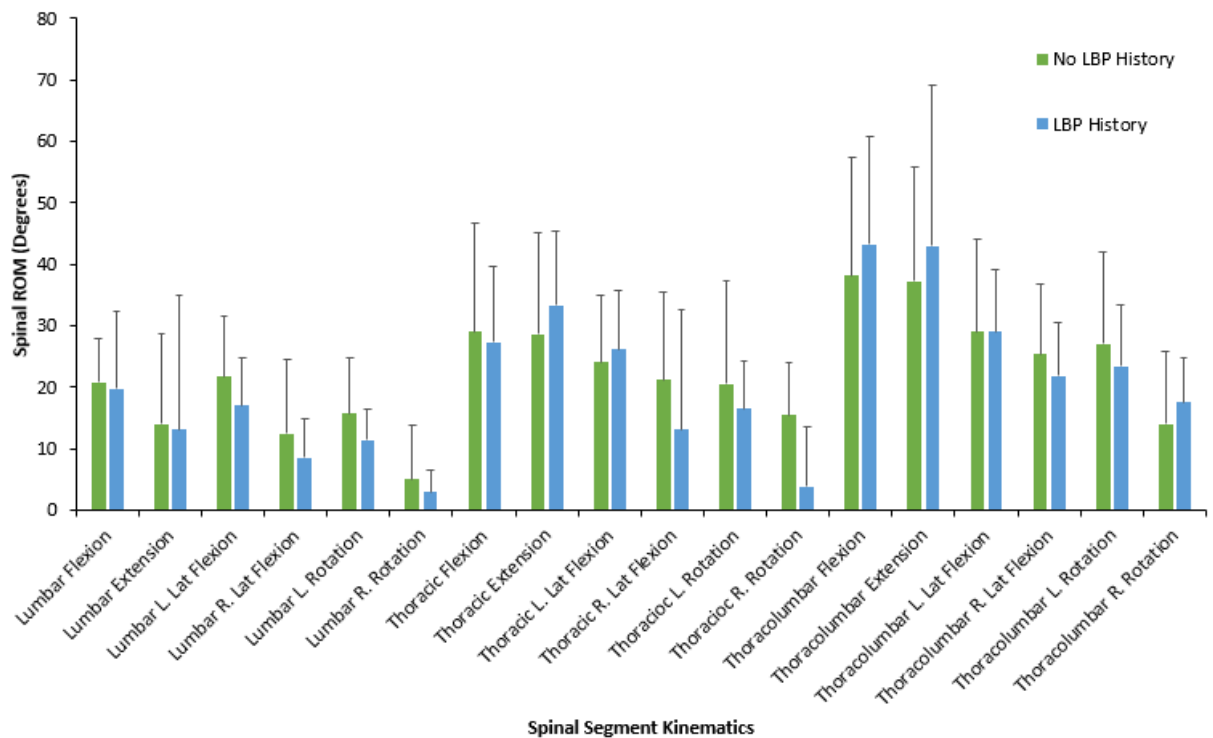
Figure 5.2.1. A comparison of mean three-dimensional spinal kinematics during fast bowling in junior bowlers with (n=8) and without (n=13) a history of lower back pain.

5.2.3.2 Senior fast bowlers

5.2.3.2.1 Prevalence and incidence of lower back pain

History of LBP was reported in 57% of the senior fast bowler sample. 50% of LBP was a symptom of lumbar stress fractures (n = 4, radiologically confirmed). Two bowlers were diagnosed with stress fractures on the contralateral side of the L5 vertebrae, one had contralateral stress fractures at L3, 4 and 5, whilst the remaining bowler was diagnosed with bilateral stress fractures at L3, 4 and 5. One bowler's LBP was a result of herniated discs at L3/4 and L4/5 whilst the remaining three bowlers' LBP had no formal radiographical diagnosis. The mean age (±SD) of first occurrence was 16.6 ± 2.6 years. The mean time missed due to LBP was 15.6 weeks. This time increases to 23.7 weeks away from cricket when only LBP as a symptom of a diagnosed lower back pathology is taken into account. Only two cases reported no cricket missed, both of whom had no formal diagnosis. One bowler had diagnosed bilateral lumbar stress fractures at L5, but pain was not a symptom.

Injury surveillance of senior fast bowlers in the 2015 season reported four instances of LBP. Three instances were due to muscle strains, resulting in two weeks, three weeks and six weeks cricket missed. The fourth case was due to bilateral stress fractures at the

pars interarticularis at L4 and L5 resulting in the whole season being missed. All four bowlers had reported a previous history of LBP. The bowlers that developed muscle strains all had history of lumbar stress fractures or herniated discs. The bowler that was diagnosed with lumbar stress fractures during the 2015 season had no previously diagnosed pathology, but had reported a history of LBP. Consequently, a risk ratio of 4.0 was calculated; reporting that risk of future LBP is four times greater if you have been previously diagnosed with a spinal pathology compared with bowlers with no history of spinal pathology.

5.2.3.2.2 Fast bowling impact characteristics

A comparison of impact characteristics between senior fast bowlers with and without a history of LBP can be seen in table 5.2.3. Following Bonferroni correction for multiple significance tests, no variables were significantly different between the two groups. However, some large effect sizes were observed. Peak tibial acceleration along the z axis at BFI was greater in the 'no history of LBP' group and time-to-peak resultant tibial acceleration was faster in the 'LBP' group both with effect sizes >1.4.

A comparison of impact characteristics of senior fast bowlers that did or did not develop LBP in the 2015 season can be seen in table 5.2.4. After Bonferroni correction for multiple significance tests no significant differences were observed. However, several large effect sizes were reported. Peak tibial acceleration along the z axis and resultant tibial acceleration were higher in the 'no LBP' group at BFI. Time-to-peak vertical and resultant sacral acceleration was also faster in the 'no LBP' group at BFI.

At FFI large effect sizes were observed in a number of variables. Peak tibial acceleration along the y axis, peak resultant acceleration and normalised resultant tibial acceleration were larger in the 'no LBP' group. Time-to-peak tibial acceleration along the x axis was faster in the 'no LBP' group, consequently, tibial loading rate (along the x axis and resultant) was larger in the 'no LBP' group. Additionally, peak sacral acceleration (vertical, mediolateral and resultant) were larger in the 'no LBP' group. A summary of these findings can be found in table 5.2.7.

5.2.3.2.3 Fast bowling spinal kinematics

Senior fast bowlers with and without a history of LBP (retrospective) display a different relationship compared with junior bowlers. After Bonferroni corrections, no significant

differences in any spinal kinematics were observed between bowlers that had a history of LBP and bowlers that did not (Table 5.2.5). Large effect sizes were observed in thoracolumbar extension at BFI, with bowlers without a history of LBP displaying less extension as seen in figure 5.2.2 (d=1.02). Consequently, smaller range of thoracolumbar flexion/extension was observed between BFI and FFI in bowlers without a history of LBP (d=0.88). Moderate effect sizes highlighted that lumbar lateral flexion away from the direction of delivery at BFI was larger in bowlers with no history of LBP. Less lumbar and thoracic flexion at FFI was observed in bowlers without a history of LBP. Thoracic and thoracolumbar lateral flexion at FFI was also smaller in bowlers without LBP history. Range of thoracic flexion between BFI and FFI was smaller in bowlers without a history of LBP.

Kinematics expressed as a percentage of static range of motion produced no significant differences but some large effect sizes. Lumbar flexion, thoracic extension and thoracic right lateral flexion were larger in bowlers with no LBP history (d= 1.32, 0.89 and 1.05). Conversely, thoracic left and right rotation (d= 1.16 and 1.80), and left and right thoracolumbar lateral flexion (d= 0.87 and 1.75) and left and right rotation (d= 1.29 and 1.69) were smaller in bowlers without a history of LBP.

Analysis of senior bowlers that developed LBP in the 2015 season and those that did not (prospective) produced some large effect sizes (Table 5.2.6). Bowlers that did not develop LBP displayed less lumbar extension at BFI seen in figure 5.2.3 (d=1.92). Bowlers that did not develop LBP also displayed smaller lumbar flexion at FFI and therefore displayed smaller range of lumbar flexion/extension between BFI and FFI. Hip-shoulder separation was also smaller in bowlers that did not developed LBP (d=0.82). Additionally, lumbar lateral flexion away from the direction of delivery at BFI was larger in bowlers that did not develop lower back pain (d=1.03). Lumbar rotation at FFI was also larger in bowlers that did not develop LBP (d= 1.26), as was range of thoracic rotation between BFI and FFI (d=0.90).

Comparisons of kinematics expressed as a percentage of static ROM produced some large effect sizes. Lumbar flexion and right rotation were larger in bowlers that did not develop LBP (d= 1.31 and 0.94). Thoracic right rotation and thoracolumbar flexion were also larger in the 'no prospective LBP' group (d= 1.06 and 0.81). Conversely, lumbar left rotation, thoracic flexion and thoracolumbar right lateral flexion was smaller

in the 'no LBP' group ($d=$ 2.33, 1.26 and 1.15). A summary of these findings can be found in table 5.2.8.

Table 5.2.3. Mean (±SD) tibial and sacral impact characteristics of senior fast bowlers with (n=8) and without (n=6) a history of lower back pain.

	LBP (n = 8)		No LBP (n = 6)		*p*-value		Effect size (*d*)	
Tibial Acceleration	**Back-Foot Impact**	**Front-Foot Impact**	**Back-Foot Impact**	**Front-Foot Impact**	BFI	FFI	BFI	FFI
Peak Tibial Acc *x* (g)	13.7 (7.4)	21.8 (10.0)	14.4 (6.5)	28.3 (17.3)	.843	.392	-0.11	-0.44
Peak Tibial Acc *y* (g)	6.8 (5.7)	10.1 (8.2)	6.7 (3.8)	11.8 (10.0)	.982	.725	0.01	-0.19
Peak Tibial Acc *z* (g)	10.0 (5.2)	29.4 (20.7)	20.28 (7.9)	21.3 (11.5)	.016	.402	-1.49	0.51
Resultant Tibial Acc (g)	20.7 (5.4)	37.5 (19.0)	24.9 (8.1)	38.8 (20.7)	.342	.911	-0.53	-0.06
Time-to-peak Tibial Acc *x* (ms)	31.6 (2.4)	28.3 (12.0)	27.9 (8.6)	26.9 (15.7)	.274	.345	0.55	0.10
Time-to-peak Resultant Tibial Acc (ms)	65.9 (5.4)	49.6 (10.7)	67.3 (12.7)	67.0 (12.8)	.796	.017	-0.13	-1.45
Mean Tibial Loading Rate *x* (g.s^{-1})	486.8 (229.6)	1112.0 (665.9)	600.5 (306.3)	1673 (1328.0)	.443	.324	-0.41	-0.51
Mean Resultant Tibial Loading Rate (g.s^{-1})	347.5 (133.8)	825.9 (453.7)	411.3 (145.8)	736.8 (459.0)	.413	.724	-0.45	0.20
Normalised Peak Tibial Acc *x* (g.kg^{-1})	0.2 (0.1)	0.2 (0.1)	0.2 (0.1)	0.3 (0.2)	.722	.384	-0.20	-0.46
Normalised Resultant Tibial Acc (g.kg^{-1})	0.2 (0.1)	0.4 (0.2)	0.3 (0.1)	0.4 (0.2)	.214	.707	-0.71	-0.20
Sacral Acceleration								
Peak Vertical Sacral Acc (g)	3.1 (0.8)	3.2 (0.4)	2.7 (0.7)	3.3 (0.3)	.389	.609	0.49	-0.28
Peak Mediolateral Sacral Acc (g)	1.5 (1.1)	0.9 (0.4)	1.2 (0.7)	1.2 (0.7)	.568	.372	0.35	-0.46
Peak Anterior-Posterior Acc (g)	1.4 (0.9)	1.5 (0.7)	1.7 (0.9)	1.4 (0.6)	.538	.902	-0.34	0.07
Resultant Sacral Acc (g)	3.5 (1.1)	3.8 (0.6)	3.4 (0.8)	4.0 (0.7)	.832	.677	0.12	-0.22
Time-to-peak Vertical Sacral Acc (ms)	71.7 (26.1)	66.7 (14.4)	67.1 (14.8)	69.0 (14.1)	.710	.772	0.23	-0.16
Time-to-peak Resultant Sacral Acc (ms)	75.2 (25.4)	67.1 (17.6)	70.0 (15.2)	69.8 (17.2)	.643	.804	0.25	-0.16
Mean Sacral Vertical Loading Rate (g.s^{-1})	50.9 (24.6)	60.1 (16.8)	46.3 (21.4)	54.3 (14.2)	.718	.510	0.20	0.38
Normalised Peak Vertical Sacral Acc (g.kg^{-1})	0.03 (0.01)	0.04 (0.01)	0.03 (0.01)	0.04 (0.01)	.674	.852	0.23	-0.17
Resultant Attenuation (%)	18.4 (7.1)	14.5 (9.0)	14.4 (3.5)	14.1 (7.6)	.246	.939	0.75	0.04

LBP, lower back pain; acc, acceleration; g, gravity; ms, milliseconds; s, seconds; kg, kilograms.

Table 5.2.4. Mean (±SD) tibial and sacral impact characteristics of senior fast bowlers that did (n=4) and did not (n=10) develop lower back pain in the following season.

	LBP (n = 4)		No LBP (n = 10)		*p*-value		Effect size (*d*)	
Tibial Acceleration	**Back-Foot Impact**	**Front-Foot Impact**	**Back-Foot Impact**	**Front-Foot Impact**	BFI	FFI	BFI	FFI
Peak Tibial Acc x (g)	10.8 (6.3)	17.5 (10.9)	15.4 (6.6)	28.7 (15.0)	.217	.162	-0.70	-0.79
Peak Tibial Acc y (g)	5.7 (2.7)	3.8 (2.0)	7.1 (5.1)	14.0 (9.0)	.616	.006	-0.32	-1.30
Peak Tibial Acc z (g)	10.7 (6.1)	17.8 (6.0)	18.0 (8.6)	27.6 (18.0)	.116	.177	-0.89	-0.61
Resultant Tibial Acc (g)	19.6 (2.4)	23.7 (9.0)	24.5 (9.0)	44.0 (19.4)	.139	.022	-0.62	-1.16
Time-to-peak Tibial Acc x (ms)	27.1 (6.2)	35.39 (15.8)	30.5 (7.0)	24.3 (12.2)	.406	.142	-0.50	0.84
Time-to-peak Resultant Tibial Acc (ms)	58.4 (12.3)	61.8 (13.8)	70.0 (7.0)	58.6 (15.5)	.155	.720	-1.34	0.21
Mean Tibial Loading Rate x (g.s^{-1})	459.8 (180.0)	744.18 (678.2)	588.5 (301.9)	1708.1 (1136.0)	.351	.081	-0.47	-0.93
Mean Resultant Tibial Loading Rate (g.s^{-1})	389.3 (119.2)	461.0 (163.7)	381.8 (152.6)	900.5 (459.4)	.925	.022	0.05	-1.08
Normalised Peak Tibial Acc x (g.kg^{-1})	0.1 (0.1)	0.2 (0.1)	0.2 (0.1)	0.3 (0.2)	.240	.231	-0.69	-0.73
Normalised Resultant Tibial Acc (g.kg^{-1})	0.2 (0.0)	0.3 (0.1)	0.3 (0.1)	0.5 (0.2)	.138	.034	-0.62	-1.22
Sacral Acceleration								
Peak Vertical Sacral Acc (g)	2.7 (0.8)	3.0 (0.5)	2.9 (0.7)	3.3 (0.3)	.721	.328	-0.24	-0.87
Peak Mediolateral Sacral Acc (g)	1.6 (0.8)	0.7 (0.5)	1.2 (0.9)	1.2 (0.6)	.525	.147	0.37	-0.88
Peak Anterior-Posterior Acc (g)	1.5 (1.2)	1.2 (0.8)	1.6 (0.8)	1.6 (0.6)	.811	.437	-0.18	-0.59
Resultant Sacral Acc (g)	3.2 (1.3)	3.5 (0.7)	3.5 (0.7)	4.0 (0.6)	.814	.219	-0.33	-0.89
Time-to-peak Vertical Sacral Acc (ms)	88.3 (18.7)	70.8 (13.9)	61.3 (14.5)	66.8 (14.2)	.054	.647	1.73	0.28
Time-to-peak Resultant Sacral Acc (ms)	89.8 (16.6)	72.3 (14.6)	64.2 (16.4)	69.0 (11.6)	.163	.772	1.55	0.25
Mean Sacral Vertical Loading Rate (g.s^{-1})	33.2 (9.6)	53.7 (19.0)	54.3 (23.0)	58.0 (14.2)	.031	.705	-1.03	-0.27
Normalised Peak Vertical Sacral Acc (g.kg^{-1})	0.03 (0.01)	0.04 (0.01)	0.03 (0.01)	0.04 (0.00)	.574	.454	-0.33	-0.64
Resultant Attenuation (%)	16.7 (5.6)	18.6 (9.2)	15.9 (5.7)	12.5 (7.1)	.825	.265	0.14	0.78

LBP, lower back pain; acc, acceleration; g, gravity; ms, milliseconds; s, seconds; kg, kilograms.

Table 5.2.5. Mean (±SD) three-dimensional spinal kinematics of senior fast bowlers with (n=8) and without (n=6) a history of lower back pain.

