Exploring limb symmetry index for balance across a range of functional tasks

By Michael Neil Gara

A thesis submitted in partial fulfilment of the requirements of Bournemouth University for the degree of Masters of Research

May 2018

Bournemouth University

<u>Exploring limb symmetry index for balance across a range of functional tasks</u>

by Michael Neil Gara

Introduction: Body symmetry and functional reciprocity represent key components of normal movement (Lu & Chang 2012, Sadeghi et al 2000, Watkins 1999) making them essential components of clinical examination. To this end a limb symmetry index (LSI) of 80 – 90% of the unaffected limb has been proposed by previous authors (Daniel et al 1982, Barber et al 1990, Sapega 1990, Petschnig et al 1998). Whilst LSI has been reviewed for a large variety of potential variables, the LSI for balance remains largely unexplored. Balance is viewed as an integral part of maintaining everyday physical activity, a good quality of life and reducing health burden (Clark et al 2016). Therefore, this study aimed to determine the LSI for balance across a variety of functional tasks; whilst reviewing the use of novel yet clinically reproducible methodology. ***Method:*** A cross-sectional observational design was used. Seventeen participants (mean age 27.6±5.7 years) were recruited from the student population at Bournemouth University. Participants reported no existing injury or other balance affecting condition. Balance was measured using two devices: an instrumented wobbleboard (SMARTwobble, THETAmetrix, UK) and a sacral mounted accelerometer (Balance Sensor, THETAmetrix, UK). Participants completed a variety of tasks including forward, lateral and medial hop landing where sacral acceleration was measured for 1 second following landing. Task analysis was completed using SPSS v23, MatLab and Excel. ***Results:*** No statistically significant differences occurred between dominant and non-dominant limb for any of the assessed tasks. The absolute mean percentage difference between limbs was 4.9%±3.7% (95% CI 1.8% - 8.0%). ICC values ranged from 0.73 – 0.96 suggesting moderate to excellent test-retest reliability for accelerometry and wobbleboard. ***Discussion:*** The LSI for balance should be expected to be around 5% regardless of task. Sacral mounted accelerometry, represents a valid and reliable measurement device, for a variety of complex balance assessment tasks including hop landing. Instrumented wobbleboards may also provide a valid and reliable, clinically accessible method for measuring limb symmetry, but may not be appropriate for evaluating a variety of tasks.

Contents

List of Illustrations and Tables

Illustrations

Acknowledgements

I would like to offer my deepest gratitude to Dr Jonathan Williams and Dr Carol Clark for their phenomenal support, patience and guidance throughout my long and complex journey.

Declaration

I declare that this thesis is presented solely for the consideration for the award of Masters by Research (MRES) from Bournemouth University and has not been presented elsewhere in any form.

This project and thesis is the sole work of the author except for the bespoke algorithm used to establish the hop landing scores, which was devised by Dr Jonathan Williams.

1.Introduction

Balance has many varying definitions dependent upon the circumstances of their use. It is important to recognise the need for correct interpretation of balance as it is an important component of human function. Two key dictionary definitions are important in understanding the need for this project:

1. an even distribution of weight enabling someone or something to remain upright and steady.
2. a situation in which different elements are equal or in the correct proportions (Oxford Dictionary 2018)

The first definition is a priority of human development, as it represents a key challenge in the transition from quadrupedal to bipedal ambulation, which indicates an increasing functional maturity. The second definition is intrinsically linked to the first due to the need for all the elements that effect human balance to be present in the correct proportions. Several variables are thought to affect an individual's balance – visual & vestibular input, neuromuscular impulses, muscle strength, joint range of movement (ROM), bone density and proprioception (Cug et al 2014); although it is not fully understood in what proportions these variables normally occur, or what the correct alignment of these variables are, when a person demonstrates functional balance.

The comprehension of these balance inputs is further complicated, by the necessity for separate classifications of balance, faced as a construct of normal human movement. The most common classifications of balance are static and dynamic; static balance is the ability to maintain postural stability and orientation with centre of mass over a stationary base of support, whereas dynamic balance is the ability to maintain postural stability and orientation by maintaining an equilibrium between a moving centre of mass and a fluctuating base of support (Meyer & Ayalon 2006, Sullivan & Portnry 2014). However, this dichotomous definition of balance fails to recognise the subtle overlapping required during everyday balance tasks. Thus, balance might best be viewed as a spectrum which combines elements of the static and dynamic definitions to create the stability needed to perform everyday functions. Gait kinematics perfectly emphasise the need to view balance as a spectrum; the break down between stance phases and swing phases (Uustal & Baerga 2004) highlighting the regular transition from static to dynamic balance. Adding in everyday functional tasks, such as climbing and descending stairs or standing from a seated position, further enhances the need to envisage the construct of balance as more than just static or dynamic.

Why is balance important?

Balance is viewed as an integral part of maintaining everyday physical activity, a good quality of life and reducing health burden (Clark et al 2016). Falls are a key problem for the NHS and are estimated to cost around £2.3bn per year (NICE 2013). With 30% of over 65-year olds and 50% of over 80-year olds falling at least once per year (NICE 2013), gait deficiencies and balance disorders represent the second largest cause of falls in older adults (Noohu et al 2014). However, despite the recognition that falling has gained as a symptom of problematic function; there are many variables in balance measurement and a limited understanding of normal values for balance. Key balance measurement tools range from the large and expensive centre of pressure (COP) force plates to the cheap and easily available balance assessments such as the Berg balance scale or Timed Up and Go. This variability in balance measurement makes it increasingly difficult to establish the degree of impaired balance across the spectrum of functional balance and thus the efficacy of training programmes for individuals in a clinical environment.

Balance Measurement & Testing

The literature is full of examples of methods to measure balance; ranging from the complex, expensive laboratory-based devices to simpler clinical measures. It is evident that different measures are likely to measure different constructs of balance along the static/dynamic continuum. Furthermore, each method has its own unique set of specific strengths and limitations.

Force Plate

Centre of pressure (COP) is often measured using a force plate. This allows for measurement of pressure changes in both anterio-posterior (AP) and medio-lateral (ML) positions, which is correlated to postural stability via the inverted pendulum theory (Winter 1995), where the smaller the pressure area the better the postural stability. Whilst force plates are often used to assess postural stability by researchers, their cost and size make them less popular in clinical settings when compared to more readily available cheaper alternatives. A commonly used example of force plates in research is the Biodex Stability System; Hinman (2000) reviewed the test-retest reliability of this equipment and suggested an Intraclass Correlation Coefficient (ICC) of 0.86-0.87, whilst Golriz et al (2012) reviewed a more clinically accessible portable force plate (Midot Posture Scale Analyzer (MPSA)) and suggest ICC values of 0.63-0.83 when measuring postural sway. The more portable and clinically accessible force plate being deemed less reliable than the Biodex Stability System.