Spinal Kinematics (°)	BFI				FFI				Range			
	LBP	No LBP	P value	Effect Size (d)	LBP	No LBP	P value	Effect Size (d)	LBP	No LBP	P value	Effect Size (d)
Shoulder counter-rotation	35.7 (11.7)	35.6 (6.5)	1.00	0.01								
Hip-shoulder separation	41.2 (30.8)	41.8 (10.7)	.963	-0.03								
T1 orientation	274.6 (34.6)	264.1 (37.9)	.600	0.29								
S1 orientation	292.3 (10.8)	281.7 (32.6)	.412	0.41								
Lumbar flexion	-15.3 (11.0)	-14.7 (8.0)	.908	-0.07	24.2 (6.7)	21.1 (5.7)	.345	0.50	39.5 (15.2)	34.3 (11.1)	.492	0.41
Lumbar lateral flexion	-8.6 (12.2)	-15.8 (5.9)	.228	0.79	19.6 (5.5)	20.0 (7.1)	.910	-0.06	28.3 (14.8)	33.0 (7.6)	.501	-0.42
Lumbar rotation	-1.5 (11.3)	0.8 (8.2)	.950	-0.24	14.7 (6.5)	14.4 (6.0)	.932	0.05	16.2 (14.8)	13.6 (7.2)	.703	0.24
Thoracic flexion	-26.9 (15.2)	-22.8 (27.3)	.731	-0.18	40.0 (14.7)	31.7 (11.5)	.281	0.64	66.8 (27.9)	48.9 (32.3)	.288	0.59
Thoracic lateral flexion	-11.9 (17.6)	-10.5 (11.9)	.871	-0.10	33.9 (17.5)	26.1 (9.4)	.356	0.58	45.8 (33.4)	34.0 (16.4)	.455	0.47
Thoracic rotation	-7.1 (9.8)	-8.9 (13.8)	.778	0.15	19.7 (11.5)	19.9 (5.9)	.963	-0.03	26.7 (11.7)	28.8 (17.5)	.797	-0.13
Thoracolumbar flexion	-40.5 (10.2)	-29.4 (11.3)	.079	-1.02	48.4 (13.0)	45.0 (11.3)	.625	0.28	88.9 (22.2)	70.8 (19.3)	.143	0.88
Thoracolumbar lateral flexion	-19.2 (23.5)	-20.3 (17.1)	.923	0.06	36.1 (9.7)	31.1 (9.2)	.347	0.54	55.3 (24.6)	51.4 (19.)	.755	0.18
Thoracolumbar rotation	0.3 (14.7)	-0.3 (24.2)	.949	0.03	27.1 (7.3)	29.5 (12.6)	.652	-0.23	26.7 (15.3)	29.9 (20.6)	.748	-0.17

BFI, back-foot impact; FFI, front-foot impact; °, degrees; SCR, shoulder counter-rotation; HSS, hip-shoulder separation; LBP, lower back pain.

Table 5.2.6. Mean (±SD) three-dimensional spinal kinematics of senior fast bowlers that did (n=4) and did not (n=10) develop lower back pain in the following season.

Spinal Kinematics (°)	BFI				FFI				Range			
	LBP	No LBP	P value	Effect Size (*d*)	LBP	No LBP	P value	Effect Size (*d*)	LBP	No LBP	P value	Effect Size (*d*)
Shoulder counter-rotation	39.3 (11.2)	34.1 (7.6)	.443	0.60								
Hip-shoulder separation	53.2 (29.5)	36.9 (15.5)	.356	0.82								
T1 orientation	268.8 (32.8)	268.5 (38.3)	.987	0.01								
S1 orientation	284.7 (19.7)	286.9 (28.3)	.870	-0.08								
Lumbar flexion	-24.3 (7.7)	-11.2 (6.5)	.031	-1.92	26.3 (2.6)	20.9 (6.5)	.047	0.93	50.6 (8.5)	30.9 (9.3)	.008	2.16
Lumbar lateral flexion	-6.3 (13.9)	-15.3 (6.3)	.287	1.03	19.3 (5.1)	20.1 (6.8)	.833	-0.11	25.6 (14.4)	33.1 (9.4)	.539	-0.69
Lumbar rotation	-0.3 (11.6)	-0.1 (9.0)	.830	-0.02	19.3 (7.9)	12.6 (4.1)	.188	1.26	19.6 (13.3)	12.8 (9.5)	.396	0.65
Thoracic flexion	-16.5 (8.3)	-27.8 (25.5)	.236	0.50	30.8 (15.3)	37.0 (12.6)	.509	-0.46	47.3 (20.4)	60.3 (34.3)	.404	-0.41
Thoracic lateral flexion	-7.4 (24.7)	-12.5 (8.4)	.708	0.36	35.0 (21.6)	27.2 (9.3)	.534	0.57	42.4 (45.0)	37.7 (14.2)	.852	0.18
Thoracic rotation	-2.8 (16.1)	-10.2 (9.9)	.441	0.63	22.0 (13.3)	18.9 (6.2)	.685	0.36	24.8 (16.6)	29.1 (14.7)	.667	-0.29
Thoracolumbar flexion	-33.6 (14.8)	-34.4 (11.5)	.930	0.06	47.3 (17.1)	46.1 (9.9)	.905	0.10	80.9 (31.0)	77.6 (19.1)	.854	0.14
Thoracolumbar lateral flexion	-20.9 (24.5)	-19.4 (18.2)	.918	-0.07	33.3 (14.1)	33.2 (7.9)	.987	0.01	54.2 (23.7)	52.6 (21.4)	.911	0.07
Thoracolumbar rotation	4.6 (7.0)	-1.9 (23.4)	.446	0.31	29.9 (4.2)	27.9 (12.2)	.661	0.18	25.3 (29.8)	29.8 (20.8)	.582	-0.24

BFI, back-foot impact; FFI, front-foot impact; °, degrees; SCR, shoulder counter-rotation; HSS, hip-shoulder separation; LBP, lower back pain.
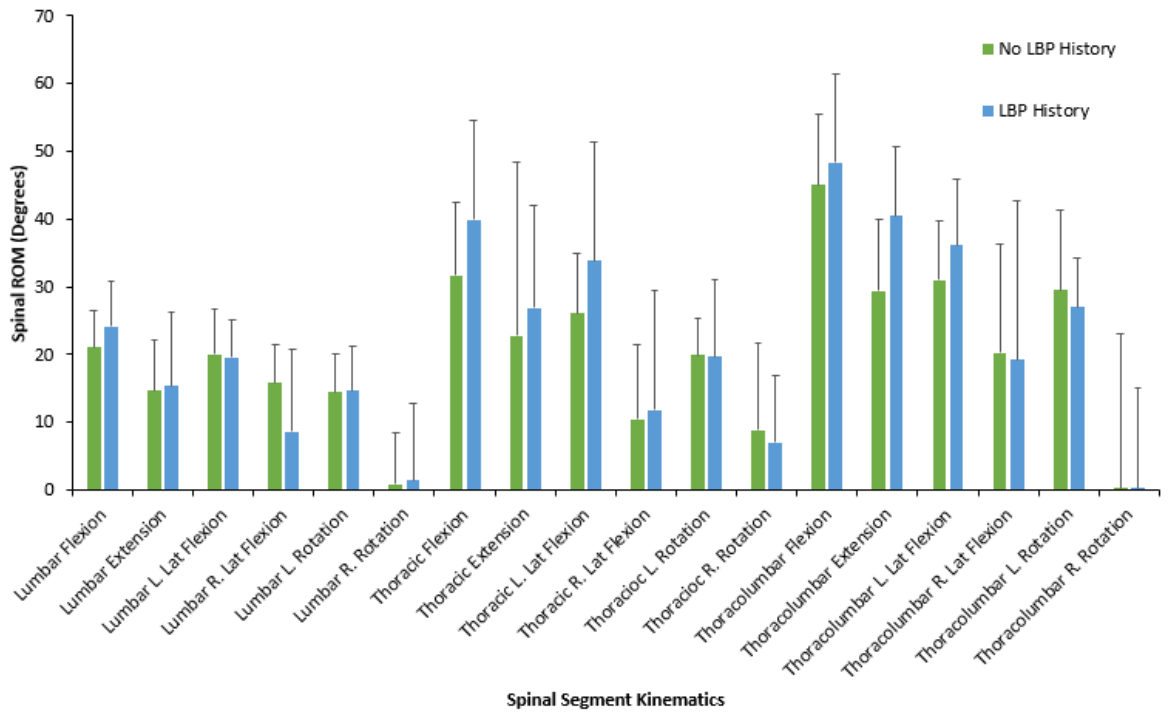
Figure 5.2.2. A comparison of three-dimensional spinal kinematics during fast bowling in senior bowlers with (n=8) and without (n=6) a history of lower back pain.
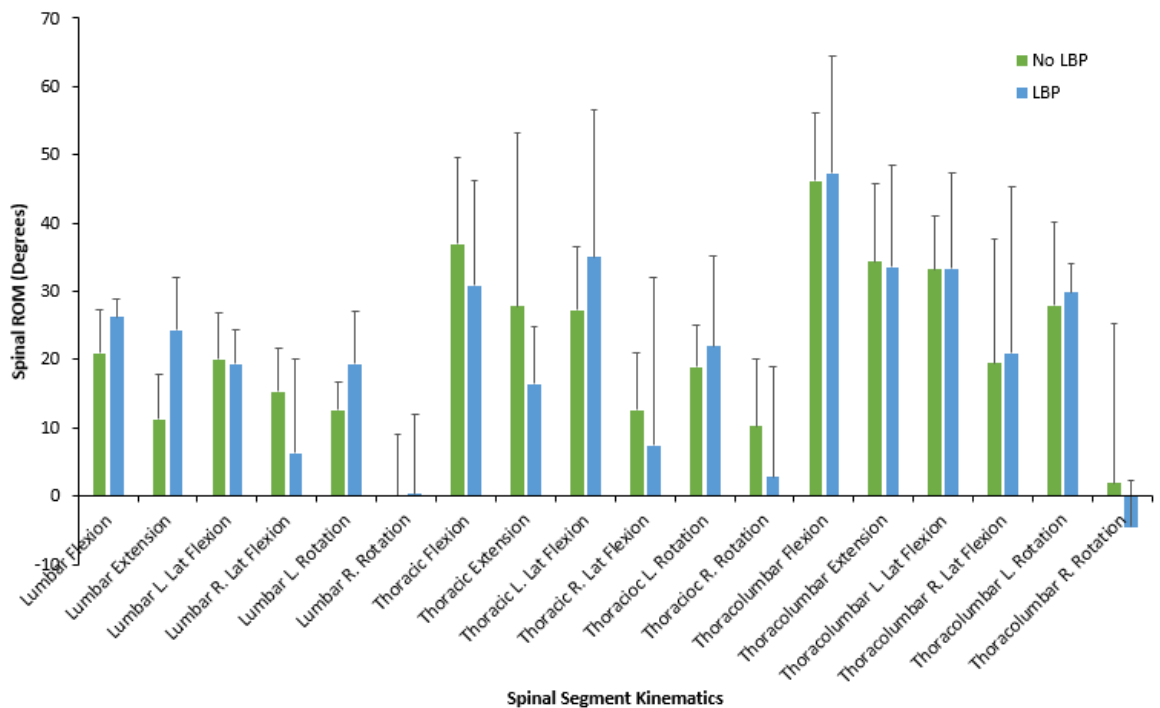
Figure 5.2.3.A comparison of three-dimensional spinal kinematics during fast bowling in senior bowlers with (n=4) and without (n=10) prospective lower back pain.

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Table 5.2.7. Summary of differences in impact characteristics during bowling in junior and senior fast bowlers with and without retrospective or prospective lower back pain.

Tibial Acceleration	Juniors		Seniors		Seniors		Common Themes
	LBP History	No LBP History	LBP History	No LBP History	Prospective LBP	No Prospective LBP	
Peak Tibial Acc x (g)						↑ at BFI ↑ at FFI	
Peak Tibial Acc y (g)						↑ at FFI	
Peak Tibial Acc z (g)		↑ at BFI	↑ at FFI	↑ at BFI		↑ at BFI ↑ at FFI	↑ at BFI = ↓ Risk
Resultant Tibial Acc (g)		↑ at BFI		↑ at BFI		↑ at BFI ↑ at FFI	↑ at BFI = ↓ Risk
Time-to-peak Tibial Acc x (ms)	↑ at BFI		↑ at BFI		↑ at FFI	↑ at BFI	
Time-to-peak Resultant Tibial Acc (ms)				↑ at FFI		↑ at BFI	
Mean Tibial Loading Rate x (g.s^{-1})				↑ at FFI		↑ at FFI	
Mean Resultant Tibial Loading Rate (g.s^{-1})						↑ at FFI	
Normalised Peak Tibial Acc x (g.kg^{-1})						↑ at BFI ↑ at FFI	
Normalised Resultant Tibial Acc (g.kg^{-1})		↑ at BFI		↑ at BFI		↑ at BFI ↑ at FFI	↑ at BFI = ↓ Risk
Sacral Acceleration							
Peak Vertical Sacral Acc (g)						↑ at FFI	
Peak Mediolateral Sacral Acc (g)		↑ at BFI ↑ at FFI				↑ at FFI	↑ at BFI = ↓ Risk
Peak Anterior-Posterior Acc (g)						↑ at FFI	
Resultant Sacral Acc (g)						↑ at FFI	
Time-to-peak Vertical Sacral Acc (ms)					↑ at BFI		
Time-to-peak Resultant Sacral Acc (ms)					↑ at BFI		
Mean Sacral Vertical Loading Rate (g.s^{-1})		↑ at BFI				↑ at BFI	
Normalised Peak Vertical Sacral Acc (g.kg^{-1})						↑ at FFI	
Resultant Attenuation (%)			↑ at BFI			↑ at FFI	

LBP; Lower back pain, BFI; Back-foot impact, FFI; Front-foot impact, ↑; larger than other group ($d \geq 0.8$), ↑; larger than other group ($d \geq 0.5$).

Table 5.2.8. Summary of differences in bowling spinal kinematics of junior and senior fast bowlers with and without retrospective or prospective lower back pain.

Variable	Juniors		Seniors		Seniors		Common Themes
	LBP History	No LBP History	LBP History	No LBP History	Prospective LBP	No Prospective LBP	
Shoulder counter-rotation					↑		
Hip-shoulder separation					↑		
T1 orientation							
S1 orientation		↑ side on at BFI					
Lumbar flexion			↑ Flex at FFI		↑ Ext at BFI ↑ Flex at FFI ↑ Range		**↑ Flex at FFI = ↑ Risk**
Lumbar lateral flexion		↑ at FFI ↑ Range	↑ at BFI			↑ at BFI ↑ Range	**↑ at BFI and Range = ↓ Risk**
Lumbar rotation		↑ at FFI ↑ Range			↑ at FFI ↑ Range		
Thoracic flexion			↑ Flex at FFI ↑ Range			↑ Ext at BFI	
Thoracic lateral flexion			↑ at FFI		↑ at FFI		**↑ at FFI = ↑ Risk**
Thoracic rotation		↑ at BFI ↑ Range				↑ at BFI	**↑ at BFI = ↓ Risk**
Thoracolumbar flexion			↑ Ext at BFI ↑ Range				
Thoracolumbar lateral flexion			↑ at FFI				
Thoracolumbar rotation							

LBP; Lower back pain, BFI; Back-foot impact, FFI; Front-foot impact, Ext; Extension, Flex; Flexion. ↑, larger than other group ($d \geq 0.8$); ↑, larger than other group ($d \geq 0.5$).

5.2.4 Discussion

This study aimed to assess tibial and sacral accelerations and three-dimensional spinal kinematics in junior and senior fast bowlers and their relationship to history of LBP and subsequent development of LBP in the follow up season. It is the first of its kind to look at both a retrospective and prospective analysis of LBP in relation to fast bowling impacts and spinal kinematics in cricket. Consequently, this study offers novel insight into the relationship of these variables with LBP risk.

5.2.4.1 Prevalence and incidence of lower back pain

Research has placed prevalence of fast bowling LBP between 40-64% (Elliott et al. 1992; Hardcastle et al. 1992; Burnett et al. 1996; Dennis et al. 2005; Kountouris et al. 2012). Findings for senior fast bowlers in this study were in line with this at 57%; however, prevalence of LBP in junior fast bowlers was slightly lower at 38%. Whilst, this may have been due to a multitude of factors, one important consideration may be difficulties with retrospective recall of LBP. The injury history questionnaires in this study relied on the player's recall of previous occurrences and as such may have omitted minor episodes, which may have been captured, in other studies, using alternate surveillance methods, such as diary or Medical/Physiotherapy records. This may to some extent explain why retrospective prevalence of LBP in senior bowlers was higher, as all senior bowlers were professional, player accounts were supplemented with club medical records. These results are in line with previous studies and support the idea that cricket fast bowling seems to be associated with a high prevalence of historical back pain.

Prospective injury surveillance highlighted a low incidence of LBP in the follow up 2015 season for junior fast bowlers at 5%. As this only accounted for one occurrence of LBP in this sample, no calculation of risk ratios or further statistical analysis was possible. These findings suggest that a previous history of LBP in junior bowlers is not a guarantee of back pain in the coming season. It is not clear whether adjustments had been made to bowling style, frequency or whether greater physical development contributed to this, but this message is positive for junior bowlers with LBP. The incidence of LBP for senior bowlers in the follow up season was higher at 29%, yet still lower than previously reported figures of between 40-64% (Elliott et al. 1992; Hardcastle et al. 1992; Burnett et al. 1996; Dennis et al. 2005; Kountouris et al. 2012). It

must be noted that all instances of LBP in senior bowlers were reported in individuals with known spinal pathology. It is possible therefore that these pathological changes become 'flared up' by the bowling season, however this pain only interrupts part of the bowling season. To this end the individuals seem able to bowl for some of the season with the presence of pathological changes further strengthening the case to record pain information rather than just pathological/radiographic diagnoses. As bowling workload has been a well reported risk factor for lower back injury and pain, medical practitioners and coaches are more closely monitoring players' workloads in the professional environment (Dennis et al. 2005). This may explain the lower incidence of LBP reported for senior bowlers in this study. Whilst junior bowlers' workloads are significantly lower, immature structures in the spine and weaker trunk musculature have been shown to increase their risk of developing spinal pathologies such as stress fractures (Crewe et al. 2013; Morton et al. 2013). However, due to the nature of not being fully professional junior bowlers are not as closely monitored, therefore incidences of LBP may go unchecked and unrecorded with the possibility of these developing into significant pathological changes (Crewe et al. 2012).

This presentation of both LBP history and prospective LBP surveillance has not been previously reported in current fast bowling literature. As no bowlers without a history of LBP developed pain in the 2015 season, risk ratios based on pain alone were not able to be produced. However, analysis of risk of LBP in the 2015 season follow-up based on previously diagnosed spinal pathology produced a risk ratio highlighting that senior bowlers with a previous lower back pathological diagnosis were four times more likely to have future episodes of LBP. Thus, this study recommends that bowlers with a history of spinal pathology as well as those that may have had an undiagnosed spinal pathology receive additional injury surveillance as an 'at risk' group.

5.2.4.2 Fast bowling impact characteristics in junior fast bowlers with and without lower back pain

Previous research has not been able to identify any variables relating to GRF which are related to risk of developing LBP. Despite this, such a link continues to be hypothesised and this study offers new insights into the relationship as it investigated a retrospective and prospective link to LBP, as well as considering FFI and BFI during live bowling.

A comparison of impact characteristics between junior bowlers with and without a history of LBP produced no significant differences due to the cautious nature of the Bonferroni correction applied. Consequently, effect sizes were also calculated to assess the practical differences between the sub-groups. Peak mediolateral sacral acceleration at FFI displayed a large effect size with the 'no LBP' group displaying greater values than the 'LBP' group'. Whilst two studies have previously reported mediolateral forces at the lumbar spine (none have reported as segment accelerations), no previous research has reported its association with LBP or injury. This finding may suggest bowlers without a history of LBP may adopt a more 'side-on' pelvis position at FFI. As the player approaches the bowling crease a switch to position the pelvis 'side-on' to the wicket would result in the anterior translation of the player's mass to 'fall' across the 'side-on' positioned pelvis. Therefore, it is possible that this anterior translation (recorded as mediolateral acceleration when side-on), may reduce the magnitude of vertical acceleration by translating momentum in the direction of delivery instead of downwards. Alternately, it could be explained by those bowlers with a history of LBP having modified their bowling technique in response to the LBP experienced. It has been speculated that pain can result in altered movements in back pain sufferers (Williams et al. 2010) with higher order movement such as velocity and acceleration specifically affected (Shum et al., 2005; Williams et al., 2013). However, if this were the case a more global reduction in acceleration might be expected, something not supported by the current study. With a moderate effect size for mediolateral sacral acceleration at BFI and a difference in the same direction as FFI, it may suggest that bowlers without LBP history stay more side-on throughout the whole delivery stride, avoiding a 'mixed' action which has been reported to increase risk of lower back injury (Elliott et al. 1993; Portus et al. 2004; Morton et al. 2013).

A number of larger impacts were noted at BFI in the 'no LBP' group. This may support previous conclusions that increased risk of LBP it is not a result of individual high magnitudes of force but a multitude of factors including the body's ability to deal with high magnitudes of force and the amount of repetition of this loading (Dennis et al. 2005; Orchard et al. 2006). In addition, this may suggest a strategy of increasing impacts at BFI, in an attempt to reduce the larger impacts experience at FFI. This is a previously unreported hypothesis and thus further investigation into its effects on injury

risk and performance is needed. However, this may potentially offer a quick and easy instruction for coaches to reduce front-foot impacts.

5.2.4.3 Fast bowling spinal kinematics in junior bowlers with and without lower back pain

This study's retrospective analysis of LBP risk in junior fast bowlers has highlighted that bowlers without a history of LBP displayed greater right thoracic rotation (away from the direction of delivery) at BFI compared to bowlers with a history of LBP. It is clear that right spinal rotation is a key mechanism of 'wind-up' to generate bowling pace, however which region of the spine this right rotation is generated from will have a significant impact of injury risk. The thoracic spine is anatomically designed for rotation motion and therefore the thoracic right rotation observed in these junior bowlers seems to mitigate risk of reporting a history of LBP. In contrast if spinal rotation (critical to bowling pace) were achieved through right lumbar rotation then significant impaction would occur through the zygopophyseal joints and ultimately stress the pars on the left perhaps explaining the significant injury risk to the left lower lumbar spine. The data collected in this study seems to suggest that enhancing right rotation through the thoracic (potentially removing the need for right lumbar rotation) may be critical in reducing the risk of LBP reporting.

Whilst previous studies have reported overall range of motion, this study suggests that, as the group without LBP history displayed larger range of motion, larger range of motion may either not be considered a risk factor or the timing and location of where spinal range is high may be a more appropriate indicator (Burnett et al. 1998; Ranson et al. 2008; Ferdinands et al. 2009; Ranson et al. 2009; Stuelcken et al. 2010; Crewe et al. 2013; Bayne et al. 2016). Furthermore, it may be suggested that what may be considered 'excessive' range of motion is dependent on the individual. Consequently, this study completed an analysis of spinal range of motion during the delivery stride relative to static range of motion trials.

Large effect sizes highlighted that thoracolumbar flexion and extension relative to static range of motion was larger in junior bowlers without a history of lower back pain. As previously suggested, uniplanar kinematics have not been highlighted as increasing risk of LBP and this relative increase in flexion and extension may ultimately reduce the requirement of concurrent lateral flexion and/or rotation (as seen in fast bowling).

5.2.4.4 Fast bowling impact characteristics in senior fast bowlers with and without lower back pain

Large effect sizes were determined in impact characteristics of senior bowlers demonstrating similar results to those in the junior bowlers. A large effect size was observed at BFI for peak tibial acceleration along the z axis with peak values greater in the 'no LBP' group. This may suggest the same strategy seen in junior bowlers: adopting higher braking forces at BFI. In support of this, peak tibial acceleration along the z axis at FFI is higher in the LBP group (d= 0.51). In addition, a large effect size was observed for time-to-peak resultant acceleration at FFI where those with no history of LBP had significantly greater time-to-peak values. This means that individuals with no history of LBP used greater breaking accelerations at BFI and allowed more time for the FFI resultant accelerations to be attenuated. It is not clear whether these mechanisms are sufficient enough to explain the presence of back pain history or are indeed the result of subtle adjustments in response to pain but this is the first time findings relating to BFI and FFI have been reported.

Comparison of senior bowlers that did and did not develop LBP in the 2015 season follow up produced no significantly different results, but some large effect sizes were observed. At BFI time-to-peak resultant tibial acceleration was faster in the 'LBP' group (d= 1.34) however time-to-peak vertical sacral acceleration was slower in the 'LBP' group (d= 1.73). This suggests bowlers in the 'LBP' group experience more rapid accelerations at the tibia but ultimately employ a strategy to decrease the vertical acceleration rate at the sacrum. This has resulted in a large effect size for vertical sacral loading rate with the 'no LBP' group displaying higher values. The fact that the 'LBP' group have decreased time-to-peak acceleration and loading rate at the sacrum may be out of necessity to decrease stress on pain sensitive structures. Although impacts in the 'LBP' group are lower they may be comparatively more damaging (as suggested in the previous study in this chapter) as a result of physiological differences that may pre-dispose the 'LBP' group to increased risk. An example of this may be previously injured tissues which have a lower tolerance to load (Bahr and Bahr, 1997). Similar passive regulation has been observed with front knee technique; bowlers experiencing the highest GRF at FFI displayed increased knee flexion (Worthington et al. 2013).

A number of tibial and sacral accelerations displayed high effect sizes at FFI (seen in table 3.4.3). The general trend of these differences suggest that the 'no LBP' group experience higher peak accelerations and faster time-to-peak accelerations and therefore larger loading rates. This suggests that it is not the magnitude or faster rate of loading in isolation that may increase bowlers' risk of injury and is likely a result of other factors. Previous research has suggested a combination of large impacts and certain bowling kinematics put bowlers at 'increased risk', however no previous research has successfully reported significant findings relating to this hypothesis (Glazier, 2010).

5.2.4.5 Fast bowling spinal kinematics in senior bowlers with and without lower back pain

The analysis of senior bowlers with and without a history of LBP in this study has highlighted that a different relationship exists between LBP history and three-dimensional spinal kinematics compared with junior bowlers in this study. Senior bowlers with a history of LBP displayed larger thoracolumbar extension at BFI compared with bowlers without LBP history. Extension of the lumbar spine has been identified as a critical element in the development of spondylolysis and disc lesions (Adams et al. 1988) particularly when the movements are highly repetitive (Green et al. 1994). Furthermore, the addition of other movements whilst the spine is in the extended position is likely to add significant further load to the spine. Available range of lateral bending and rotation in a spine which is in extension is reduced and thus smaller magnitudes of lateral bending and rotation are likely to have a relatively larger effect on the resultant loading (Burnett et al 2008). However, is should be remembered that this was at BFI where the bowler has a tendency to laterally flex to the right thus reducing the load on the left posterior elements of the spine.

Senior bowlers without a history of pain displayed greater lumbar right lateral flexion. Total range of lumbar lateral flexion (right to left) during the delivery stride is critical to developing bowling pace, again through the process of wind-up and follow through. Greater range of lumbar right lateral flexion requires relatively less left lumbar lateral flexion to achieve the same overall total range of lateral flexion. Left lumbar lateral flexion results in significant impaction of the left zygopophyseal joint and ultimately stresses the pars thus increasing the risk of develop back injury and/or pain. Indeed, this has been proposed as a significant mechanism previously (Ranson et al. 2008). The data

collected in this study seems to suggest that enhancing right lumbar lateral flexion (potentially removing the need for large amounts of left lumbar lateral flexion) may be critical in reducing the risk of LBP reporting.

As with junior bowlers, kinematics were also expressed relative to individual's static range of motion: Lumbar flexion, thoracic extension and right lateral flexion were larger in bowlers without a history of LBP. Even though range of flexion/extension was reported as larger when kinematics were expressed in degrees, the opposite relationship is seen when reported in relation to static ROM. This suggests that whilst overall range was greater in the 'LBP' group, static range was lower in the 'no LBP' group. This in isolation, without concurrent lateral flexion or rotation, may not increase LBP risk (Glazier et al. 2010). However, thoracic left and right rotation, thoracolumbar left and right rotation and lateral flexion were larger in the 'LBP' group, suggesting that although flexion/ extension may place comparatively less stress on vertebrae of bowlers in this group, the larger concurrent lateral flexion and rotation alongside this is likely to place the vertebrae in a position of mechanical weakness (Swaminathan et al. 2016).

A comparison of senior bowlers that did or did not go on to develop LBP in the 2015 season was also carried out in this study. Hip-shoulder separation was larger in bowlers that developed LBP compared to those who did not. Whilst hip-shoulder separation has been developed as a more accurate representation of spinal rotation to shoulder counter-rotation, it has not previously been highlighted as a risk factor for LBP or injury in previous literature, and has not been highlighted in the retrospective analysis in this study (Burnett et al. 1995; Portus et al. 2004). It is possible that hip-shoulder separation values are a contributing factor when coupled with other spinal kinematics, as hypothesised in previous studies (Burnett et al. 2008; Glazier et al. 2010). The previous chapter identified that at most 30% of the HSS value was explained by thoracolumbar lateral flexion therefore a large amount of the HSS would incorporate other kinematics. Greater lumbar extension at BFI and lumbar flexion at FFI, illustrates a greater range of motion overall in the LBP group. Such a demand has been identified as being particularly damaging to the pars (Green et al. 1994) and may offer an important coaching target. This study has found that lateral flexion away from the direction of delivery at BFI is greater in the bowlers that did not develop LBP which further strengthens the arguments made previously. When these kinematics were analysed in relation to static range of motion differences were observed in range of spinal flexion

with lumbar and thoracolumbar flexion larger in bowlers that did not develop LBP. Furthermore, lumbar and thoracic right rotation was higher in bowlers that did not develop LBP. In contrast, thoracic flexion was higher in the LBP group, suggesting that the two groups relied on different segments to produce spinal flexion when compared to their static range of motion. These findings mirror those seen in the comparison of junior and senior bowlers in the previous study. Previous research has highlighted that LBP may alter spinal kinematics, thus it may be the case that bowlers who developed LBP (who also tended to have LBP history) may have altered kinematics based on previous pain, which as a result has increased risk of future pain.

5.2.4.6 Kinematic predictors of LBP

Results of this study suggest that, for junior bowlers, encouraging increased thoracic rotation at BFI may decrease the risk of LBP in junior bowlers. Bowlers that did not have a history of LBP displayed more than 10° extra thoracic rotation (16° compared with 4° in the LBP group). This may result in less lumbar rotation and place less stress on the commonly injured lumbar spine.

Recommendations for senior bowlers may be focussed on reducing excessive spinal extension at BFI and spinal flexion and rotation at FFI. Bowlers that developed LBP displayed 24° extension compared with 11° for those who did not develop LBP. This decrease may be attained by encouraging increased lateral flexion at BFI, which was higher in the no LBP group. Bowlers without a history of LBP or that didn't go on to develop LBP displayed mean lumbar lateral flexion of 16° compared with the LBP group displaying between 6-9° lateral flexion. This may result in less left lumbar lateral flexion and reduce the stress placed on the highly injured left lumbar spine.

5.2.5 Conclusion

This study has presented novel data relating to impact characteristics, spinal kinematics and risk of LBP in junior and senior bowlers. This is also the first study to the author's knowledge that has analysed LBP both retrospectively and prospectively in relation to tibial and sacral impact characteristics and spinal kinematics. Junior bowlers without a history of LBP demonstrated high effect sizes showing higher tibial impacts at BFI and lower tibial impacts at FFI, suggesting that BFI may be used to dissipate some of the acceleration experienced during bowling prior to FFI. Results consistently showed bowlers that did not have a history of LBP or do not develop LBP in the 2015 season

reported larger magnitudes of acceleration at BFI. Whilst this is a novel finding, its meaning remains speculative and therefore further work should investigate the viability of this mechanism in relation to LBP risk and performance.

Furthermore, this study proposes a more 'side-on' position at BFI may result in increased mediolateral acceleration (in the direction of delivery in a side-on action) and as a result decrease downwards momentum, lowering vertical impact magnitude. At FFI, peak tibial accelerations were similar in both groups for junior fast bowlers, but this was not the case in senior fast bowlers with some larger effect size demonstrating higher magnitudes in the 'no LBP' groups. This suggests that junior and senior bowlers display a different relationship with fast bowling impacts and LBP, and thus, must be investigated as such.

This study also highlights that the relationship between LBP and fast bowling spinal kinematics is different in junior and senior bowlers and thus must be considered as separate issues. Furthermore, overall range between BFI and FFI may not be an appropriate for assessment of LBP risk. This study has highlighted that it is likely to be a combination of high ranges at specific time points during the delivery stride that may place the spine in a position of increased weakness and thus these orientations may be more representative of LBP risk. General trends highlighted that larger rotation away from the direction of delivery at back-foot impact may be advantageous for increasing overall range of motion (to increase ball release speed), whilst reducing large amounts of rotation at FFI where impact forces are higher. However, higher ranges of spinal extension at BFI were seen in LBP groups, suggesting an upright, but not extended spinal orientation is desirable at BFI. Comparisons of spinal kinematics relative to static range of motion corroborate that concurrent movements at points of high spinal loads may increase LBP risk more than greater ranges of uniplanar movements. Whilst it is desirable that interventions can be put into place before the development of LBP, it remains that the best predictor of future LBP is a previous history of lower back pathology.

5.3 Summary of Results and Key Findings

This chapter aimed to investigate the relationship between fast bowling impacts and three-dimensional spinal kinematics on risk of LBP in junior and senior bowlers.

Whilst large impacts have been hypothesised to increase risk of fast bowling lower back pain (LBP) and junior bowlers being highlighted as having further increased risk, this chapter presents the first study to directly compare junior and senior fast bowling impacts. Senior bowlers displayed greater y axis acceleration and time-to-peak acceleration at back-foot impact (BFI), which may be a result of a more front-on lower limb orientation. Whilst high effect sizes were noted for peak x axis tibial acceleration at front-foot impact (FFI) and peak mediolateral and vertical sacral acceleration at BFI, the lack of significant differences between junior and senior bowlers may suggest similar loads experienced by junior and senior bowlers. With immature structures in the spine and weaker supporting musculature, impacts experienced by junior bowlers may place more stress on the spine compared with senior bowlers experiencing similar loads.

A comparison of lumbar, thoracic and thoracolumbar spinal kinematics during bowling in junior and senior fast bowlers produced no significant differences, suggesting bowling technique may remain reasonably consistent throughout maturation and thus spinal kinematics in isolation may not be responsible for junior bowlers' higher risk of lower back pain. Analysis of effect sizes highlighted that junior bowlers displayed greater rotation away from the direction of delivery at back-foot impact, possibly as a mechanism to generate more pace on the ball. Whether similar bowling kinematics have similar effects on junior and senior LBP risk is unknown, thus separate analyses of the relationship between junior and senior bowling spinal kinematics and risk of LBP was conducted.

This chapter's analysis of the relationship between fast bowling impacts and both retrospective and prospective LBP highlights that junior and senior bowlers display different relationships with pain. Junior bowlers without a history of back pain adopt greater tibial impacts at the BFI phase which subsequently serves to reduce resultant tibial impact in the FFI phase. This mechanism suggests that BFI could be utilised as a method of dissipating resultant acceleration experienced during fast bowling and may be a key coaching intervention to reduce FFI in juniors.

In seniors, BFI was again an area of interest between those with a history of and those without a history of LBP. Bowlers without a history of LBP used greater z-axis impact at BFI suggesting the leg was positioned more side-on to the direction of travel as it landed. This could be a braking mechanism similar to that seen in junior bowlers, with the specific difference in the axis of acceleration. In addition, the resultant tibial time-to-peak acceleration at FFI was faster in those with a history of LBP suggesting reduced damping at this bowling phase. Therefore, coaching senior bowlers to increase their BFI (towards the direction of travel side-on) may positively affect the follow up FFI or indeed interventions to enhance damping could be sought; such as bowling with a 'soft front knee' or specifically designed footwear to facilitate FFI damping.

Whilst relationships between LBP and spinal kinematics differed slightly between junior and senior bowlers, general trends highlighted greater thoracic rotation at back-foot impact may be favourable as a mechanism of wind-up without increasing range of motion at front-foot impact, which may place the spine in a position of increased weakness when larger loads are experienced. In addition greater right lumbar lateral flexion at BFI may reduce the requirement for large amounts of left lumbar lateral flexion at FFI, a position know to load the spine signifiacntly and proposed to be linked to back pain in bowlers. Furthermore, greater spinal extension at back-foot impact was observed in the LBP group, this may be a result of extension limiting range of rotation and lateral flexion, increasing the probability of loading vertebrae at end range, increasing compression on facet joints and thus increasing risk of LBP. LBP groups also displayed greater flexion at front-foot impact, which may produced similar restrictions to lateral flexion and rotation, when even higher load is placed on the spine. Consequently, this study proposes bowlers utilise greater thoracic right rotation or greater lumbar right lateral flexion and avoid spinal extension at back-foot impact. In addition bowlers should aim to limit spinal flexion throughout the delivery stride to minimise their risk of low back pain.

The prospective analysis identified that senior bowlers were more likely to develop back pain if they had lower z-axis and resultant tibial acceleration at BFI suggesting an absence of the mechanism speculated above. At FFI the findings suggest that individuals who go on to develop LBP actually have lower impact magnitudes for a number of variables, including accelerations at the sacrum. Indeed, acceleration magnitude and rate all suggest that lower acceleration values at this point of the

bowling cycle may hold the key to identifying who is likely to develop LBP in the follow up season. Despite this it is clear that currently, the best predictor of future LBP is previous history of spinal pathology in senior fast bowlers.

Whilst the use of inertial sensing technology has produced some novel findings, the inherant limitations of this technology must also be acknowledged. Whilst, the lack of line-of-sight has enabled greater portability and flexibility in testing environment, this also means that key points in the bowling action, such as ball release, cannot be indetified with inertial sensors in isolation. Bayne et al. (2016) highlights that the largest force experienced in the lumbar spine is after FFI. If FFI was defined in a similar fashion to previous studies this would insinuate studies in this thesis have missed 'key points' in the bowling action. However, as FFI was defined as peak sacral acceleration this study is able to define kinematics and impacts when the lower back is theoretically at greatest risk of damage. This does however mean the time between BFI and FFI as defined in this thesis may be longer than in previous studies, however all kinematics between these points were able to be analysed and as such is an acceptable limitation. Furthermore, the fact the same sample was used to compare both juniors and seniors (study 5.1) and LBP in juniors and seniors (study 5.2), may raise the argument that previous history of LBP may affect variables in the sample in study 5.1 where both 'LBP history' and 'no LBP history' were in the same group. The fact that no significant differences were observed between 'LBP' and 'no LBP' groups in study 5.2 highlights that it is acceptable to place both in the same group for study 5.1.