Accelerometry

Whilst COP is thought to represent postural stability, using the inverted pendulum theory (Winter 1995), it is also debated whether changes in COP correlate to changes in centre of mass (COM) (Winter 1995). Accelerometry has been suggested as an alternative measurement device for postural sway. Such technology has been proposed as an instrument to investigate balance (Moe-Nilssen & Helbostad 2002) and similar to a force plate it can measure linear movement in an AP and ML direction (Williams et al 2016). However, unlike a force plate the accelerometer can be positioned close to the suspected COM for the individual (Williams et al 2016) allowing for quantification of COM acceleration along the respective linear axes. Positioning the sensor near the suspected COM, in theory, takes account of the composite sway of all lower limb joint stability (hip, knee, ankle), rather than just the ankle stability associated with COP measurement (Winter 1995). Winter’s (1995) theory of inverted pendulum suggests a greater sway area at the COM when compared with COP sway; but that over a sustained period the COM sway may correlate with COP. Investigations comparing accelerometry and force plate measurement suggest this to be true (Whitney et al 2011, Seimetz et al 2012) and with COM providing a larger sway area, it may be expected that measuring COM will provide more accurate data relating to balance ability. Tri-axial accelerometry can also record movement along a vertical axis which can be used to account for tilt in space of the accelerometer during balance tasks. ICC values for accelerometry range from 0.74-0.97 (Saunders et al 2015, Williams et al 2016).

<u>Wobbleboard</u>

Wobbleboards are a commonly used modality, for both sport-specific and non-specific training programmes, aiming to increase proprioception and balance (Emery et al 2005, Ogaya et al 2011). The reported links between balance and wobbleboard training (Ogaya et al 2011) have highlighted the need to quantify wobbleboard performance so that changes can be objectively measured (Williams & Bentman 2014). To this end wobbleboards have been constructed to record permutations in tilt angle during task performance. The ThetaMetrix ‘Smart’ wobbleboard has been shown to have good test re-test reliability (ICC = 0.71), when measuring wobbleboard performance, across a series of tasks (Williams & Bentman 2014). Williams & Bentman (2014) also suggest that the ‘smart’ wobbleboard can differentiate between task difficulty, thus it can be considered as a valid balance measurement device, although it has not been compared against other validated balance measurement tools.

<u>Berg Balance Scale</u>

The Berg Balance Scale was created in 1989 to objectively quantify balance in the elderly (Downs 2015). It is reported to be a valid and reliable tool which can be easily administered in a variety of settings with minimal equipment (Downs 2015). The scale comprises 14 tasks of varying difficulty beginning with assessment of the participants sit to stand and progressing to standing on one leg. Each task is scored from 0 (unable) to 4 (highest level of function) (Downs 2015). It is often used to predict risk of falls in a clinical setting – particularly for people with disabling conditions including advanced age (Downs 2015). The Berg Balance Scale has been shown to have excellent reliability (ICC 0.97) (Downs 2015). Despite its recognition in clinical practice, the Berg Balance Scale is of limited use as a balance tool for assessment on participants under the age of 75, without an associated condition affecting balance (Downs 2015). The lack of complexity in the challenges assessed, means participants reach a ceiling rapidly (Downs 2015) and therefore the appraisal of balance is desensitised, for 'normative' clinical data.

Timed Up and Go

The Timed up and go (TUG) is another assessment tool frequently used by clinicians to predict falls risk in elderly or physically challenged populations (Beauchet et al 2011). The participant starts from a seated position in an armchair. They are asked to stand, walk 3 metres, turn and walk back to the chair (Beauchet et al 2011). The timer stops when the participant returns to a seated position in the chair. The clinically significant value for falls prediction is thought to be a task completion time of greater than 14 seconds (Shumway-Cook et al 2000), however varying studies into the TUG test suggest the clinically significant value lies somewhere in a range of values from 11 to 32.6 seconds (Beauchet et al 2011). The study by Resnik & Borgia (2011) investigated the minimum detectable change (MDC) when using the TUG test. They determined the MDC to be 3.6 seconds – therefore any change in TUG time must exceed this MDC figure to be recognised as a true clinical change. This relatively large MDC represents an easily achieved floor effect, where participants will rapidly be unable to show clinical improvement. The TUG is recognised as a functional evaluation tool however similarly to the Berg Balance Scale it is not sensitive enough to detect balance deficiencies in a normative population. Resnik & Borgia (2011) documented an ICC value of 0.88 when testing participants with a lower limb disability.

Star Excursion Balance Test

The Star Excursion Balance Test (SEBT) is referred to as an assessment of dynamic postural control (Bouillon & Baker 2011). Participants are required to single leg squat whilst reaching in 8 different directions with their contralateral limb (Gribble et al 2012). Distances reached are then recorded to provide an objective outcome measure. Kinzey & Armstrong (1998) recorded ICC values for test-retest reliability for the SEBT to be 0.67-0.87. The SEBT is credited as sensitive enough to differentiate between an injured and uninjured lower limb (Gribble et al 2012). Gribble et al (2012) recorded significant differences ($p<0.05$) between the injured and uninjured limb, for unilateral chronic ankle instability, in the anterior, posterior and medial direction. Although the SEBT has been investigated for multiple variables – injury prediction, fatigue, rehabilitation outcomes (Gribble et al 2012, Bouillon & Baker 2011); it has not been used to investigate normal between limb differences. The Y balance test is a smaller version of the SEBT – it consists of just 3 directional challenges (anterior, posterolateral and posteromedial), it was deemed a reliable alternative due to strong correlations with the SEBT (Gribble et al 2012). Alhnadi et al (2015) reviewed the difference between dominant and non-dominant lower limb and suggest that the normal between limb difference under the Y balance test is approximately 99% of the contralateral limb.