Whilst this chapter has highlighted some novel findings in regards to the relationship between junior and senior fast bowling and the risk of LBP, the only outcome measure to which fast bowing biomechanics have been compared against is LBP incidence. Before the uptake of the recommendations outlined in this chapter, an analysis of the effect of the variables measured in this chapter on fast bowling performance must be analysed. This will allow for a clear understanding of how recommendations can be 'tailored' to provide a balance between decreasing risk of LBP and maximising performance.

Summary of key findings

- Similar magnitudes of tibial and sacral impacts were observed in junior and senior bowlers, with senior bowlers positioned more side-on.
- Bowlers without LBP history or did not go on to develop LBP reported larger peak impacts at BFI than LBP groups (although not statistically significant).
- Impact characteristics in isolation may not be an appropriate predictor of LBP risk.
- History of lower back pathology resulted in four times greater risk of developing future lower back pain.
- Fast bowling spinal kinematics remain similar between junior and senior bowlers.
- Junior bowlers spinal range of motion during static trials was only significantly different for thoracolumbar rotation but no differences were observed during bowling.
- Thoracic rotation away from the direction of delivery at back-foot impact may reduce risk of LBP.
- Lumbar lateral flexion away from the direction of delviery at back-foot impact may reduce the risk of LBP.
- Large ranges of spinal extension at back-foot impact or flexion at front-foot impact may increase risk of LBP.

Chapter 6

Fast Bowling Performance

6.1 The relationship between ball release speed and spinal kinematics and tibial and sacral accelerations during fast bowling. A pilot study.

6.1.1 Introduction

The previous studies in this thesis have produced novel conclusions concerning spinal kinematics and tibial and sacral impacts and their relationship with risk of lower back pain (LBP) in fast bowlers. Greater lumbar lateral flexion and thoracic rotation at back-foot impact (BFI) as well as greater magnitudes of impacts at BFI is proposed to decrease risk of LBP, whilst increased spinal extension at this point may increase risk. Whilst these conclusions may be useful to practitioners when looking at technique interventions or coaching, uptake is likely to be limited without the knowledge of the effects of these recommendations on fast bowling performance.

The previous chapter has hypothesised that lower limb orientation at back-foot impact may affect magnitude and direction of impacts experienced at the lower limb during fast bowling. However, due to no video analysis being utilised in the previous studies in this thesis, this theory was not able to be verified. Bartlett et al. (1996) highlights two common techniques seen at back-foot impact, describing the orientation of the foot and lower limb as either 'front-on' (facing the direction of delivery) or 'side-on' facing perpendicular to the direction of delivery (as seen in figure 6.1.1). As such, it may be hypothesised that a 'front-on' lower limb orientation would enable greater knee flexion and ankle dorsiflexion, allowing impact force to be dissipated more effectively through greater range of motion. Whilst previous research has highlighted the effect of front leg knee flexion on ball release speed, no previous research has investigated the effect of the back leg (Worthington et al. 2013).

Aside from front leg technique, previous literature have highlighted a number of key variables that may contribute to increased performance (Salter et al. 2007; Worthington et al. 2013). Worthington et al. (2013) highlight increased upper trunk flexion at front-foot impact (FFI) and delayed arm circumduction as key variables for increasing ball release speed. Faster run-up speed has also been shown to increase ball release speed; however, 'excessive' speeds have been reported to have a detrimental effect on accuracy due to a 'rushed' delivery stride (Bartlett et al. 1996; Salter et al. 2007). Whilst previous literature provides some understanding as to factors that may affect bowling performance, the method of analysis used to produce the findings in this thesis utilised

different technology and as such produced some novel metrics. Therefore, the relationship between these metrics (such as tibial accelerometry) and ball release speed are unknown. Consequently, an analysis of the relationship between these variables and fast bowling performance is warranted.
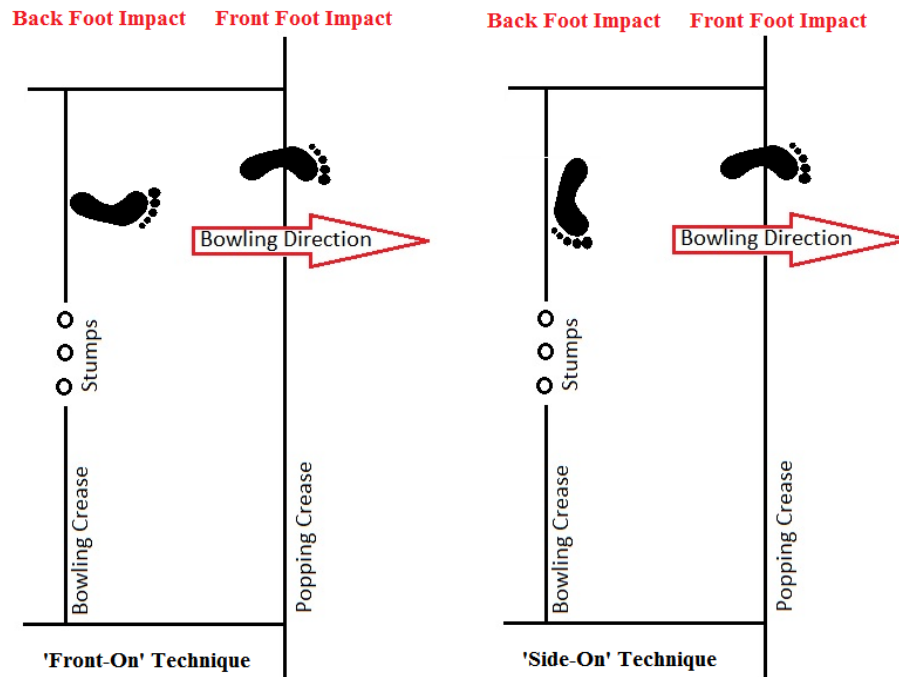
Figure 6.1.1. Orientations of the lower limb at back and front-foot impact.

6.1.2 Aim of the study

This study aims to explore how spinal kinematics, sacral and tibial acceleration and lower limb orientation at back foot impact may affect ball release speed during cricket fast bowling.

6.1.3 Results

Of the thirteen participants analysed in this study, eight (62%) bowlers displayed 'front-on' techniques, three (23%) 'side-on' and two (15%) mixed.

6.1.3.1 Ball release speed, impact characteristics and spinal kinematics

Mean (±SD) ball release speed was recorded at 27.4 (±2.7)m/s. Mean tibial and sacral accelerations and spinal kinematics are reported in table 6.1.1 and 6.1.2. Sacral vertical loading rate at BFI displayed a significant positive, moderate correlation with ball release speed (r=.521, p=.041). Thoracic lateral flexion in the direction of delivery at

FFI also displayed a significant positive, moderate correlation with ball release speed (r=.629 ,p=.014).

Table 6.1.1. Mean (±SD) tibial and sacral accelerations during back and front foot impact and their relationship with ball release speed (n=13).

	Back Foot Impact	r	*P*	Front Foot Impact	r	*P*
Tibial Acceleration						
Peak Tibial Acc X (g)	11.65 (3.68)	-.097	.382	25.41 (10.30)	.193	.264
Peak Tibial Acc Y (g)	14.42 (6.23)	.245	.222	18.99 (12.31)	.363	.112
Peak Tibial Acc Z (g)	3.17 (2.27)	-.081	.401	17.33 (9.42)	-.044	.443
Resultant Tibial Acc (g)	20.52 (5.02)	.186	.281	40.15 (18.68)	.210	.245
Time to Peak Tibial Acc X (ms)	47.32 (21.22)	.276	.193	27.68 (14.94)	.145	.318
Time to Peak Resultant Tibial Acc (ms)	29.23 (9.91)	.209	.257	21.76 (13.23)	.013	.483
Mean Tibial Loading Rate X (g.s^{-1})	361.97 (203.43)	-.300	.172	1156.16 (592.63)	.095	.379
Mean Resultant Tibial Loading Rate (g.s^{-1})	819.07 (234.75)	-.197	.270	2271.40 (1365.37)	.216	.240
Sacral Acceleration						
Peak Vertical Sacral Acc (g)	1.77 (0.56)	.308	.165	2.67 (1.11)	-.246	.209
Peak Mediolateral Sacral Acc (g)	1.27 (0.21)	.131	.342	0.90 (0.48)	-.003	.496
Peak Anterior-Posterior Acc (g)	0.29 (0.21)	-.295	.176	0.21 (0.16)	.324	.140
Resultant Sacral Acc (g)	2.57 (0.39)	.099	.380	3.23 (1.08)	-.285	.173
Time to Peak Vertical Sacral Acc (ms)	85.03 (44.50)	.122	.353	110.43 (55.23)	.290	.168
Time to Peak Resultant Sacral Acc (ms)	84.65 (44.59)	.118	.357	45.92 (13.27)	-.140	.324
Mean Sacral Vertical Loading Rate (g.s^{-1})	39.66 (24.10)	.521	.041*	41.38 (22.98)	-.261	.195
Mean Sacral Resultant Loading Rate (g.s^{-1})	60.36 (31.29)	.433	.080	79.67 (30.26)	-.204	.251
Resultant Attenuation (%)	86.31 (12.69)	-.037	.454	86.29 (6.32)	.383	.099

Acc, acceleration; g, gravity; ms, milliseconds; s, seconds; kg, kilograms; *, p<.05.

Table 6.1.2. Mean (±SD) spinal kinematics during back and front foot impact and their relationship with ball release speed (n=13).

	Back Foot Impact			Front Foot Impact		
	Spinal Kinematics (°)	r	p	Spinal Kinematics (°)	r	p
Shoulder counter-rotation	21.42 (8.17)	.135	.338			
Hip-shoulder separation	33.41 (30.05)	.093	.387			
T1 orientation	260.85 (10.07)	.467	.063			
S1 orientation	259.59 (14.25)	.050	.438			
Lumbar flexion	-15.01 (14.43)	-.025	.469	22.78 (18.30)	-.160	.310
Lumbar lateral flexion	-18.00 (19.47)	.390	.105	11.08 (20.45)	-.195	.271
Lumbar rotation	14.27 (15.41)	-.074	.410	21.84 (29.28)	-.241	.225
Thoracic flexion	-9.97 (14.48)	-.470	.062	21.39 (20.66)	-.114	.362
Thoracic lateral flexion	0.58 (22.84)	-.078	.405	17.70 (21.68)	.629	.014*
Thoracic rotation	-12.30 (10.26)	-.029	.465	22.68 (24.85)	.029	.465
Thoracolumbar flexion	-30.37 (13.95)	-.162	.308	35.74 (14.61)	-.432	.080
Thoracolumbar lateral flexion	-7.93 (26.51)	.034	.458	18.06 (20.97)	-.076	.407
Thoracolumbar rotation	-2.06 (12.44)	.332	.146	22.14 (11.16)	-.266	.201

°, degrees; *, P< .05.

6.1.3.2 Effect of back leg orientation on ball release speed, impact characteristics and spinal kinematics

Bowlers with a 'front-on' lower limb orientation displayed significantly faster ball release speeds than bowlers with a 'side-on' lower limb orientation at BFI at 28.3m/s and 24.41m/s respectively (p = 0.038).

An analysis of tibial and sacral accelerations at BFI and FFI highlighted no significant differences between bowlers with a 'front-on' and 'side-on' lower limb orientation at BFI. However, a large effect size was observed, highlighting faster resultant tibial loading rate at BFI in bowlers with a side-on lower limb (*d*=0.99). Additionally, a large effect size reported faster time-to-peak tibial acceleration along the longitudinal axis of the tibia at FFI in bowlers with a 'side-on' back leg at BFI (d=0.81).

No significant differences were observed in three-dimensional spinal kinematics between 'front-on' and 'side-on' BFI lower limb orientation groups, however large

effect sizes were observed. Larger thoracic rotation away from the direction of delivery was seen in bowlers with a 'front-on' lower limb orientation ($d=1.11$). Greater lumbar and thoracolumbar flexion at FFI was observed in 'side-on' lower limb bowlers ($d=0.93$ and 1.17). Additionally, total range of thoracolumbar flexion was greater in the 'side-on' technique ($d=1.12$). Conversely, larger thoracic lateral flexion at FFI was seen in 'front-on' lower limb bowlers ($d=1.20$).

Table 6.1.3. Comparison of fast bowling tibial and sacral accelerations between bowlers with front-on (n=6) and side-on (n=7) lower limb orientation at back-foot impact.

	Back-Foot Impact				Front-Foot Impact			
	Back-Foot Front-On	Back-Foot Side-On	p-value	Effect size (d)	Back-Foot Front-On	Back-Foot Side-On	p-value	Effect size (d)
Tibial Acceleration								
Peak Tibial Acc x (g)	12.44 (3.19)	11.08 (4.14)	.553	0.37	26.20 (11.93)	24.73 (9.61)	.810	0.14
Peak Tibial Acc y (g)	3.34 (2.70)	3.05 (2.13)	.839	0.12	16.64 (9.80)	17.92 (9.82)	.819	-0.13
Peak Tibial Acc z (g)	16.47 (8.41)	12.96 (4.25)	.360	0.53	15.89 (8.92)	21.65 (14.79)	.424	-0.47
Resultant Tibial Acc (g)	20.80 (6.95)	20.36 (3.71)	.881	0.08	38.99 (19.08)	41.15 (19.79)	.846	-0.11
Time-to-peak Tibial Acc x (ms)	44.97 (18.32)	49.00 (24.38)	.762	-0.19	34.09 (20.01)	22.19 (6.03)	.160	0.81
Time-to-peak Resultant Tibial Acc (ms)	33.47 (8.42)	26.20 (10.34)	.226	0.77	27.20 (18.66)	17.10 (2.23)	.180	0.76
Mean Tibial Loading Rate x (g.s^{-1})	369.57 (235.18)	356.54 (197.27)	.919	0.06	1156.02 (837.11)	1156.28 (344.18)	.999	<-0.01
Mean Resultant Tibial Loading Rate (g.s^{-1})	696.63 (172.64)	906.53 (244.42)	.132	-0.99	2023.94 (1469.69)	2483.50 (1347.12)	.568	-0.33
Sacral Acceleration								
Peak Vertical Sacral Acc (g)	1.89 (0.47)	1.69 (0.64)	.593	0.36	2.64 (1.43)	2.64 (0.87)	.997	0
Peak Mediolateral Sacral Acc (g)	1.21 (0.30)	1.32 (0.12)	.394	-0.48	0.93 (0.48)	0.87 (0.51)	.811	0.12
Peak Anterior-Posterior Acc (g)	0.23 (0.22)	0.33 (0.21)	.410	-0.46	0.24 (0.16)	0.19 (0.17)	.580	0.30
Resultant Sacral Acc (g)	2.54 (0.32)	2.60 (0.46)	.813	-0.15	3.18 (1.25)	3.28 (1.00)	.879	-0.09
Time-to-peak Vertical Sacral Acc (ms)	100.12 (68.30)	74.25 (13.92)	.333	0.52	111.74 (46.03)	109.31 (65.82)	.941	0.04
Mean Sacral Vertical Loading Rate (g.s^{-1})	47.36 (28.55)	34.17 (20.90)	.375	0.53	40.40 (18.39)	42.21 (27.79)	.894	-0.08
Mean Sacral Resultant Loading Rate (g.s^{-1})	69.65 (42.79)	53.73 (21.24)	.411	0.47	80.86 (30.65)	78.65 (32.34)	.902	0.07
Vertical Attenuation (%)	16.91 (6.60)	20.56 (16.12)	.645	-0.30	12.66 (10.49)	12.75 (5.88)	.984	-0.01
Resultant Attenuation (%)	13.99 (6.54)	13.49 (3.36)	.864	0.10	9.91 (6.25)	9.23 (3.93)	.816	0.13

Table 6.1.4 Comparison of fast bowling spinal kinematics between bowlers with front-on (n=6) and side-on (n=7) lower limb orientation at back-foot impact.

Spinal Kinematics (°)	Back-Foot Impact				Front-Foot Impact				Range			
	Back-foot Front-on	Back-foot Side-on	P value	Effect Size (d)	Back-foot Front-on	Back-foot Side-on	P value	Effect Size (d)	Back-foot Front-on	Back-foot Side-on	P value	Effect Size (d)
Shoulder counter-rotation	25.45 (8.81)	19.40 (7.60)	.244	0.74								
Hip-shoulder separation	34.85 (9.83)	32.68 (37.14)	.913	0.08								
T1 orientation	261.34 (2.42)	260.85 (12.52)	.912	0.05								
S1 orientation	261.96 (14.45)	258.40 (15.00)	.704	0.24								
Lumbar flexion	-12.44 (6.01)	-16.30 (17.50)	.684	0.30	12.38 (15.20)	27.99 (18.29)	.174	-0.93	24.82 (16.66)	44.28 (34.93)	.324	-0.71
Lumbar lateral flexion	-11.08 (16.24)	-21.47 (21.01)	.409	0.55	10.37 (31.27)	11.43 (15.41)	.937	-0.04	21.45 (36.23)	32.90 (32.61)	.592	-0.33
Lumbar rotation	12.35 (12.94)	15.23 (17.27)	.777	-0.19	11.03 (27.23)	27.24 (30.48)	.391	-0.56	-1.32 (15.87)	12.02 (18.45)	.247	-0.78
Thoracic flexion	-11.75 (19.56)	-9.97 (14.48)	.779	-0.10	25.79 (28.81)	19.19 (17.27)	.625	0.28	37.54 (15.34)	28.28 (23.27)	.492	0.47
Thoracic lateral flexion	7.18 (11.58)	-2.72 (26.92)	.506	0.48	31.76 (8.97)	10.67 (23.12)	.115	1.20	24.58 (17.26)	13.39 (46.57)	.658	0.32
Thoracic rotation	-19.50 (10.82)	-8.70 (8.42)	.084	-1.11	15.79 (18.45)	26.13 (27.99)	.523	-0.44	35.29 (15.52)	34.82 (32.59)	.979	0.02
Thoracolumbar flexion	-25.56 (16.49)	-32.78 (13.01)	.423	0.49	25.95 (10.56)	40.63 (14.34)	.102	-1.17	51.50 (13.61)	73.41 (24.14)	.127	-1.12
Thoracolumbar lateral flexion	-0.30 (23.55)	-11.74 (28.57)	.507	0.44	19.89 (14.18)	17.15 (24.53)	.843	0.14	20.18 (23.08)	28.90 (26.18)	.586	-0.35
Thoracolumbar rotation	-5.84 (5.55)	-0.17 (14.76)	.483	-0.54	22.29 (4.74)	22.06 (13.63)	.976	0.02	28.13 (10.07)	22.23 (17.85)	.559	0.41

6.1.4 Discussion

This study aimed to use novel methods of analysis of tibial and sacral accelerations as well as three-dimensional spinal kinematics to investigate the relationship between ball release speed and fast bowling technique. Whilst previous studies have extensively investigated this relationship, the novel method of analysis in this thesis has highlighted previously unreported variables (Portus et al. 2004; Salter et al. 2007; Worthington et al. 2013). As such, the addition of this knowledge may provide coaches and practitioners with valuable data when using inertial sensors in practice. Furthermore, it may help to inform the previously collected data in this thesis as to whether the suggested interventions to lower risk of lower back pain simultaneously affect performance.

6.1.4.1 Ball release speed, impact characteristics and spinal kinematics

As is common within the fast bowling literature, this study used ball release speed as a measure of fast bowling performance (Salter et al. 2007; Worthington et al. 2013). Studies have reported average ball release speeds of between 24.4-37.4m/s (Stockhill and Bartlett, 1992; Elliott et al. 1993; Crewe et al. 2013). This study reports mean ball release speed in line with the lower end of this range at 27.4±2.7m/s, this may be a result of this study using club level bowlers, where most of the previous cited speeds are analysing elite level fast bowlers. Furthermore, whilst some studies have analysed ball release speed using two-dimensional measures (as in this study), three-dimensional analysis of ball release may elicit faster release speeds due to the absence of perspective error. However, due to the relatively small mediolateral deviation of the ball expected at ball release, this effect is likely to be minimal.

Previous studies have highlighted a number of key variables that affect ball release speed; faster approach speed has been reported to significantly increase ball release speed (r= .543-.728) (Glazier et al. 2000; Salter et al. 2007). However, studies have also reported that excessive approach speed may affect bowling accuracy due to a 'rushed' delivery stride (Bartlett et al. 1996). Thus, a faster ball release speed may in some instances result in decreased accuracy. This may challenge the most commonly utilised method of performance analysis within the literature of ball release speed quantification without a metric of bowling accuracy. This study only included deliveries of a 'good' line and length and thus any trends observed to have increased ball release speed are unlikely to compromise accuracy.

Additionally, faster shoulder angular velocity of the delivery arm has been shown to significantly correlate to faster ball release speeds in a number of studies (r= .358-.688)(Glazier et al. 2000; Salter et al. 2007; Worthington et al. 2013). Salter et al. (2007) also highlighted vertical velocity of the non-bowling arm as a significant factor (r=.801). The fact that this current study did not measure approach speed or arm velocities may be considered to limit a holistic description of contributing factors; however, the main aim of this study was to investigate the contributing factors to ball release speed, relative to the guidelines previously highlighted in this thesis as potentially decreasing risk of LBP. This may help to provide an understanding of whether interventions could potentially decrease LBP risk without affecting performance. The variables highlighted above have been well reported previously and it may be argued that additional insight into this relationship is not needed. In conjunction with this, reporting of these variables will not address the aim of this study as run-up speed or arm velocities were not reported as LBP risk factors in this thesis. However, previous literature has reported some relationships with variables that have been measured in this study.

An extended front knee at FFI has been reported to elicit higher ball release speeds (r=.520-.720) (Portus et al. 2004; Loram et al. 2005; Wormgoor et al. 2010; Worthington et al. 2013). The relationship of front knee technique with ground reaction force (GRF) has been well documented (Bartlett et al. 1996; Worthington et al. 2013). Typically, studies report higher GRF in bowlers with an extended or 'braced' front knee at FFI (Portus et al. 2004). However, Worthington et al. (2013) challenges this relationship reporting higher GRF with a 'flexed knee' technique as a protective mechanism to avoid excessive loading of the lower limb. Whilst GRF was not reported in this study, tibial accelerations were measured, which may be indicative of magnitudes of GRF due to the high correlation between the two metrics (Tran et al. 2010). No significant relationships were reported between tibial accelerations and ball release speed in this study. This supports the conclusions from Worthington and colleagues (2013) stating that higher GRF does not necessarily elicit higher ball release speed. Furthermore, no variable relating to tibial or sacral acceleration at FFI reported a significant relationship with ball release speed in this study. Only sacral vertical loading rate at BFI highlighted a significant relationship (r=.521, p=.041).

The finding that vertical sacral loading rate at BFI displays a significant positive correlation with ball release speed is a novel one, with few studies reporting impact characteristics at BFI in relation to performance. Crewe et al. (2013) is the only study to report loading further up the body (rather than just GRF) and this only highlights lumbar load between FFI and ball release. The fact that loading rate and not peak values showed a significant relationship, highlights it may not be magnitude of force, but rate at which force is loaded that may contribute to faster deliveries. The same trend was not seen in back foot tibial accelerations. Thus, while impacts may not differ at the tibia, faster bowlers are able to tolerate or produce higher loading rates at the sacrum (potentially as a result of a 'stiffer' kinetic chain) and therefore transfer more momentum from the run-up into the delivery stride (Bartlett et al. 1996). Furthermore, physical factors may also explain this trend; weaker bowlers may not be able to tolerate similarly high loading rates, consequently having to regulate it through lower limb mechanics. This theory aligns with that of Worthington et al. (2013) stating that a 'flexed front knee' may be a result of high GRF, rather than technique dictating the magnitude of force. This is also seen in passive regulatory mechanisms is barefoot running (Franklin et al. 2015). Previous studies in this thesis has outlined that increasing back tibial acceleration at BFI may decrease risk of LBP. Thus, alongside findings from this study, it may be suggested that if an increase in tibial acceleration at BFI can be achieved without compromising loading at the sacrum this intervention may be viable. However, if larger magnitudes of back tibial acceleration at BFI cannot be tolerated (decreasing 'stiffness' as a result) it may result in a braking effect that could compromise ball release speed.

Alongside impact variables, a number of kinematic parameters measured by this study have been reported as displaying relationships with ball release speed in previous literature. Shoulder rotation has been described as playing a key role in faster ball release speeds; delayed shoulder rotation in the direction of delivery, resulting in larger hip-shoulder separation and smaller rotation angles at FFI have been correlated to faster ball release speeds. Spinal rotation is the only kinematic variable that has been correlated to ball release speed in previous literature. Consequently, the fact that this study reports a moderate, positive correlation between thoracic lateral flexion at FFI and ball release speed is novel. This may be explained by differences in bowling technique between cohorts; 85% of bowlers in this study were 'front-on' or 'side-on' bowlers

whereas previous literature commonly reports a greater percentage of 'mixed' bowling actions. Thus, similar hip and shoulder alignments may allow increased lateral flexion due to less concurrent rotation at these time points. Movement through a greater range is likely to allow more opportunity to generate force and thus increase ball release speed. In contrast, the difference in hip and shoulder alignment in the mixed action is likely to limit concurrent lateral flexion and as such is more reliant or shoulder rotation to generate pace on the ball, as is reported in previous literature (Glazier, 2010).

Whilst the variables reported in this study give an insight into fast bowling technique and the relationship with ball release speed, it must be acknowledged that thoracic lateral flexion and BFI sacral loading rate described in this study account for 40% and 27% of variance seen in ball release speed respectively (r^2= 0.396 and 0.271). Thus, a large proportion of variance is unaccounted for by this analysis. Biomechanically, this is likely to consist of approach speed, lower limb and shoulder kinematics as highlighted by previous work (Portus et al. 2004; Glazier et al. 2010; Worthington et al. 2013). Aside from biomechanical variables, it must be acknowledged that physical factors will likely result in variance in ball release speed between bowlers. As this study was not able to physiologically profile bowlers, this information was unknown and was therefore not able to be factored into the analysis. For this reason, junior bowlers were excluded from this study due to the greater risk of differences in maturation and physical factors, increasing the potential to confound the above analysis. Whilst these variables may add to the description of performance, they would not provide information on whether the recommendations made previously in this thesis are feasible as interventions to decrease LBP risk without compromising performance. Thus, a more focussed approach was taken in this analysis.

6.1.4.2 Effect of back leg orientation on ball release speed, impact characteristics and spinal kinematics

As highlighted in the above discussion, lower limb technique has been investigated within the current body of literature. However, this investigation has been focussed around lower limb technique at FFI (Portus et al. 2004; Worthington et al. 2013). An analysis of lower limb technique at BFI has not been reported previously, despite the fact differences in lower limb orientation at this point have been noted (Bartlett et al. 1996). Therefore, this study investigated the effect of lower limb orientation on ball

release speed, impact characteristics at the tibia and sacrum and spinal kinematics. Bartlett et al. (1996) has categorised lower limb techniques into front-on (foot facing direction of delivery) and side-on (foot facing perpendicular to the direction of delivery) at BFI. As such, this study has grouped bowlers according to these guidelines.

Ball release speed in bowlers bowling with a front-on lower limb orientation was significantly faster than in the side-on group at 28.3m/s and 24.41m/s respectively. As such, further analysis into why this is the case is warranted. Comparison of impact characteristics and spinal kinematics between the front-on and side-on groups displayed no significant differences but a number of large effect sizes were observed. A large effect size was reported for resultant tibial loading rate at BFI (d=0.99), with the side-on lower limb group demonstrating faster loading rates. It may be suggested that a side-on orientation limits knee flexion and as such a 'stiffer' lower limb and higher loading rates at BFI are displayed. Nonetheless, in order to verify this suggestion knee flexion angle is needed. This was not able to be collected in this study due to the use of two-dimensional video analysis, and as such the variances in the orientation of the lower limb did not allow the accurate calculation of knee flexion angle without potential out of plane errors. Alongside the above finding, a large effect size was observed for time-to-peak tibial acceleration along the longitudinal axis of the tibia at FFI (d=0.81). This finding describes a faster time-to-peak in bowlers with a side-on technique. However, as no other variables showed any differences at FFI, there is no evidence to suggest this is a result of lower limb orientation at BFI, and is therefore likely to be a result of individual technique isolated to FFI. Nonetheless, it is interesting to note, with a significantly faster ball release speed in the front-on group there were still no differences in magnitudes of impacts at FFI as highlighted in previous studies.

Large effect sizes were also observed in the comparison of spinal kinematics between the two groups. Larger thoracic rotation away from the direction of delivery was seen in the front-on group (d=1.11). This rotation is likely a 'wind-up' mechanism in order to generate pace on the ball as seen with increased shoulder counter-rotation in junior bowlers (Elliott et al. 2005). Whilst the same counter-rotation may be utilised in the side-on group, the side-on orientation of the lower limb would enable easier pelvis counter-rotation and as such decrease the need for resultant spinal rotation. This hypothesis may offer a balance between performance and risk of injury and pain; producing similar momentum without utilising excessive range of motion at the spine.

However, this may only be effective if similar patterns of decreased spinal range of motion are seen at FFI where impacts are highest.

Large effect sizes were also observed in spinal kinematics at FFI. Lumbar and thoracolumbar flexion at FFI and total thoracolumbar flexion range were greater in the side-on group (d=0.93, 1.17 and 1.12 respectively). Conversely, thoracic lateral flexion was greater in the front-on group (d=1.20). These differences support the hypothesis of different kinematic strategies to generate pace highlighted previously in this thesis. Additionally, the fact that the front-on bowlers displayed greater ball release speed as well as larger lateral flexion aligns with the correlation analysis in this study highlighting larger lateral flexion elicits faster ball release speed.

6.1.5 Conclusion

The findings in this study highlighting positive correlations between BFI sacral loading rate and thoracic lateral flexion and ball release speed are novel findings. These findings suggest that this thesis' previous recommendations to decrease risk of LBP by decreasing spinal extension and increasing impact magnitude at BFI may not compromise ball release speed and may therefore be viable interventions. The analysis of differences in back limb orientation reported bowlers with a 'front-on' lower limb orientation at BFI bowled faster that the 'side-on' bowlers. This is likely a result of the 'front-on' group displaying larger thoracic lateral flexion (which is reported to positively correlate to ball release speed in this study), however there is little evidence to suggest a link between back limb orientation and spinal kinematics at FFI. As such, the faster ball release speed seen in the 'front-on' group is likely a result of strength and technical factors unrelated to back leg orientation. Consequently, it may be recommended that bowlers use whichever back leg orientation feels most comfortable. Further investigation should focus on the implementation of these recommendations and whether they are effective in practice.

Chapter 7

General Discussion

This thesis aimed to determine which factors during fast bowling may affect risk of lower back pain (LBP) in junior and senior fast bowlers. This was achieved by using new methods of analysing impacts and spinal kinematics, not previously utilised during live fast bowling. Analyses using these methods along with retrospective and prospective LBP data were able to highlight impact and kinematic factors during the fast bowling delivery stride that may increase risk of LBP.

LBP and lower back injury have been highlighted as significant issues in the fast bowling population, resulting in substantial time loss and expense within the professional game (Orchard et al. 2002; Financial Times, 2014; ECB, 2016). Research has highlighted that junior fast bowlers are at increased risk of LBP and injury as a result of immature structures in their spine (Johnson et al. 2012; Crewe et al. 2013; Morton et al. 2013). Furthermore, increased risk of LBP has been reported in bowlers with excessive bowling workloads (Dennis et al. 2005; Orchard et al. 2006). Whilst research has given a clearer picture of 'non-technique based' risk factors, LBP's relationship with fast bowling impacts and kinematics during the delivery stride remains unclear. A review of available literature highlighted that this may be a result of alternate aims, common methodological flaws and heterogeneity between studies.

Whilst sample sizes in this and previous literature are small, the professional fast bowling population is small and thus samples used in this thesis are typical of those in previous studies (Portus et al. 2000; Elliott et al. 2005; Ranson et al. 2009; Stuelcken et al. 2010; Crewe et al. 2013; Worthington et al. 2013; Bayne et al. 2016; Middleton et al. 2016). The exploratory nature of this research resulted in the analysis on large numbers of variables and as a result, large sample sizes were needed in order to achieve adequately powered studies. Consequently, effect sizes were used in order to highlight trends where no statistical significance was present. Whilst previous studies suffer from similarly low sample sizes, few have chosen to report effect sizes and thus, few, if any, differences, have been reported in the literature in relation to LBP and injury (Morton et al. 2013).

Studies investigating ground reaction force (GRF) during fast bowling have hypothesised that large impacts may increase risk of LBP or injury, however no significant relationships have been reported to support this hypothesis (Hurrion et al. 2000; Portus et al. 2004; Stuelcken and Sinclair, 2009; Crewe, et al. 2013; Worthington

et al. 2013; Spratford and Hicks, 2014; Bayne et al. 2016; King et al. 2016; Middleton et al. 2016). A more holistic analysis of the relationship between LBP and impacts, including both retrospective and prospective analyses of junior and senior fast bowlers, aimed to explore this relationship further.

Studies have reported that shoulder counter-rotation in excess of 30 degrees increases risk of LBP and injury (Portus et al. 2004; Stuelcken et al. 2010). However, shoulder counter-rotation can only describe the orientation of the shoulders in the transverse plane, which explains very little in relation to three-dimensional spinal kinematics. Thus, as with fast bowling impacts, the exact mechanisms of injury in relation to kinematic variables remain unclear. Consequently, this thesis aimed to address and overcome the environmental limitations seen in force plate and kinematic studies through the use of inertial sensors to measure 'live' fast bowling.

Previous research has validated the use of inertial sensors for the measurement of impacts and spinal kinematics, however this technology has not been utilised for the analysis of fast bowling, thus a reliability and validity analysis was warranted (van den Noort et al. 2009; Crowell et al. 2010; Theobald et al. 2010; Tran et al. 2010; Charry et al. 2011; Hu et al. 2014; Sell et al. 2014). Comparisons of inertial sensor data with gold standard devices (force plate and optoelectronic motion analysis systems) produced correlations similar to those reported in previous literature for tasks such as running and jumping (Najafi et al. 2015; Raper et al. 2018). Correlations were strong to very strong ($r > 0.8$) for 79% of all variables measured (tibial accelerations and lumbar kinematics) with root mean square error of prediction values of $0.3\text{-}1.5°$ for lumbar kinematics. Thus, it may be assumed that the inertial sensors used in this study are a valid measure of impacts and spinal kinematics during fast bowling.

Additionally, a reliability analysis was conducted. Intraclass correlation coefficients were shown to be good to excellent for repeated measurement of both fast bowling tibial and sacral impacts and spinal kinematics. Furthermore, standard error of measurement was $<3g$ for impacts and $<9°$ for spinal kinematics; values comparable or lower than previous studies using accelerometers for impact measurement in landing tasks and fast bowling spinal kinematics using optoelectronic motion analysis or electromagnetic sensors which have reported values between $1\text{-}17°$ (Ranson et al. 2008; Ranson et al. 2009). Furthermore, the inertial sensors used in this study have

demonstrated good sensitivity to true changes in bowling technique with minimum detectable change (MDC) values placed at <9g for fast bowling impacts and 13° for lumbar kinematics respectively. This indicates true differences in impacts or spinal kinematics and not just natural variations in technique. These values may be used as thresholds for such analyses as whether kinematics may have been altered as a result of pain or whether an intervention has had a real effect. As MDCs for fast bowling analysis have not previously been reported in the literature, this novel information may be a valuable and practical measure for coaches.

Previous research in other sports such as running have highlighted that different surfaces may result in different impact characteristics and therefore affect risk of injury (Hardin et al. 2004). However, as no previous research has analysed this hypothesis in cricket fast bowling, its effects and relationship with risk of injury were unknown. This study highlighted that different magnitudes of tibial accelerations were evident on different playing surfaces, although a 'harder' surface did not always elicit a larger magnitude or faster time to peak as may be expected. This suggests bowlers may employ an intrinsic, regulatory mechanism to cope with higher impacts. This hypothesis is backed by previous fast bowling literature suggesting higher ground reaction force may result in a more flexed front knee in order to dissipate exposure to high impact forces (Worthington et al. 2013). Furthermore, this intrinsic regulation has also been reported in running literature (Hardin et al. 2004). Results suggested that bowling on outdoor artificial surfaces may increase magnitudes of impacts when compared with other surfaces. Conversely, indoor wooden 'sports hall' surfaces resulted in the lowest impacts and may be desirable for high bowling workloads or 'return to play' scenarios.