Subjective Assessment of Balance

Given the association of balance to function, it has been speculated that functional questionnaires may be able to provide an indication of expected poor balance performance (Resnik & Borgia 2011, Clark et al 2016). The 36-item short form health questionnaire (SF-36), the patient-specific functional scale (PSFS) and the functional difficulties questionnaire (FDQ-9) all represent valid questionnaires for determining reduced function (Resnik & Borgia 2011, Clark et al 2013). The SF-36 recorded an ICC of 0.61 for physical function (Resnik & Borgia 2011) and the PSFS recorded an ICC of 0.81 (Resnik & Borgia 2011). The FDQ-9 was found to have a significant moderate correlation with the ThetaMetrix Smart wobbleboard (Clark et al 2016); demonstrating that increased perceived functional difficulties were associated with lower balance scores (Clark et al 2016). The reliability of the FDQ-9 was recorded as 0.81 (Clark et al 2016). Therefore, the use of subjective questionnaires, although not objectively able to measure balance, may provide an interesting method to preview expectations for balance performance (Clark et al 2016).

Limb symmetry index

A limb symmetry index (LSI) was proposed to provide guideline clinical values for between limb comparison (Daniel et al 1982, Barber et al 1990, Sapega 1990, Petschnig et al 1998). Several functional traits of the lower limb have been reviewed for symmetry. The suggestion for strength is that between limb similarity should be between 85-95% (Armstrong & Oldham 1999, Knapik et al 1991, Lanshammar & Ribom 2011). Barbieri et al (2015) suggest that LSI for lower limb power would be approximately 88% when kicking a ball. Vaisman et al (2017) measured power using a vertical hop task and identified an LSI of ≥85% in 95% of their participants. The study by Barbieri et al (2015) would also suggest an LSI for lower limb accuracy of approximately 58%. Single limb hop forward for distance limb symmetry is predicted to be 85-96% (Barber et al 1990, Munro & Herrington 2011, Gokeler et al 2017). LSI for weight distribution in dual limb quiet standing was suggested to be approximately 93% (Eliks et al 2017). Whilst the literature review conducted as part of this project (chapter 2 p.16) predicts lower limb balance symmetry to vary from 82-93% between limbs during single limb quiet standing. Lower limb symmetry marks an important consideration for functional balance – e.g. during the phases of gait, a person transitions from one leg to the other and must adjust the determinants of balance, so that they can remain upright and stable (Uustal & Baerga 2004). However, single limb balance in quiet standing isn’t representative of all the balance challenges faced during everyday functional activity. For example, reciprocal ascension of stairs requires an alternation of single limb balance from right to left leg, combined with knee and hip extension.

For more advanced function, such as participation in sport, stability during a more complex plyometric loading and unloading of a single limb is necessary to successfully complete tasks. Therefore, hop testing is often used as an outcome measure for post-operative anterior cruciate ligament (ACL) rehabilitation – the aim to establish limb symmetry for either hop height or distance or both as being representative of a return to full function (Rohman et al 2015, Logerstedt et al 2012, Petschnig et al 1998). Although this method works as a basic determinant of limb symmetry, it focuses on comparisons of lower limb power, rather than measuring stability of the lower limb when stressed. The high re-injury rates for ACL following rehabilitation would also suggest that current criteria for return to sport are not sufficient (Wellsandt et al 2017).

Alongside rehabilitation goals, injury prediction is an important consideration for athletes, coaches and healthcare professionals (Moran et al 2017). As such, attempts have been made to develop injury prediction screening tools (Jackson et al 1978, Gabbett 2010, Bird & Markwick 2016, Lawrence et al 2017, Moran et al 2017). Limb symmetry is considered fundamental to injury avoidance; however, standard variations in between limb values are not fully understood for reproducible and relevant clinical tasks (Lawrence et al 2017). To date, a large number of different tests have been suggested to examine limb symmetry. However, given the relationship between balance and a large number of variables, it would stand to reason that balance asymmetry could be considered a highly sensitive predictor of injury. Therefore, to fully understand deficits in lower limb balance symmetry, it is important to attain normative LSI values for single limb balance tasks other than in quiet standing.

Summary

The reviewed balance testing procedures and equipment represent a selection of the popular clinical assessment methods. However, the variety of testing procedures and the criteria needed for successful measurement, mean that accelerometry likely represents an important valid and reliable balance assessment tool to be used for multiple functional tasks in clinical practice. The portability of the accelerometer makes it an easy to administer, minimally invasive balance measurement device, with potentially excellent reliability; whilst its ability to measure postural sway during several different tests makes it an ideal choice for functional balance assessment. The ‘smart’ wobbleboard meets the same criteria of easy to administer and minimal invasiveness but has a reduced reliability and lacks the concurrent validity testing of accelerometry. There is also the potential for subjective estimation of balance deficits using simple easily administered clinical questionnaires.

The LSI lower limb values range from 58-96% dependent on characteristic being measured. The majority of LSI characteristics fall in the range of 85-95%. The LSI for balance has only been reviewed for single limb quiet standing and as a result there is a need to further explore balance asymmetry across a variety of more complex tasks to provide a greater understanding of the likely range of LSI for balance.

Aims

The aims of this project are to:

1. Establish an expanded understanding of LSI for balance through testing more complex tasks.
2. Assess the validity of the instrumented wobbleboard through comparison with accelerometry.
3. Review the reliability of accelerometry in measuring hop landing performance.
4. Examine the relationship between the FDQ-9 and complex balance tasks.

2.Literature Review

Introduction

Body symmetry and functional reciprocity represent key components of normal movement (Lu & Chang 2012, Sadeghi et al 2000, Watkins 1999), making them essential aspects to review in a clinical examination. Comparison of limb function with the contralateral side often provides a consistent identifier of impaired function, 'auto-normalised' to the individual, whilst providing a target outcome for rehabilitation. To this end a limb symmetry index (LSI) of 80 – 90% of the unaffected limb has been proposed by previous authors (Daniel et al 1982, Barber et al 1990, Sapega 1990, Petschnig et al 1998), suggesting that limb variables exhibit approximately a 10 - 20% asymmetry as standard. A large amount of research has identified that LSI pertaining to strength ranges from 85-95% symmetry (Armstrong & Oldham 1999, Knapik et al 1991, Lanshammar & Ribom 2011), thus providing clinicians with a clear clinical target for strength rehabilitation of an affected or injured limb. Despite this being clearly outlined for strength, to date no such information is evident for balance. Balance is known to be affected by injury (Baierle et al 2013, Burnett et al 2015) and is a common component of many rehabilitation protocols (Owen et al 2006). Despite this, clear clinical targets such as LSI for single limb balance are currently lacking. As a result, the question remains as to what difference between single limb balance is considered normal human variation and what should be considered impairment. Therefore, the aim of this review is to synthesise the available literature to quantify the level of symmetry for single limb balance.

Method

Search Strategy

A systematic search of MEDLINE, ScienceDirect, J-Stage, SPORTDiscuss, Directory of Open Access Journals, PsychINFO and CINAHL was conducted in November 2016. A breakdown of search terms and associated Boolean logic can be reviewed in appendix 1. Non-peer reviewed articles and articles not published in the English language were excluded at the search stage and after removal of duplicates, seventy-three articles of potential interest were found. Titles and abstracts were initially screened for inclusion before full texts were reviewed against inclusion and exclusion criteria. A flow chart outlining the article selection process can be seen in appendix 2.

Inclusion and Exclusion Criteria

As the aim of this review was to discover normative values for single limb balance symmetry, only articles investigating healthy participants were used. No age or date restrictions were applied and no limitations with respect to measurement methods were imposed. All included studies had to report measures of single limb static balance on each leg, however interventional studies were acceptable if this pre-intervention data was accessible.

Quality Index and Analysis

A data extraction table was created to enable identification of underlying commonalities for further discussion (Appendix 3). A modified version of the Downs and Black (Downs & Black 1998) checklist for non-randomised studies was used to analyse the quality of selected studies allowing for comparison based on strength of evidence. For this review question 27 was removed from the checklist secondary to the lack of data provided by any study to perform power calculations. The scoring was also modified so that where a criterion was unable to be determined (UTD) it is not penalised in the same way as a clear no; to do this the scoring was YES = 1 UTD = 0 NO = -1. Final scores were then given as percentages of potential achievable score. This is a new method established for this review and has not yet been reviewed for validity or reliability.

Meta-Analysis

A meta-analysis was created to collate and analyse the results of all the studies. SPSS v23 was used to review the results for normality and significant differences between means.

Results

Ten relevant articles were identified for inclusion by the search strategy. The majority of papers were cross sectional observational studies, three of the studies tested single limb balance against an intervention (Johnson & Leck 2010, Kilroy et al 2016, Pau et al 2012) and two of the studies included a control population (Kilroy et al 2010, Pau et al 2012) whilst the third study incorporated a cross over design (Johnson & Leck 2010). A total number of 439 participants were reviewed (male n= 131, female n=190, n=118 unrecorded).

Quality Index

Downs and Blacks checklist scoring was converted into a percentage and reported in Appendix 3. The quality scores ranged from 35-67% with a mean score of 53±10%. These scores would suggest an overall moderate threat to methodological bias and therefore represents a potential limitation of the findings. Despite this, often common threats to validity would be unlikely to affect the outcomes of interest for this review. Questions 8 & 9 failed to score; but as these studies were measuring 'normal' uninjured participants no adverse events would be likely and no patients would be lost to follow up due to the cross-sectional nature of the studies. Questions 12 & 13 relate to the participants and the environment, under the specifications of the checklist, they scored negatively but there is the argument that the study participants would be representative of the normal population because they were 'normal' uninjured participants. The facilities were not representative of what patients would receive but the participants were not patients and no interventions were being received. Questions 14 & 15 relate to blinding of participants and examiners from the intervention; but there was no intervention received and the majority of measurements were taken using automated computerised methods which would limit subjective bias. Failing to score on these six elements would reduce each study's methodological score by approximately 23% but would likely have a much smaller effect on recorded outcomes and transferability of the studies to the wider population.

Balance measurement

Seven of the studies used force plates/platforms to measure centre of gravity/pressure/mass (Chew-Bullock et al 2012, Cug et al 2014, Kilroy et al 2016, Masu et al 2014, Matsuda et al 2008, Pau et al 2012, Teranishi et al 2011), one study used tri-axial accelerometery (x2 sensors mounted to head and waist) to measure postural sway (Eguchi & Takada 2014), one used an optoelectronic motion capture system to monitor sway throughout the ankle, knee and hip (Clifford & Holder-Powell 2010) and one simply timed how long an individual could remain balancing on one leg (Johnson & Leck 2010).

Limb Symmetry Index Values

The collected study findings for between-limb balance variation can be seen in Appendix 4. The differences between dominant and non-dominant limb were calculated as a percentage difference against the dominant limb and weighted means were calculated against total participants. Matsuda et al (2008) did not provide clear figures in their results (only providing illustrations) and could not be included in the analysis; two other studies (Cug et al 2014, Teranishi et al 2011) failed to provide standard deviations. The overall mean balance difference between dominant and non-dominant lower limb was 11.3%±17.8%. The confounding variables male/female, under 18/over 18 and athletic/non-athletic were evaluated. A Shapiro-Wilks test was conducted to establish if results were distributed normally; none of the results from the confounding variables followed normal distribution and a Mann-Whitney U non-parametric test was used to calculate the likelihood of significance between groups. The breakdown between males (n=131) and females (n=190) was 8.2%±12.9% and 18.1%±19.3% respectively with a $p<0.679$ suggesting there is no significant difference in limb symmetry index between males and females. A significant difference was found ($p = 0.01$) between under 18's (16.8%±20.1%) (n=198) and over 18's (7.7%±17.5%) (n=241). As several of the studies included participants with a focused 'athletic' background (Kilroy et al 2016, Masu et al 2014, Chew-Bullock et al 2012, Pau et al 2012), calculations were done to assess the difference between 'trained' and 'untrained' participants; the breakdown between ‘trained’ (n=87) and ‘untrained’ (n=352) participants was 17.9%±33.9% and 9.9%±14.3% respectively with a $p<0.373$ suggesting no significant difference in limb symmetry index between trained and untrained individuals.

Discussion

This study aimed to establish a potential value for normal variation in between-limb balance performance. The evidence would suggest the dominant leg displays on average 11.3% better balance than the non-dominant leg. This figure falls into the previously indicated LSI range (10 – 20% (Daniel et al 1982, Barber et al 1990, Sapega 1990, Petschnig et al 1998)) and is similar to that previously discussed in relation to LSI for strength (5 – 15% (Armstrong & Oldham 1999, Knapik et al 1991, Lanshammar & Ribom 2011)); suggesting that LSI for balance may be closely associated to strength. Given the large number of variables that affect balance, it may be purely coincidental that the limb symmetry difference between strength and balance are similar for dominant and non-dominant lower limb, or perhaps muscle strength plays the primary role in balance ability. Several studies support this theory by highlighting a strong association between strength and balance (Handrigan et al 2010, Mackey & Robinovitch 2006, Pijnappels et al 2008a, Pijnappels et al 2008b). This would then infer that if muscle strength is the key determinant of balance, the other factors such as visual and vestibular input, proprioception and joint range of movement only serve to increase or limit the efficiency of muscular strength and in turn an individual’s balance ability. The implications of this could mean that where balance is compromised, without a clear method of injury, then a lower limb strength training programme might be the go to clinical suggestion; to return the balance deficit without specific balance training. Thus, increased balance asymmetry, outside of the specified values, may provide a diagnostic tool for lower limb deficits.