Whilst high impact forces have only been hypothesised to increase risk of injury, with no significant findings reported in previous literature, shoulder counter-rotation has repeatedly shown a relationship with risk of LBP and injury. Shoulder counter-rotation in excess of 30° has been reported to significantly increase risk of LBP and injury (Portus et al. 2004; Stuelcken et al. 2010). Whilst this is a simple and quick measurement for coaches, shoulder counter-rotation is only able to describe orientation of the shoulders in the transverse plane, relative to the wickets. Changes in shoulder orientation may be achieved via spinal or whole body rotation, thus the extent to which shoulder counter-rotation represents true three-dimensional spinal kinematics is unknown. This study highlighted that shoulder counter-rotation is not a representative

measure of three-dimensional spinal kinematics. However, a moderate relationship was observed between shoulder counter-rotation and thoracic and thoracolumbar lateral flexion at the start of the bowling action. This may suggest a mechanism of generating pace on the ball that consists of multiplanar kinematics (including lateral flexion and rotation) which may increase risk of injury. This mechanism of injury and LBP has been proposed in previous literature, but with no significant relationships reported, further investigation is needed to confirm this hypothesis (Glazier et al. 2010). Thus, further analyses into three-dimensional kinematics and their association with LBP were carried out.

Previous pathological research has highlighted that junior fast bowlers are at increased risk of LBP and injury compared with senior bowlers. Studies reported initial incidence of LBP or injury occurring before the age of 18 and prevalence of conditions such as disc degeneration increasing from 21% to 65% between the ages of 13-18 years (Elliott and Khangure, 2002). However, no previous studies have offered a direct comparison of junior and senior fast bowling technique and large heterogeneity in reported methodologies means that comparisons between junior and senior bowling studies are difficult. Consequently, this thesis reports novel findings that may inform practitioners and researchers of any differences between junior and senior fast bowling impacts or spinal kinematics that may predispose junior bowlers to increased risk of LBP.

Senior bowlers displayed significantly greater acceleration opposing movement in the direction of delivery (if the back-foot is orientated 'front-on') and time-to-peak acceleration at back-foot impact than juniors. As these finding were not mirrored at the sacrum, these differences may be a result of orientation of the lower limb at BFI which has been highlighted to vary between a 'front-on' or side-on' orientation between bowlers (seen in figure 5.1.1) (Bartlett et al. 1996). It may be hypothesised that a 'front-on' lower limb position would allow greater knee flexion and ankle dorsiflexion to dissipate impacts and increase time-to-peak acceleration. Whereas a 'side-on' lower limb position would have to rely on hip abduction, which with a smaller range of motion, may be less effective. However, previous studies have only looked at lower limb kinematics at front-foot impact (Portus et al. 2004; Worthington et al. 2013). This study was not able to measure lower limb kinematics, as such this is investigated further in this thesis.

Comparisons of impacts at front-foot impact (FFI) highlighted that time-to-peak resultant tibial acceleration was faster and peak acceleration perpendicular to the direction of delivery (with a 'front-on' foot orientation) was greater in junior bowlers. This may suggest that the greater y axis accelerations seen in seniors at back-foot impact (BFI) may aid in reducing impacts at FFI. Although previous studies have not reported this relationship, methods of reducing ground reaction forces (usually at FFI) have been reported (Elliott, 2000; Portus et al. 2004; Worthington et al. 2013). Greater front knee flexion and front ankle plantarflexion at contact, are some methods that have been reported to reduce impact magnitudes and time-to-peak and therefore assumed to reduce LBP and injury risk. Consequently, it may be the case that the more experienced senior bowlers use these mechanisms to more effectively dissipate front-foot impacts and increase time-to-peak. However, further analysis was needed to indicate whether the differences displayed between junior and senior bowlers may have some influence on risk of LBP.

Further kinematic comparisons of junior and senior bowlers highlighted large effect sizes but no statistically significant differences in spinal kinematics at BFI. Junior bowlers displayed greater spinal rotation away from the direction of delivery at back-foot impact. This may suggest a similar mechanism to increased shoulder counter-rotation, suggested in previous research to generate more pace on the ball (Elliott et al. 2005). The similarity between all other spinal kinematics suggests that technique remains reasonably consistent through maturation, however whether similar kinematics have similar effects on LBP risk of junior and senior bowlers needed further investigation. Consequently, this thesis conducted separate analyses of junior and senior bowlers and the effect of fast bowling impacts and spinal kinematics on LBP risk.

Fast bowling injury, and more specifically lower back injury, has received much attention in the literature, highlighting age and workload as two key risk factors in injury risk (Dennis et al. 2005; Orchard et al. 2006; Johnson et al. 2012; Crewe et al. 2013; Morton et al. 2013). Bowlers in this study (alongside coaches to help inform them) were required to fill out predicted bowling workload in terms of overs bowled per week for the follow-up season. Junior bowlers have limited bowling workloads due to the laws of the game (stated at the beginning of this thesis). These workload limitations were further enforced due to the controlled elite environment of the junior bowlers. The senior bowlers used in this study were professionals and were able to accurately

measure and predict upcoming workloads due to coach support. As such, predicted bowling workloads of 50 overs per week (the threshold for increased injury risk suggested by Dennis et al. 2005) were not exceeded in any bowlers. Whilst lower back injury has received much attention in the literature, it has been postulated that pain and injury are not always synonymous and thus must be regarded as separate issues (Millson et al. 2004). LBP has received less attention within the literature and consequently, while technique factors that affect lower back injury risk remain ambiguous, mechanisms of LBP are even more uncertain (Morton et al. 2013). This thesis conducted both retrospective and prospective analysis of LBP, analysing junior and senior fast bowlers' LBP history, as well as incidence of LBP during the 2015 season. Previous studies looking at LBP in relation to bowling technique tend to report either retrospective or prospective pain data and thus cannot provide a holistic analysis (Elliott et al. 1992; Hardcastle et al. 1992; Dennis et al. 2005; Kountouris et al. 2012).

This study reported 38% incidence of retrospective LBP in junior bowlers and 57% for senior bowlers. These values (as seen in table 7.1.1) are similar to those seen in previous studies. Junior retrospective LBP incidence was slightly lower than the next lowest reported incidence whilst senior retrospective LBP incidence was 6% higher than the mean but within a range of one standard deviation. The much lower prospective LBP incidence in both junior and senior bowlers could be a result of only tracking bowlers for the following season, thus more chronic issues highlighted in more longitudinal studies may have been missed (Dennis et al. 2005). Whilst a number of previous LBP studies have analysed some form of bowling technique, no previous research has analysed three-dimensional spinal kinematics and bowling impacts in relation to risk of LBP (Elliott et al. 1992; Hardcastle et al. 1992; Burnett et al. 1996).

Table 7.1.1. Incidence of lower back pain in fast bowlers reported in previous research and the current study.

Author	Participants	LBP
Current Study (Retrospective)	N = 21 (mean age 16.9yrs)	38%
Current Study (Retrospective)	N = 14 (mean age 24.1yrs)	57%
Current Study (Prospective)	N = 21 (mean age 16.9yrs)	5%
Current Study (Prospective)	N = 14 (mean age 24.1yrs)	29%
Kountouris et al. 2012	N=38 (mean age 15.5yrs)	45%
Dennis et al. 2005	N= 44 (mean age 14.7yrs)	52%
Burnett et al. 1996	N= 19 (mean age 13.6)	53%
Elliott et al. 1992	N=20 (mean age 17.9yrs)	40%
Hardcastle et al. 1992	N=22 (mean age 17.9yrs)	64%
Mean (SD) of previous studies		$50.8 \pm 9.1\%$

Yrs; years, SD; standard deviation; n, number; LBP, lower back pain.

Analysis of junior fast bowlers and history of LBP highlighted bowlers without a history of LBP displayed greater impacts at BFI, and consequently decreased impacts at FFI, when compared with bowlers with a history of LBP. Bowlers without a history of LBP displayed resultant acceleration of 20g at BFI, resulting in 31g at FFI, compared to the LBP history group who reported 15g at BFI and 36g at FFI. Thus, it may be proposed that a ratio of 2:3 may be a reasonable target for BFI to FFI magnitudes for bowlers looking to decrease risk of injury (as opposed to 1:2 seen in the LBP group). This relationship has not previously been reported in the literature with few studies reporting impacts at both back and front-foot impact (Mason et al. 1989; Saunders and Coleman, 1991; Elliott et al. 1992; Hurrion et al. 2000). This increased impact at BFI could be a mechanism to decrease momentum through the delivery stride and thus result in lower impacts at FFI. However, as this is the first study to report this finding, it is not known whether this mechanism may have an impact on fast bowling performance or whether momentum is conserved further up the body in order to release the ball with a high ball release speed. As such, this relationship is investigated further in this thesis.

Analysis of junior spinal kinematics and history of LBP revealed similar compensatory mechanisms at BFI. Bowlers without a history of LBP displayed greater thoracic rotation away from the direction of delivery at BFI. By increasing this rotation at BFI, it may decrease the need for lumbar rotation (the area most at risk of injury) and thus

decrease the risk of LBP and injury as highlighted by previous studies (Portus et al. 2004; Stuelcken et al. 2010). Bowlers that had no history of LBP displayed over 10° more thoracic rotation (mean of 16°) at BFI compared with the LBP group (mean of 4°). Consequently, it may be suggested that greater contralateral rotation at BFI seen in junior bowlers (reported in the juniors and seniors comparison) is not likely to be a factor in junior bowlers' increased LBP risk. It may be hypothesised that if overall range between BFI and FFI remains the same, greater contralateral rotation at BFI allows for less ipsilateral rotation at FFI and thus may place the spine in a position more able to cope with high impacts. Conversely, if range at FFI is not decreased then the increase in range of rotation will likely elicit an increase in ball release speed (Portus et al. 2004). A 'front-on' bowling technique may more easily facilitate this extra rotation as a result of less time needed to orientate the rest of the body in the pre-delivery stride, thus an adequate pre-delivery stride would enable more contralateral rotation before BFI. These cues of a 'more front-on position' with increased 'away shoulder rotation' at BFI are two simple commands that may be implemented to address these issues.

Although senior bowlers displayed a slightly different relationship with history of LBP, similar trends were observed. Greater peak tibial acceleration perpendicular to the direction of delivery (with a front-on foot position) was observed at BFI for bowlers without a history of LBP, suggesting a breaking mechanism in the direction of delivery with a side-on foot position, similar to the mechanism reported in junior bowlers. As with junior bowlers, bowlers with no LBP history displayed a BFI:FFI resultant acceleration ratio closer to 2:3 than the LBP group, reaffirming that this may be a possible predictor of LBP risk. Similar results were observed in the analysis on senior bowlers that did or did not develop LBP in the 2015 season. This supports the previously stated hypotheses that increased braking at BFI may allow more effective dissipation of impact forces. These findings have not previously been reported in the literature and may aid coaches in the design of injury prevention and rehabilitation interventions.

FFI impact did not agree with the typically assumed hypothesis that higher impacts at FFI increase the risk of LBP (Johnson et al. 2012). Impact magnitude and rate at both the tibia and sacrum were higher in the 'no prospective LBP' group. This may suggest that large impacts in isolation may not increase risk of LBP. It may be the case that lower impacts experienced when the spine is in a position of weakness may potentially

place bowlers at greater risk than larger impacts with the spine in a more mechanically advantageous position. Furthermore, all bowlers that developed LBP had a reported previous history of LBP and thus the lower impacts observed in this group may be a result of previous alterations as a result of pain (Williams et al. 2010). Nonetheless, these findings advocate the analysis of kinematics in conjunction with impact characteristics.

Kinematic analysis of senior bowlers with and without a history of LBP highlighted bowlers without a history of LBP increased lumbar lateral flexion away from the direction of delivery at BFI. This may be a similar mechanism to junior bowlers who increased contralateral rotation at BFI. This increase in lateral flexion was also observed in senior bowlers who did not develop LBP prospectively. A side-on delivery may better facilitate greater lateral flexion towards the direction of delivery as a result of this increased 'wind-up' away from the direction of delivery at BFI.

In contrast to this, increased spinal extension at BFI, as well as increased flexion at FFI were displayed in senior bowlers with a history of LBP and bowlers that developed prospective LBP. Previous studies have suggested that during increased spinal extension, range of lateral flexion and rotation is decreased (Burnett et al. 2008). Thus, it may be the case that a combination of increased range in the sagittal plane, alongside lateral flexion and rotation resulting in a greater likelihood of loading the spine at end range (a known risk factor of injury and pain) (Chosa et al. 2004; Smith et al. 2005; Swaminathan et al. 2016). Furthermore, greater extension at BFI is likely to reduce lateral flexion and rotation range and thus larger range may be required at FFI to bowl effectively, thus placing the spine in a position of increased mechanical weakness at the time when the largest impacts are experienced. Bowlers with a history of LBP displayed 41° of thoracolumbar extension at BFI compared to 29° in no LBP history bowlers. Additionally, with bowlers that developed LBP displaying greater flexion and rotation at FFI, recommendations focussing on movements predominantly along a single plane may be warranted.

Prior to the uptake of the interventions suggested above, this thesis aimed to highlight whether any of the key variables in relation to LBP risk affected fast bowling performance. Previous studies have highlighted a number of contributing factors including run-up speed, delayed shoulder rotation and a 'braced' front knee (Portus et

al. 2004; Salter et al. 2007; Worthington et al. 2013). As this thesis has used a novel method of analysis and produced novel metrics, the same method was used to allow direct comparison to previous suggestions regarding LBP risk. This study highlighted that faster ball release speeds were observed with a larger sacral loading rate at BFI (r=.521). As few studies have investigated loads at the lower back alongside performance, this is a novel finding. It may be suggested that a 'stiffer' lower limb at BFI allows for more effective transfer of momentum through to ball release, thus increasing ball release speed. Greater thoracic lateral flexion also displayed a significant positive correlation to ball release speed (r=.629). This may highlight that increased range of motion elicits faster ball release speeds, but the plane in which this occurs is likely to vary with bowling technique. This may explain why previous studies have reported rotation as a contributing factor with cohorts containing a greater percentage of 'mixed' bowling actions, where concurrent lateral flexion is likely to be limited (Glazier, 2010). This analysis also highlights that recommendations provided in this thesis to reduce risk of LBP are unlikely to decrease ball release speed, as no variables highlighted as 'decreasing risk' correlated to ball release speed, however this may vary between individuals.

This thesis also highlighted that orientation of the lower limb may affect bowling biomechanics, as well as ball release speed. These different orientations of a 'side-on' back leg and 'front-on' back leg have been previously highlighted within the literature but not explored. As such, this study compared fast bowling impacts, spinal kinematics and ball release speeds in 'front-on' and 'side-on' lower limb orientations. As no previous studies have investigated this link, significant findings obtained in this study may provide valuable insight into mechanisms of bowling relating to injury and performance. Significantly faster ball release speeds were observed in the 'front-on' orientation group as well as larger lateral flexion at FFI, this finding agrees with the correlation analysis above. Greater spinal counter-rotation was observed in the 'front-on' group also, a similar mechanism to generate pace as those observed in previous studies (Elliott et al. 2005). Lastly, a 'softer' back knee was observed in the 'front-on group' who displayed a slower tibial loading rate. This may suggest that the correlation between a faster sacral loading rate and ball release speed, is not reliant on a 'stiffer' knee and as such a 'softer' knee but 'stiffer' hip at BFI may provide a compromise between performance and injury.

In summary, whilst previous research has been clear that LBP in fast bowlers is an important issue, its relationship with bowling impacts and three-dimensional spinal kinematics has remained ambiguous (Morton et al. 2013). This thesis has highlighted that inertial sensors are a valid and reliable method for the analysis of 'in field' fast bowling impacts and three-dimensional spinal kinematics; allowing the analysis of variables (such as different playing surfaces) that are difficult to measure with previously utilised methods. Whilst studies in this thesis also suffer from small sample sizes seen in previous studies, the reporting of effect sizes have allowed new insight into junior and senior fast bowlers' relationship with LBP. While junior bowlers have been highlighted to be at increased risk of LBP, differences in spinal rotation between junior and senior bowlers was not highlighted as a variable that may increase risk of LBP when compared with LBP history and prospective back pain. This thesis proposes that larger degrees of spinal extension at BFI or flexion at FFI may increase risk of LBP when coupled with lateral flexion or rotation by decreasing the available range of motion. Conversely, larger impacts at BFI may have a protective effect against LBP by decreasing impact magnitude at FFI (which are typically larger).

Chapter 8

Conclusions and

Recommendations

8.1 Conclusion

This thesis has demonstrated that inertial sensors are a valid and reliable method of measuring fast bowling lower limb impacts, as well as three-dimensional fast bowling spinal kinematics. Validity and reliability was as good as or better than previously reported values using different methodologies. Therefore, inertial sensors may offer a method of measuring bowling impacts and kinematics in the field.

Analysis of fast bowling impacts on different playing surfaces suggest that the 'hardest' surface does not always result in the highest magnitudes of impacts or fastest time-to-peak impacts. Outdoor artificial surfaces produced the largest impacts and thus, high workloads should be avoided on these surfaces. Conversely, the wooden indoor surface recorded the lowest impacts and may therefore be advantageous in rehabilitation or high workload scenarios.

This thesis has highlighted that shoulder counter-rotation or hip-shoulder separation may not be representative of three-dimensional spinal kinematics during fast bowling. Lateral flexion away from the direction of delivery increased with larger degrees of shoulder counter-rotation. Thus, it may be the case that the coupling of these two movements may increase risk of injury or LBP.

The lack of differences observed between junior and senior fast bowling impacts and kinematics suggests that bowling technique remains relatively consistent throughout maturation. Greater rotation away from the direction of delivery at back-foot impact was observed in junior bowlers (possibly as a mechanism to increase pace). However, further analysis highlighted this is not likely to increase risk of LBP. Consequently, it may be assumed that junior bowlers increased risk of LBP cannot be attributed to differences in impacts or spinal kinematics.

Junior and senior bowlers displayed slightly different relationships with LBP, however some common trends highlighted that greater lumbar lateral flexion or thoracic rotation

away from the direction of delivery at back-foot impact may decrease risk of LBP. In addition, larger impacts at back-foot impact decreased risk of LBP, possibly as a mechanism of reducing impacts at front-foot impact. Conversely, greater spinal extension at back-foot impact or flexion at front-foot impact may increase risk of LBP.

A correlation analysis of bowling impacts and spinal kinematics with ball release speed, highlighted that increased sacral loading rate at BFI and larger thoracic lateral flexion at FFI increased with higher ball release speed. As no relationships were observed with variables highlighted in the LBP studies, the recommendations in this thesis to reduce LBP risk are not likely to negatively affect ball release speed.

Additionally, a comparison of back leg orientation revealed significantly faster ball release speeds with a front-on lower limb orientation at 28.3m/s, compared to 24.41m/s with a side-on orientation. Front-on orientations also displayed larger thoracic lateral flexion at FFI; as the front-on group also displayed faster ball release speeds this supports correlation analysis above.