Balance symmetry differences between males and females do not bare significance ($p=0.594$). Males seem to exhibit greater balance symmetry than females and although both show better balance performance on the dominant leg there is greater symmetry for males (7.0%) when compared with females (18.1%). With falls in the elderly (>70 years) being more prevalent in females than males (Painter & Elliot 2009, Bryant et al 2005) and poorer balance scores being associated with increased falls (Aslan et al 2008), the difference in male and female single limb balance might suggest that the poorer balance scores for women lead directly to increased falls. Painter & Elliot (2009) also documented that elderly females experienced a greater fear of falling than males and this may be a direct effect of a subconscious recognition of balance asymmetry, in particular during ambulatory tasks, where a large amount of time is spent in single limb stance during the swing phase of gait (Uustal & Baerga 2004). However, as the articles included in this review had a maximum participant age of 38, the findings of this review cannot be directly related to an elderly population; therefore, it is impossible to successfully speculate on single limb balance scores between males and females in an elderly population. Particularly as reductions in muscle strength (Low-Choy et al 2007), bone density (Cannon et al 2001, Daly et al 2013) and age-related cognition (Elosua et al 2017) provide multiple obstacles to maintaining balance as a person gets older. However, Wilder & Cannon (2009) discussed a year by year reduction in strength with up to a 40% reduction in muscle strength between age 30 & 65 years old. Whilst Cannon et al (2001) concluded that muscular innervation did not diminish with age and reductions in muscle contractile force were the result of decreased muscle mass, suggesting the possibility that neurological inputs have a decreased responsibility for poor balance compared with reduced muscle strength. Potentially further supporting the theory that muscle strength is the primary determinant of balance ability.

Despite the increased falls risk associated with aging (>70 years) (Nakagawa et al 2017), single limb balance symmetry significantly improved (p=0.01) between under 18's (16.8%) and over 18's (7.7%). The included studies examined participants with an age range of 4-38 years old and the maximum mean age of studies including over 18's was 27.2±1.4 (Clifford & Holder-Powell 2010). The findings of Eguchi & Takada (2014) would support a theory that single limb static balance improves as a child becomes more proficient at walking. It might be considered that the nature of reciprocal gait, and its requirement of the individual to place equal emphasis on both legs to ambulate normally (Uustal & Baerga 2004), lends itself to a balance symmetry and negates standard stereotypes of dominant and non-dominant limbs. Thus, as a person ages their increased single limb balance ability is likely the result of increased practice in single limb standing as a component of normal walking. This would also suggest the potential for dynamic and static balance to be intrinsically linked; with the dynamic balance practiced in normal gait perhaps giving rise to positive developments of static balance for the individual.

Further differences were noted between ‘trained’ and ‘untrained’ participants. Trained participants were defined as participants who had engaged in a recognised physical activity such as dancing or sport; whilst the untrained participants were those who had no specific focus on physical activity. No significant difference (p=0.373) exists between trained and untrained participants with percentage differences calculated to be 17.9% and 9.9% respectively. This suggests that trained participants displayed an increased level of balance asymmetry than untrained participants. Zvijac et al (2014) predicted that in elite American Football athletes muscle strength LSI may be around 41%. Given that this is much larger than the 5 – 15% suggested previously for strength, then the implication is that limb symmetry index for balance for trained individuals may also be larger than untrained individuals. Potentially, the greater asymmetry arises as a result of increased unilateral demand, placed upon the dominant limb by athletic participants (Maulder 2013). Therefore, it may be reasonable to expect slightly larger variations in dominant v non-dominant balance symmetry for athletic individuals.

However, it should be considered that this discussion is based only on values for static single leg balance symmetry. It may not appropriate to suggest that these measurements are representative of the true spectrum of single limb balance. Further research is needed to examine the relationship between dominant and non-dominant single leg balance using a variety of tasks; thus, attempting to measure the variability of single limb balance in respect of other functional challenges.

Conclusion

The LSI for balance appears to fall in line with previously established LSI ranges. LSI for balance would be expected to be around 11.3% with a potential range of approximately 7 – 18% difference between limbs. Confounding variables such as age, gender and athletic background may all have a bearing on the expected level of lower limb balance symmetry. Although the only significant difference was between over and under 18’s, it would appear that there are subtle differences between gender and athletic background which give rise to a potential spectrum for LSI for balance, based on the physical characteristics of the individual. Knowledge of these ranges for lower limb balance symmetry, may provide a diagnostic indicator of lower limb dysfunction and highlight issues such as reduced strength, proprioception, joint range and visual/vestibular deficits, albeit without specificity as to the underlying primary cause. Therefore, LSI values for balance should be established for as many variables as possible; particularly for more complex tasks than single limb stand.

3.Method

A cross-sectional observational design was employed to assess the aims of this project. This study aimed to establish dominant v non-dominant single leg balance values, comparisons between balance measurements and correlations between balance and existing function in healthy participants; with no intervention element. All participant data was collected at one point of contact.

Ethical Approval

This study was granted ethical approval by Bournemouth University's Research Ethics Committee in line with Bournemouth University's Research Ethics Code of Practice. All participants provided written informed consent prior to data collection.

Participants

An estimated sample size of 30 participants was determined from sample size calculations, using figures from previous studies comparing dominant and non-dominant lower limb balance. Previous figures indicated that with 8% expected difference and standard deviation (SD) of 17.5%, an effect size estimate of 0.46 and a sample size estimate of 30, assuming an alpha of 0.05 and 80% power.

Participants were recruited via social media from the student population at Bournemouth University. All participants were vetted against exclusion criteria to avoid conditions which may compromise balance. The exclusion criteria were determined through peer discussion and in-line with existing balance literature. All participants were provided with a participant information sheet at least 24 hours prior to providing informed consent.

Exclusion Criteria

- Previous substantial injury to either lower limb (which required surgery or plaster casting).
- Current injury/pain in either lower limb or in the last 12 months.
- Current/previous head injury or ongoing neurological disorder.

- Known balance problems (including but not limited to vertigo or dizziness).
- A history of falls
- A history of chronic debilitating illness
- Age >50.

Participant Demographics

Seventeen participants were recruited for the study, before time constraints of the project forced the cessation of data collection. Data collected from participants included age, gender, height, weight, shoe size and opinion of dominant foot to assist with the creation of a demographic profile.

Procedure

All participants were required to provide written informed consent and confirm that none of the exclusion criteria applied to them, prior to commencing data collection. Participants were asked to complete the FDQ-9 before performing any balance tasks. Demographics for height and weight were self-declared. Dominant lower limb was established by asking the participant which would be their preferred foot when kicking a ball (Hoffman et al 1998). Previous studies have demonstrated significant differences between barefoot and shod static balance (Smith et al 2015, Kilroy et al 2016); therefore, to standardise outcomes all tasks were completed barefoot. The wobbleboard was placed on a non-slip mat for safety, whilst the accelerometry challenges took place on a hard level floor. Hand support was placed in front of participants for the single leg balance and squat tasks. All data were collected by the same researcher.

Instrumentation

Wobbleboard

An instrumented wireless wobble board (SMARTwobble board, THETAmetrix, Waterlooville, Hampshire, UK) was used to quantify wobble board performance. This wobbleboard device houses a miniature tilt sensor within the dome base and communicates to a personal computer through a bluetooth connection (Williams & Bentman 2014). Tilt angle is measured at 15Hz and data is separated into bands which each represent a third of the maximal tilt achievable. Therefore, data are presented as a percentage of time spent with the board between 0 - 5^{o}, 5.1 - 10^{o} and 10.1 – 15^{o} tilt, as well as percentage time spent on the edge of the board. Additionally, the number of edge contacts is also recorded. Therefore 5 dependent variables were measured:

- % Time with the board held in the inner band
- % Time with the board held in the middle band
- % Time with the board held in the outer band
- % Time with the board held on the edge
- Number of edge contacts

The wobbleboard has been previously shown to offer face validity and moderate reliability (ICC 0.71) (Williams & Bentman 2014).

Accelerometer

A commercially available balance sensor (THETAmetrix, Waterlooville, Hampshire, UK) was used to quantify balance. The balance sensor device houses a wireless inertial sensor (fusion of triaxial accelerometer and triaxial gyroscope) which communicates via Bluetooth to a PC with specifically written software. Acceleration data are then converted into anterio-posterior (AP) and medio-lateral (ML) accelerations (corrected for sensor tilt and removal of gravitational force component) which are used to represent postural sway. The sensor is attached to the skin over vertebrae L4-S1 (located by palpation of PSIS and iliac crest) using double sided hypoallergenic tape. The software quantifies performance at 100Hz using three metrics - normalised path length, root mean square and jerk of the AP and ML acceleration traces. Triaxial accelerometry has been shown to be a valid and reliable (ICC 0.74-0.97) balance measurement device (Saunders et al 2015, Williams et al 2016).

Functional Difficulties Questionnaire (FDQ-9)

The FDQ-9 was originally created to assess for Developmental Coordination Disorder (DCD) in adults (Clark et al 2013). It demonstrated good internal reliability (ICC 0.81) and has also been validated as a predictor of balance against the wobbleboard (Clark et al 2013, Clark et al 2016). It is a 9-item questionnaire which encompasses the main areas of fine and gross motor coordination including balance (Clark et al 2016). Participants rate their abilities on a four-point Likert-type scale with: 'Very good' (1), 'Good' (2), 'Poor' (3), 'Very poor' (4), as possible options for each question asked (Clark et al 2016). Possible scores range from 9 to 36. Normal functional ability is represented by a score of ≤ 20, whilst functional difficulties are indicated by scores of >21 (Clark et al 2013). The version of FDQ-9 used also encompassed a demographic profile questionnaire and a screening tool for balance affecting disorders (Appendix 5).

Tasks

Balance tasks were divided between wobbleboard and accelerometer challenges. The task categories were randomised via coin toss. The individual tasks were completed, in a standardised order of perceived difficulty, from easiest to hardest. The aim was to standardise any learning affect that might be gained if the more complex tasks were assessed prior to the more straightforward tasks; furthermore, this was also deemed the best method for participant safety. Dominant limb was always measured first. The task order was as follows:

Accelerometer

Single leg stance eyes open and eyes closed for 30 seconds – repeated 3 times

Single leg squat eyes open only – repeated 3 times

Hop forward eyes open only – repeated 3 times

Hop to the left eyes open only – repeated 3 times

Hop to the right eyes open only – repeated 3 times

Wobble Board

Single leg stance eyes open only for 30 seconds – repeated 3 times

Single leg squat eyes open only – repeated 3 times

Task Protocol

Participants were asked to assume a single leg balance position using the hand support. When the participant indicated they were happy to begin, they were instructed to release the hand support and the researcher would commence data collection via the associated software. At the end of the task the researcher would cease collection. Participants were given no balance advice other than to stand in the middle of the wobbleboard. No parameters were set regarding technique, allowing participants to assume their most comfortable and natural position, during task assessment. Participants were advised that only their testing limb was allowed to make contact with the environment during the task.

Practice and Failure of Tasks

Participants were allowed one practice attempt for all tasks, thus participants had four attempts at each task. Where task protocol wasn't successfully followed the attempt was deemed a failure. If a participant had two consecutive task failures, then that task was stopped and the participant would move onto the next task. Any successful attempts recorded before the task was stopped were still included in the final data analysis.

Standardisation of dynamic tasks

Single leg squat and single leg hop were standardised relating to participant height. Wobbleboard single leg squat was to a marked depth of 50% of participant height whilst accelerometry squat was to a depth of 50% of participant height minus 75mm (to account for difference between standing on wobbleboard and on floor) – an adjustable plinth was placed behind the participant and they were asked to single leg squat until they could feel the plinth then rise again without resting on it. Participants were required to maintain their balance for 2 seconds following completion of the squat.

Hop tasks were only completed using accelerometry measurement. Hop distance was normalised to participant height – forward hop was required to reach 50% of participant height, whilst sideways hops were required to reach 33% of participant height. A starting point and landing point were marked on the floor with regular electrical tape. A hop was deemed 'good' provided some part of the participant's foot touched the tape on landing. If a participant landed parallel with the tape but not touching the tape it was still considered a 'good' jump, however if the participant over or under-shot the tape distance this was considered a failed jump. Participants were required to maintain their balance for 2 seconds following completion of the hop.

Data Analysis

Demographic data (age, height, weight and shoe size) were normally distributed and are reported as mean, standard deviation (SD), range and standard error of mean (Table 1).