8.2 Recommendations for future work

- Analyse the effect of different playing surfaces on bowling kinematics (specifically lower limb) in relation to injury and performance.
- Investigate the effect of real-time feedback on fast bowling impacts and spinal kinematics using inertial sensors.
- How effective are the LBP risk reduction interventions (longitudinal study)?
- How effectively can recommendations from this thesis be integrated into grass roots teaching of fast bowling technique?

8**.3 Final conclusion**

This thesis has provided new insights into the relationship between cricket fast bowling and lower back pain. The findings suggest that bowling impacts or spinal kinematics may not be responsible for junior bowlers increased risk compared with senior bowlers; however, specific technique variables may predispose or protect bowlers against lower back pain. Whilst recommendations to reduce LBP risk have been shown not to affect performance in this thesis, further research should aim to assess the feasibility and effectiveness of these long-term technique changes. The production of these guidelines, along with the introduction of a valid and reliable method of portable analysis, is aimed at further educating coaches to help reduce the impact of lower back pain in the fast bowling population. By providing the tools to assess lower back pain risk, change factors contributing to this increased risk and monitor future change, this may be achieved.

Appendix 1 Quality appraisal form adapted from Law et al. 1998

1. **Citation**
   - Include full title, all authors (last name, initials), full journal title, year, volume # and page #s.
   - This ensures that another person could easily retrieve the same article.

2. **Study Purpose**
   - Was the purpose clearly stated? − The purpose is usually stated briefly in the abstract of the article, and again in more detail in the introduction. It may be phrased as a research question or hypothesis.
   - A clear statement helps to determine if the topic is important, relevant, and of interest to you. Consider how the study can be applied to occupational therapy practice and/or your own situation before you continue. If it is not useful or applicable, go on to the next article.

3. **Literature**
   - Was the relevant background literature reviewed? − A review of the literature should be included in the article describing research to provide some background to the study. It should provide a synthesis of relevant information such as previous work/ research, and discussion of the clinical importance of the topic.
   - It identifies gaps in current knowledge and research about the topic of interest, and thus justifies the need for the study being reported.

4. **Design**
   - There are many different types of research designs. The most common types in rehabilitation research are included.
   - The essential features of the different types of study designs are outlined, to assist in determining which was used in the study you are reviewing.
   - Some of the advantages and disadvantages of the different types of designs are outlined to assist the reader in determining the appropriateness of the design for the study being reported.
   - Different terms are used by authors, which can be confusing - alternative terms will be identified where possible.
   - Numerous issues can be considered in determining the appropriateness of the methods/design chosen. Some of the key issues are listed in the Comments section, and will be described below. Diagrams of different designs, and examples using the topic of studying the effectiveness of activity programmes for seniors with dementia, are provided.
   - Most studies have some problems due to biases that may distort the design, execution or interpretation of the research. The most common biases are described at the end of this section.

5. **Appropriateness of Design**
   - Some of the important issues to consider in determining if the study design is the most appropriate include:

- Knowledge of the topic/issue: If little is known about an issue, a more exploratory method is appropriate, for example a case study or a cross-sectional design. As our level of knowledge increases, study designs become more rigorous, where most variables that could influence the outcome are understood and can be controlled by the researcher. The most rigorous design is the RCT.
- Outcomes: If the outcome under study is easily quantified and has well-developed standardized assessment tools available to measure it, a more rigorous design (eg. An RCT) is appropriate. If outcomes are not fully understood yet, such as quality of life, then a design that explores different factors that may be involved in the outcomes is appropriate, such as a case control design.
- Ethical issues: It is appropriate to use a research design that uses control groups of people receiving no treatment if there are no ethical issues surrounding the withholding of treatment.
- Study purpose/question: Some designs are well-suited to studying the effectiveness of treatment, including RCT's, before-after designs, and single-case studies. Other designs (eg. case control and cross sectional) are more appropriate if the purpose of the study is to learn more about an issue, or is a pilot study to determine if further treatment and research is warranted.

6. **Biases (3 points)**
   - There are many different types of biases described in the research literature. The most common ones that you should check for are described below under 3 main areas:
   1. Sample (subject selection) biases, which may result in the subjects in the sample being unrepresentative of the population which you are interested in;
   2. Measurement (detection) biases, which include issues related to how the outcome of interest was measured; and
   3. Intervention (performance) biases, which involve how the treatment itself was carried out.
   - The reader is directed to the bibliography if more detailed information is needed about biases.
   - A bias affects the results of a study in one direction - it either "favours" the treatment group or the control group. It is important to be aware of which direction a bias may influence the results.
   1. Sample/ Selection Bias
      - *Volunteer or referral bias:*
        - People who volunteer to participate in a study, or who are referred to a study by someone are often different than non-volunteers/non-referrals.
        - This bias usually, but not always, favours the treatment group, as volunteers tend to be more motivated and concerned about their health.

Seasonal bias:

o If all subjects are recruited and thus are evaluated and receive treatment at one time, the results may be influenced by the timing of the subject selection and intervention. For example, seniors tend to be healthier in the summer than the winter, so the results may be more positive if the study takes place only in the summer.

o This bias could work in either direction, depending on the time of year.

Attention bias:

o People who are evaluated as part of a study are usually aware of the purpose of the study, and as a result of the attention, give more favourable responses or perform better than people who are unaware of the study's intent. This bias is why some studies use an "attention control" group, where the people in the control group receive the same amount of attention as those people in the treatment group, although it is not the same treatment.

2. <u>Measurement/ Detection Biases</u>

Number of outcome measures used:

o If only one outcome measure is used, there can be a bias in the way that the measure itself evaluated the outcome. For example, one ADL measure considers dressing, eating, and toiletting but does not include personal hygiene and grooming or meal preparation.

o This bias can influence the results in either direction; eg. it can favour the control group if important elements of the outcome that would have responded to the treatment were missed.

o Bias can also be introduced if there are too many outcome measures for the sample size. This is an issue involving statistics, which usually favours the control group because the large number of statistical calculations reduces the ability to find a significant difference between the treatment and control groups.

Lack of "masked" or "independent" evaluation:

o If the evaluators are aware of which group a subject was allocated to, or which treatment a person received, it is possible for the evaluator to influence the results by giving the person, or group of people, a more or less favourable evaluation. It is usually the treatment group that is avoured. This should be considered when the evaluator is part of the research or treatment team.

Recall or memory bias:

o This can be a problem if outcomes are measured using self-report tools, surveys or interviews that are requiring the person to recall past events. Often a person recalls fond or positive memories more than negative ones, and this can favour the results of the study for those people being questioned about an issue or receiving treatment.

3. <u>Intervention/ Performance Biases</u>

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Contamination:
- o This occurs when members of the control group inadvertently receive treatment, thus the difference in outcomes between the two groups may be reduced. This favours the control group.

Co-intervention:
- o If clients receive another form of treatment at the same time as the study treatment, this can influence the results in either direction. For example, taking medication while receiving or not treatment could favour the results for people in either group. The reader must consider if the other, or additional, treatment could have a positive or negative influence on the results.

Timing of intervention:
- o Different issues related to the timing of intervention can introduce a bias.
- o If treatment is provided over an extended period of time to children, maturation alone could be a factor in improvements seen.
- o If treatment is very short in duration, there may not have been sufficient time for a noticeable effect in the outcomes of interest. This would favour the control group.

Site of treatment:
- o Where treatment takes place can influence the results - for example, if a treatment programme is carried out in a person's home, this may result in a higher level of satisfaction that favours the treatment group. The site of treatment should be consistent among all groups.

Different therapists:
- o If different therapists are involved in providing the treatment(s) under study to the different groups of clients, the results could be influenced in one direction - for example, one therapist could be more motivating or positive than another, and hence the group that she worked with could demonstrate more favourable outcomes. Therapist involvement should be equal and consistent between all treatment groups.

7. **Sample**
- N = ? The number of subjects/clients involved in the study should be clear.
- Was the sample described in detail? The description of the sample should be detailed enough for you to have a clear picture of who was involved.
- Important characteristics related to the topic of interest should be reported, in order for you to conclude that the study population is similar to your own and that bias was minimized. Important characteristics include:
  - o who makes up the sample - are the subjects appropriate for the study question and described in terms of age, gender, duration of a disability/disease and functional status (if applicable)?;

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- how many subjects were involved, and if there are different groups, were the groups relatively equal in size?;
  - how the sampling was done - was it voluntary, by referral? Were inclusion and exclusion criteria described?
  - if there was more than 1 group, was there similarity between the groups on important (confounding) factors.
- Was the sample size justified? The authors should state how they arrived at the sample size, to justify why the number was chosen. Often, justification is based on the population available for study. Some authors provide statistical justification for the sample size, but this is rare.
- Ethics procedures should be described, although they are often left out. At the very least, authors should report if informed consent was obtained at the beginning of the study.

8. **Outcomes**
   - Outcomes are the variables or issues of interest to the researcher - they represent the product or results of the treatment or exposure.
   - Outcomes need to be clearly described in order for you to determine if they were relevant and useful to your situation. Furthermore, the method (the how) of outcome measurement should be described sufficiently for you to be confident that it was conducted in an objective and unbiased manner.
   - Determine the frequency of outcome measurement. It is important to note if outcomes were measured pre- and post-treatment, and whether short-term and/or long-term effects were considered.
   - Review the outcome measures to determine how they are relevant to occupational therapy practice, ie. - they include areas of occupational performance, performance components and/or environmental components.
   - List the measures used and any important information about them for your future reference. Consider if they are well-known measures, or ones developed by the researchers for the specific study being reported. It may be more difficult to replicate the study in the latter situation.
   - The authors should report if the outcome measures used had sound (well-established and tested) psychometric properties - most importantly, reliability and validity. This ensures confidence in the measurement of the outcomes of interest.
   - Were the outcome measures reliable? - Reliability refers to whether a measure is giving the same information over different situations. The 2 most common forms of reliability are: test-retest reliability - the same observer gets the same information on two occasions separated by a short time interval; and inter-rater reliability - different observers get the same information at the same time.
   - Were the outcome measures valid? - Asks whether the measure is assessing what it is intended to measure. Consider if the measure includes all of the relevant concepts and elements of the outcome (content validity), and if the

authors report that the measure has been tested in relationship to other measures to determine any relationship (criterion validity). For example, a "valid" ADL measure will include all relevant elements of self-care, and will have been tested with other measures of daily living activities and self-care functioning to determine that the relationship between the measures is as expected.

9. **Intervention (not included in scoring)**
   - Intervention described in detail? - there should be sufficient information about the information for you to be able to replicate it.
   - In reviewing the intervention, consider important elements such as:
     - The focus of the intervention - is it relevant to occupational therapy practice and your situation;
     - Who delivered it - was it one person or different people, were they trained?
     - How often the treatment was received - was it sufficient in your opinion to have an impact? Was the frequency the same if there were different groups involved?;
     - The setting - was treatment received at home or in an institution? Was it the same for different groups of subjects if there was more than one treatment group?
   - These elements need to be addressed if you want to be able to replicate the treatment in your practice.
   - Contamination, Co-intervention avoided? - these two factors were described under Biases (see Design section). Were they addressed? If not, consider what possible issues could influence the results of the study, for example, what could happen if some of the clients in the control group received some treatment inadvertently (contamination) or if some subjects were taking medication during the study (co-intervention)? Make note of any potential influences. If there was only one group under study, mark "not applicable (n/a)" on the form.

10. **Results**
    - Results were reported in terms of statistical significance? Most authors report the results of quantitative research studies in terms of statistical significance, to prove that they are worthy of attention. It is difficult to determine if change in outcomes or differences between groups of people are important or significant if only averages, means or percentages are reported.
    - Refer to the bibliography if you wish to review specific statistical methods.
    - Outline the results briefly in this section, focusing on those that were statistically significant. If the results were not significant statistically, examine the reasons: was the sample size not large enough to show an important, or significant, difference; or were too many outcome measures used for the number of subjects involved.

- Were the analysis method(s) appropriate? Do the authors justify/explain their choice of analysis methods? Do they appear to be appropriate for the study and the outcomes. You need to consider the following:
  - The purpose of the study - is it comparing 2 or more interventions, or examining the correlation between different variables of interest. Different statistical tests are used for comparison and correlation.
  - The outcomes - if there is only one outcome measured to compare 2 different treatments, a simple statistical test such as a t-test will probably be sufficient. However, with a larger number of outcomes, involving different types of variables, more complex statistical methods, such as analysis of variance (ANOVA), are usually required.
- Clinical importance was reported? Numbers are often not enough to determine if the results of a study are important clinically. The authors should discuss the relevance of the results to clinical practice and/or to the lives of the people involved. If significant differences were found between treatment groups, are they meaningful in the clinical world? If differences were not statistically significant, are there any clinically important or meaningful issues that you can consider for your practice?

11. Drop-outs

- Drop-outs were reported? - The number of subjects/participants who drop out of a study should be reported, as it can influence the results. Reasons for the drop-outs and how the analysis of the findings were handled with the drop-outs taken into account should be reported, to increase your confidence in the results. If there were no drop-outs, consider that as 'reported' and indicate no drop-outs in the comments section.

12. Conclusions and Clinical Implications

- The discussion section of the article should outline clear conclusions from the results. These should be relevant and appropriate given the study methods and results. For example, the investigators of a well-designed RCT study using sound outcome measures could state that the results are conclusive that treatment A is more effective than treatment B for the study population. Other study designs cannot make such strong conclusions, as they likely had methodological limitations or biases, such as a lack of a control group or unreliable measures, that make it difficult to "prove" or conclude that it was the treatment alone that influenced the outcome(s). In these situations, the authors may only conclude that the results demonstrated a difference in the specific outcomes measured in this study for the clients involved. The results may not be generalizable to other populations, including yours. Further study or research should therefore be recommended.
- The discussion should include how the results may influence clinical practice - do they offer useful and relevant information about a client population, or an outcome of interest? Do they warrant further study? Consider the

implications of the results, as a whole or in part, for your particular practice and for occupational therapy in general.

Appendix 2 – Consent Form

<div align="center">

CONSENT FORM

Organisation: School of Health and Social Care

</div>

Title of Study: The lumbar spine of fast bowlers: Relationship between lower back pain and bowling actions.

Aim of Study: To establish correlations between back pain history, kinematic and impact variables of fast bowling and future back pain in young fast bowlers.

Researcher' Position: Post Graduate Researcher   **Researcher's Name:** Billy Senington

Contact Details: bsenington@bournemouth.ac.uk; 07833228335

Consent:

- The researchers have explained to my satisfaction the purpose of the study and the possible risks involved.

- I have had the procedure explained to me and have read the participant information sheet. I understand the procedures fully.

- I am aware that I will be required to complete a spell of maximal effort fast bowling with the usual associated risk of injury.

- I understand that any confidential information will be seen only by the researchers and will not be revealed to anyone else.

- I understand that I am free to withdraw from the investigation at any time, without reason.

- I understand that I will not be identified in the study and any information given in future research reports or journal manuscript will be anonymous

- I……………………………………..agree to take part in the study

Signature of Participant…………………………… Date………………………

Signature of Parent/Carer………………………… Date……………………

Signature of Coach…………………………………….Date………………………

Signature of Researcher………………………………Date…………………..…

Back pain/ injury questionnaire

We are going to use this questionnaire to find out about any back pain you may have had or are currently having. It will ask you about specific injuries, when you may have injured your back and will also ask about back pain not due to a specific injury. Please take a moment to consider your back and any pain you have had.

Name: Age:

Weight (kg):   Height (cm):

Race/ Nationality: Occupation:

Have you ever injured your back? Yes ☐   No ☐

If so, when was the first time you injured it?

What were you doing at the time you injured it?

Have you ever had a scan or X Ray? What were the findings?

Did you miss any cricket because of this injury (matches or training)? Yes ☐   No ☐

If so, How much?

Did it fully recover?

Have you injured it again since then? (Please repeat the questions above for each sprain)

How much pain did you experience at this time? (And subsequent times if applicable)

No pain   Worst pain I could imagine
   0   1   2   3   4   5   6   7   8   9   10

No pain   Worst pain I could imagine
   0   1   2   3   4   5   6   7   8   9   10

No pain   Worst pain I could imagine
   0   1   2   3   4   5   6   7   8   9   10

Did you experience any other symptoms, such as pins and needles or numbness? Where? (See pain diagram 1)

Have you experienced back pain (other than any injury outline above)?

Where did you feel the pain? (See pain diagram 2)

Did you have any other symptoms?

What activities aggravated the pain?

Did you miss any cricket because of this pain (matches or training)? Yes ☐   No ☐

If so how much?

Would you say you specifically suffer from back pain associated with cricket?

What aspects of cricket aggravate your pain?

If bowling please expand – is it every time you bowl, after a number of overs, after a number of weeks in the season, on a particular day of the week?

Are there other aspects of your training that aggravate your back pain?

Do you have pain in the 'off-season'? Yes ☐ No ☐

Where is the pain associated with bowling? (See pain diagram 3)

What is the worse pain experienced during bowling?

No pain   Worst pain I could imagine
  0   1   2   3   4   5   6   7   8   9   10

Have you had a medical diagnosis for your back pain?

How do you manage the pain?

Have you received any other significant injuries other than back injury/pain? (See pain diagram 4)

Cricket specific questions.

Please describe you training schedule below:

Mon:

Tues:

Wed:

Thurs:

Fri:

Sat:

Sun:

How many balls do you bowl in a week?

How long have you played cricket?

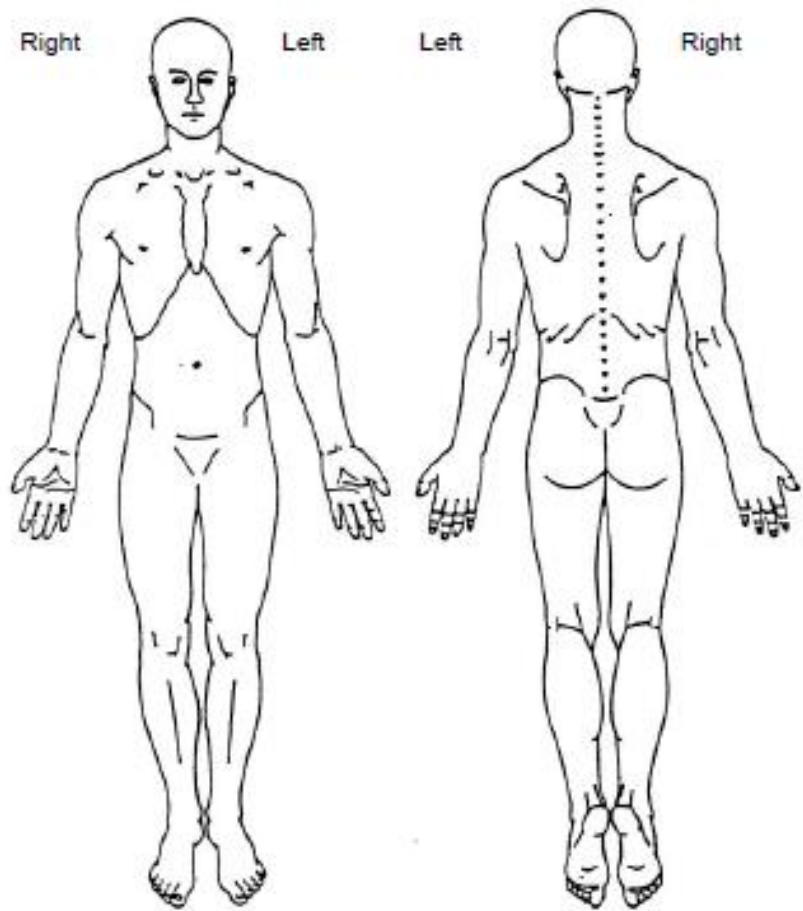How long have you been a fast bowler?