(n=17)	Age	Height (cm)	Weight (Kg)	Shoe Size
Mean	27.6	173.1	74.1	7.8
Std. Deviation	5.7	10.5	13.9	2.3
Range	19.0	34.0	57.0	7.0
Std. Error of Mean	1.4	2.5	3.4	0.5

Table 1. Demographic data: Mean, Standard Deviation, Range, Standard Error of Mean. (n=17)

Wobbleboard data were recorded via custom built software, for Windows OS, provided by THETAmetrix. This software produces a performance report by dividing the maximum tilt angle of the wobble board (15^{o}) into thirds and provides the percentage of time spent in each third as an output (Williams & Bentman 2014). Along with this, the software also provides the number of edge to floor contacts and percentage time spent with the edge in contact with the floor. As well as analysis of the individual bandings, a weighting was applied to each band (Inner x 4, Middle x 3, Outer x 2, Edge x 1), the inner banding time perceived as representing better balance than time spent on the edge. The weighting of each band allows for an overall value for each wobbleboard attempt to be collated.

The accelerometry software was also provided by THETAmetrix for Windows OS. This software converts the AP and ML linear accelerations of the pelvis into path length (Williams et al 2016). This is a calculation of the length of the sway path created by the AP and ML accelerations (Williams et al 2016). The length of the path between each sequential data point for AP acceleration (sample (x+1) - sample x) is determined & summed (Williams et al 2016) and then normalised to task time. This is repeated for ML acceleration; the normalised path length (NPL) is a combination the AP and ML path length and is measured in mg (m=milli, g=units of gravity) (Williams et al 2016). Only NPL was used because it was previously found to have the highest ICC and lowest MDC when compared with jerk and root mean squared (RMS) (Williams et al 2016).

Hop performance was measured using the trunk mounted accelerometer. Linear accelerations were corrected for sensor tilt using the on-board gyroscopes to provide linear accelerations in anteroposterior, mediolateral and superoinferior accelerations. These orthogonal accelerations were transferred to MatLab (Mathworks 2008b) and sway during hop landing was calculated using a bespoke algorithm. The landing impact peak was identified from the superoinferior acceleration and acceleration data was trimmed to 1 second following landing. Path length of the AP and ML acceleration data during this 1 second was quantified. These metrics were calculated for each hop trial and the mean of the three trials was determined.

SPSS Analysis

All data were tested for normality using a Shapiro-Wilks test. Of the 31 individual outcome variables obtained 8 were not normally distributed. However, as these 8 data sets did not meet the assumptions needed for parametric testing of not normally distributed data (sample size was ≤ 20), non-parametric testing was used throughout. Between-limb analysis was completed using Wilcoxon signed ranks test, between task analysis used Mann-Whitney U and correlations between tasks were analysed using Spearman's r coefficients. Bland-Altman plots were also created in Excel to review agreement between accelerometry and wobbleboard measurements of balance (Giavarina 2015).

Reliability analysis was completed using SPSS v23. All tasks were tested using a two-way mixed effects model for absolute agreement to establish Intraclass Correlation Coefficients (ICC). The average measures scores were recorded as each participant recorded six attempts at each task (3 dominant limb, 3 non-dominant limb). Standard Error of Measurement (SEM) was calculated using the formula: SEM = Standard Deviation (SD) x $\sqrt{(1\text{-ICC})}$ (Darter et al 2013). The SEM provides a measure of variability in the units of interest but is also important in calculating the minimum detectable change (MDC) (Darter et al 2013). The MDC allows assumptions to be made regarding changes in measurement; it can help determine whether a difference between two measurements represents a true change, or is otherwise a normal variation or measurement error (Darter et al 2013). The MDC was calculated using the formula: MDC_{95} = 1.96 x SEM x $\sqrt{2}$ (Haley & Fragala-Pinkham 2006).

4.Results

A large amount of data was collected as part of this study. The findings have been reported systematically to make collation and understanding more straightforward. Their association to the project aims will be highlighted at the end of the section.

Accelerometry

Accelerometry data were measured as the normalised path length (NPL). The NPL is the cumulation of the anteroposterior (AP) & mediolateral (ML) pathways. Illustrations of the NPL for single leg stand eyes open (SLSEO), single leg stand eyes closed (SLSEC) and single leg squat (SLSQ) tasks are included (Figures 1, 2 & 3).

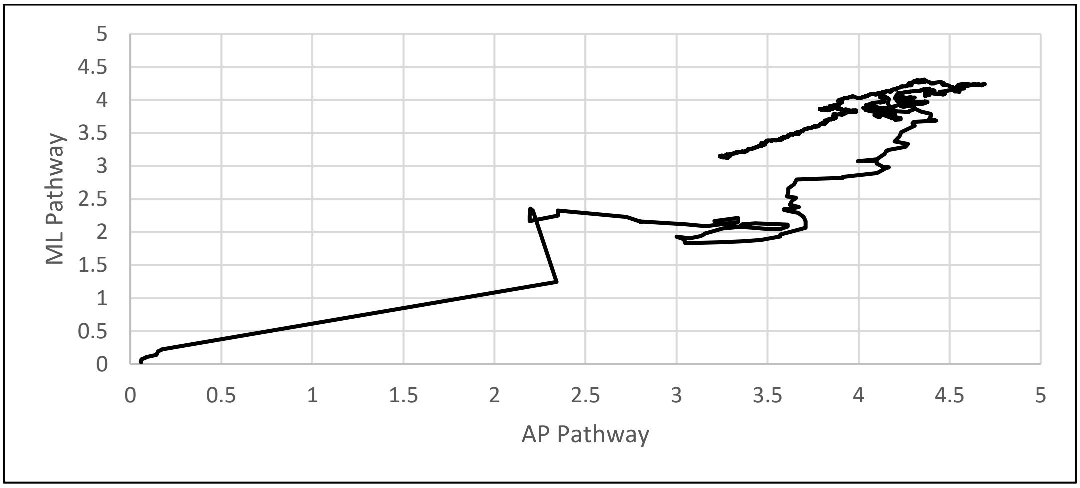


Figure 1. Anteroposterior (AP) v Mediolateral (ML) sway pathway for single leg stand eyes open (SLSEO).