What age did you start playing cricket?

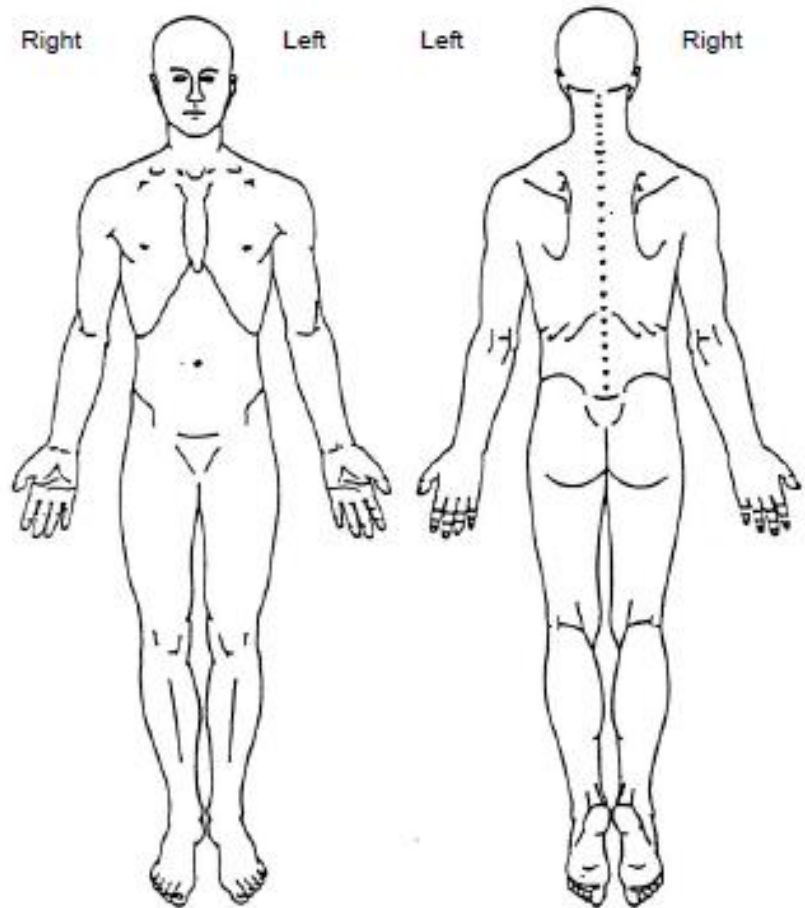What age did you start fast bowling?

Did you have back pain as a child playing cricket?
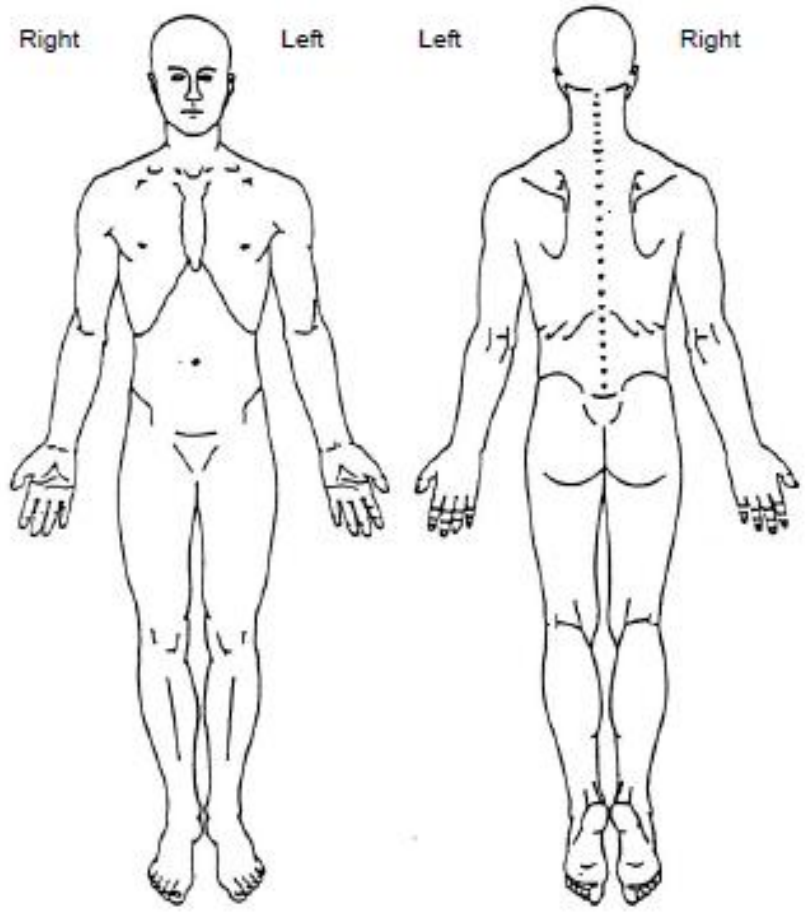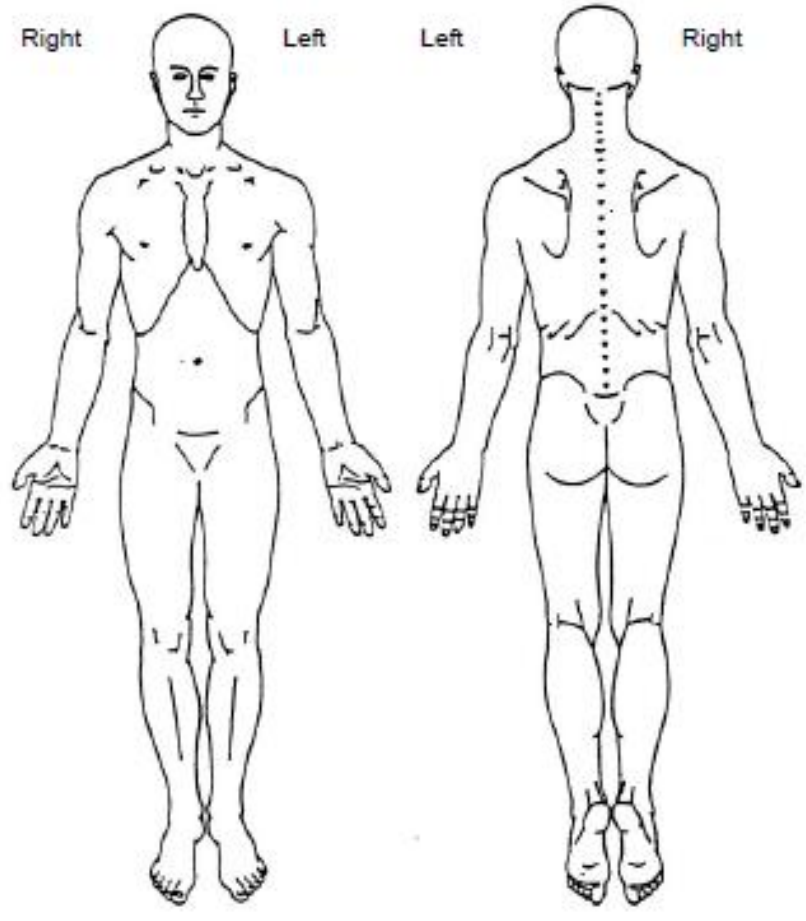
Pain Diagrams

1.

Right   Left   Left   Right

2.

Right   Left   Left   Right

3.

Right   Left   Left   Right

4.

Right   Left   Left   Right

250

Appendix 4 – Development of Methodologies

Surface impact testing

Previous studies investigating surface firmness have validated the use of a Clegg impact hammer testing device (figure I) as an objective measure of surface firmness. As displayed below a Clegg hammer consists of an impact weight with a ±500g. Typically a 2.25kg weight is used for sports surface testing, however different impact weights as low as 0.5kg have been used. This impact weight is dropped from a fixed height inside the vertical guidance tube.

Figure I. Clegg hammer impact weight and guidance tube

Thus, as the Clegg hammer method of playing surface testing has been previously validated, this study aimed to replicate this method. Some adaptations to the experimental setup were made with the consideration of replicating impact characteristics experienced by fast bowlers and use of available equipment. This set up can be seen in figure II.

In order to quantify the surface properties of different cricket playing surfaces, the above custom built impactor was used. A ±200g tri-axial accelerometer (THETAmetrix, ADXL377), sampling at 750Hz and aligned vertically with the centre of mass of an impact weight (63mm in diameter and 2.5kg in weight). The impact weight was

suspended in a guidance tube to standardise drop height to 200mm (Figure II). Furthermore, 20mm of Adiprene polyurethane foam and 3mm of rubber (taken from the heel of a typical sports training shoe) was attached to the bottom of the impact weight to more accurately simulate impact conditions during bowling.

This set up was chosen after pilot testing revealed that a sampling frequency of 750Hz was sufficient to avoid aliasing and a drop height of 200mm was high enough to allow sufficient differentiation between surfaces without exceeding the limits of the attached accelerometer. The addition of the polyurethane foam and rubber also aided in avoiding an underdamped system and thus, undersampling.
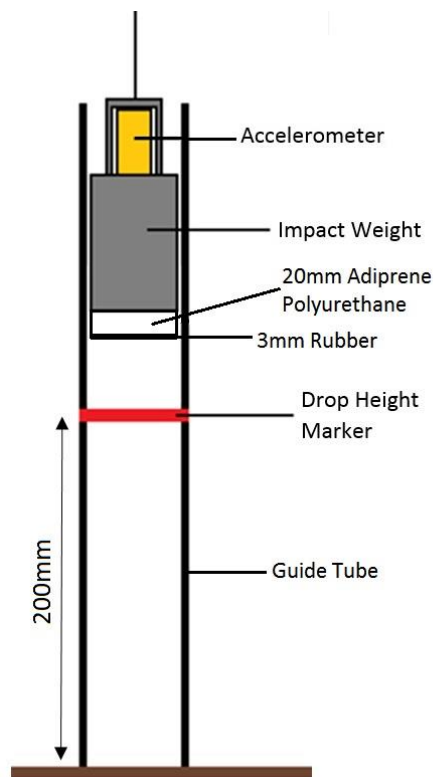
Figure II. Customised Impactor Configuration.

Measurement of tibial and sacral accelerations

Tibial accelerometer attachment

Studies investigating tibial accelerometry have cited a number of different sensor attachment locations and methods. All locations aim to reduce skin artefact by placing sensors on landmarks that provide a secure, flat surface of attachment with minimal tissue between the sensor and the bone. This study aimed to use a sensor attachment

location with the same conditions as above, whilst providing subjects with the ability to move comfortably with unrestricted range of motion when bowling. ISAK recommendations make the measurement of the length of the tibia clear and repeatable. Therefore, tibial accelerometer location was defined as the medial aspect of the mid-tibia. Mid-tibia was defined as the mid-point between the tibiale mediale and the sphyrion tibiale. As the actions being performed are classed as explosive, high impact movements, different sensor attachment methods were piloted in order to find the most secure method of attachment. Pilot studies revealed that attachment to the skin via double sided tape and re-enforcement with a re-usable elastic wrap was sufficient to secure the accelerometer to the tibia whilst still allowing the participant to move freely.

<u>Sacral sensor attachment</u>

Due to the size of the sacral vertebrae, sensor location is much more specific. A line between the left and right posterior superior iliac spines was used to identify the S1 spinous process, with the top of the sensor being placed horizontally aligned with this line (Chakraverty et al. 2007). However, as a result of location, secure sensor attachment was more difficult than with the tibial accelerometer. After pilot testing, attaching the sensor to a small plastic plate which was then attached to the skin with double-sided tape and an elastic adhesive bandage was deemed the most appropriate method of attachment. Total mass of the sensor with addition of the plastic plate was 19g compared with 12g without the plastic plate.

Measurement of spinal kinematics

<u>Attachment of spinal inertial sensors</u>

Due to the curvature of the spine and the gap produced by the paraspinal muscles, attachment of these sensors was more difficult in comparison to the tibial accelerometer. Initially, sensors were attached with double sided-tape to the skin above the S1, L1 and T1 spinous processes and re-enforced with a non-woven adhesive bandage (figure III). The S1 sensor was then further re-enforced with a re-usable elastic wrap. This was not sufficient to limit unwanted sensor movement, as sensor width exceeded that of the gap in the paraspinal muscles and was therefore affected during spinal extension.
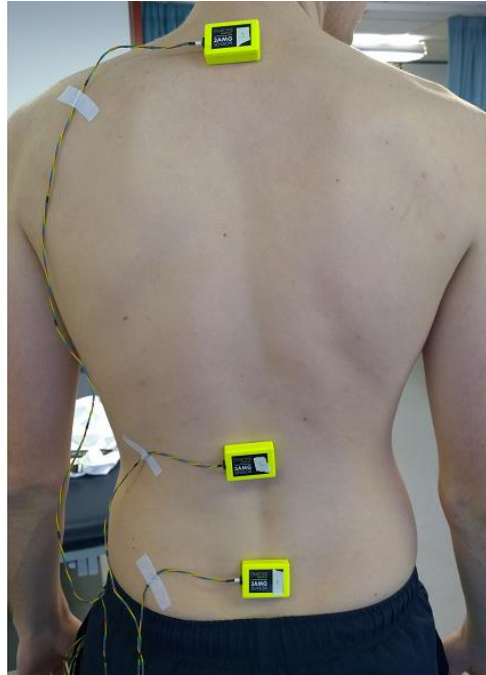
Figure III. Initial sensor attachment method.

Consequently, orientation of the sensors was changed to vertical and re-enforced in the same way as the original method of sensor attachment. This solved the problem of unwanted sensor movement but introduced an issue of unexpected 'out of plane' motions when data was processed, this was due to the re-orientation of the sensors causing 'unusual' sensor orientations during motion. Consequently, it was concluded the sensors must remain in their original orientation, thus this method of attachment was also unsatisfactory.

The next method of attachment incorporated additional attachment placed on the bottom of each sensor, to allow them to sit above the paraspinal muscle gap and remain unaffected during extension (seen in figure IV). Although this was initially successful during static range of motion trials, the attachment proved unstable during the action of bowling and did not remain fixed to the skin and was therefore was also deemed unsatisfactory.
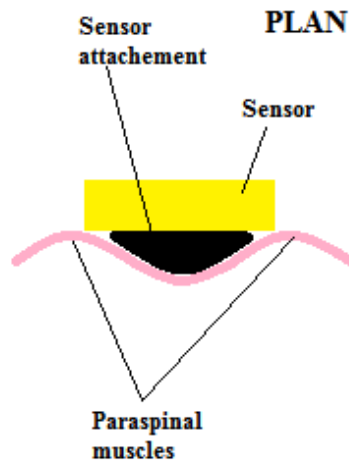
Figure IV. Pilot sensor attachment using additional material to elevate the sensors.

Consequently, a method of bridging the paraspinal muscle gap was employed at the L1 sensor (and S1, for additional support) as seen in figure V. This method provided a stable attachment as well as reliable results (as reported in this thesis).
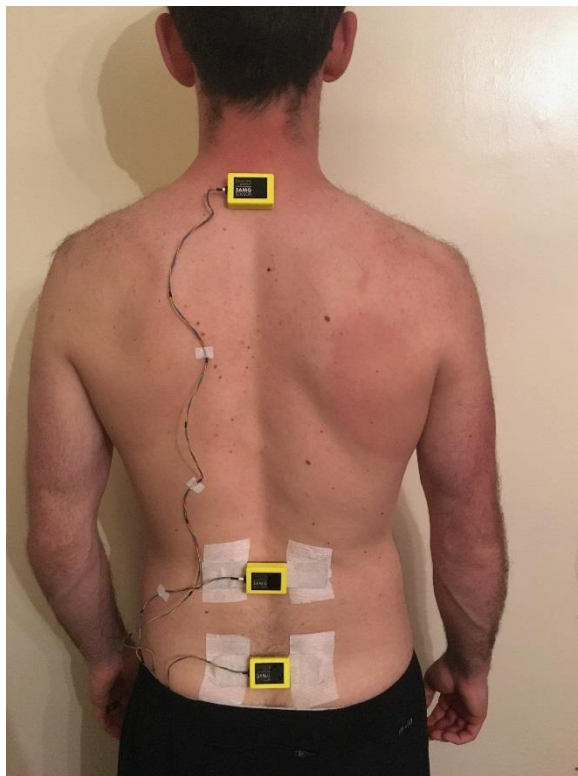

Figure V. Final sensor attachment method.

Development of injury surveillance methodologies

Inclusion Criteria

An all-male sample was chosen to avoid gender becoming a confounding variable, as a result of factors such as muscular strength and power, body composition, height, weight and bowling speed (Stuelcken et al. 2008). Furthermore, as this study looks at adolescent fast bowlers, the fact that female's physical development occurs over a different time period may also affect results.

Injury History

A questionnaire format was chosen for this study as it allows for a wealth of information to be collected in a clear and structured format. No standardised questionnaire encompassing all the required aspects was available. Consequently, a bespoke questionnaire was constructed (see appendix 3). This questionnaire was piloted on a sample of fast bowlers and appraised by three health experts independent from this study to ensure all questioning was unambiguous and only relevant questions were asked.

Before the start of the 2015 season the questionnaire regarding playing history and lower back pain and injury history, along with demographic data were collected from all participants. The questionnaire was completed with the guidance of the researcher and any unknown answers verified with records kept by the club physiotherapist.

Once all questionnaires were completed main themes were pulled from the data for further analysis, these themes can be seen below.

Questionnaire Themes

Participant Demographics: Age, Weight, Height, Race/Nationality.

Low Back Injury: Diagnosed low back injury? Results of diagnosis, Age at first occurrence, Reoccurrence? Cricket missed/ time out due to injury.

LBP: History of LBP? Diagnosis for pain? Age at first occurrence, Repeated occurrence? Cricket missed/ time out due to LBP.

Any other comments or notable injuries.

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