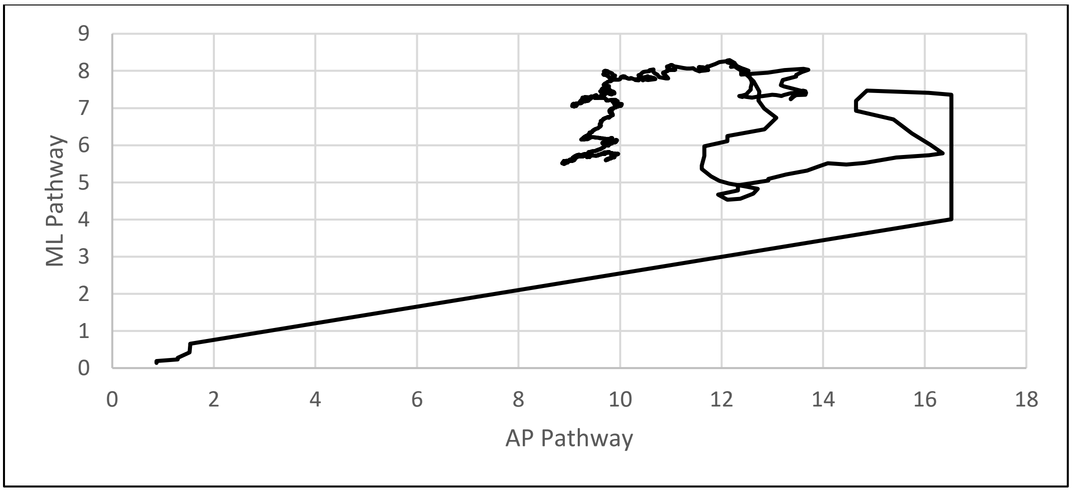


Figure 2. Anteroposterior (AP) v Mediolateral (ML) sway pathway for single leg stand eyes closed (SLSEC).

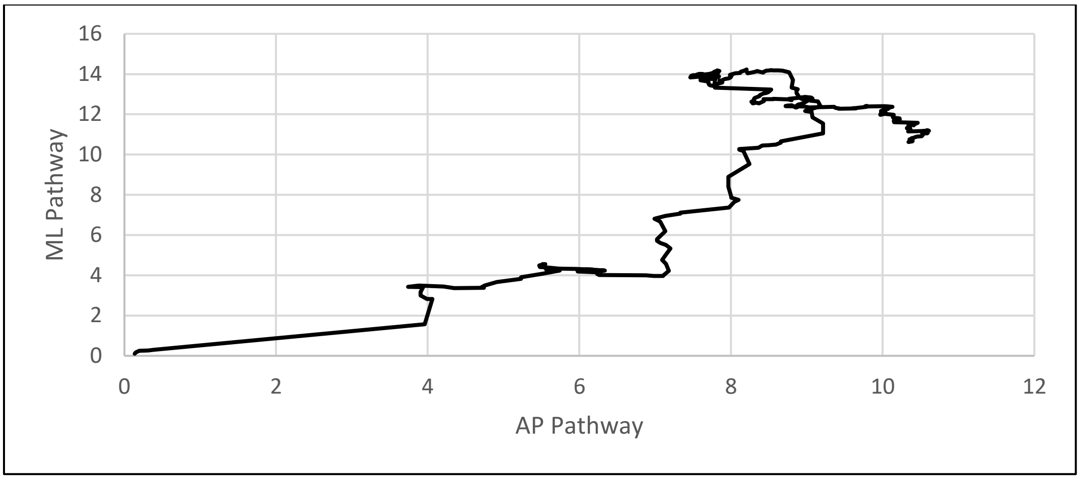


Figure 3. Anteroposterior (AP) v Mediolateral (ML) sway pathway for single leg squat (SLSQ).

As the data was not normally distributed, a Wilcoxon Signed Ranks (2-tailed significance) test was used to test for significance between dominant and non-dominant limbs, for their respective tasks. No significant differences were found between limb for any task. Figure 4 highlights the median normalised path length scores for dominant v non-dominant comparison for the balance sensor tasks. The median NPL, interquartile range (IRQ), percentage difference between dominant and non-dominant limbs and values for statistical significance are shown in Table 2.

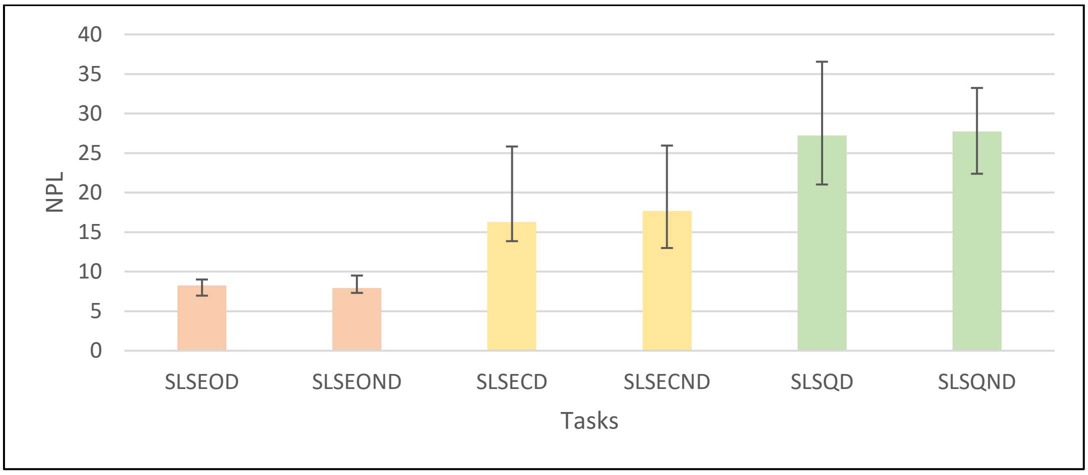


Figure 4. Median & Interquartile Range for task specific dominant v non-dominant comparisons. (NPL, Normalised Path Length; SLSEOD/ND, Single Limb Stance Dominant/Non-Dominant; SLSECD/ND, Single Limb Stance Eyes Closed Dominant/Non-Dominant; SLSQD/ND, Single Limb Squat Dominant/Non-Dominant)

	Dominant		Non-Dominant		Percentage Difference (%)	Wilcoxon Signed Ranks (2-tailed significance)
Accelerometry						
SLSEO	Median	8.3	Median	7.9	4.8	0.747
	IQR	2.3	IQR	2.4		
SLSEC	Median	16.3	Median	17.7	-8.6	0.893
	IQR	13.3	IQR	14.8		
SLSQ	Median	27.2	Median	27.7	-1.8	0.890
	IQR	17.8	IQR	14.5		

Table 2. Accelerometry median and interquartile range data for dominant and non-dominant limb for each task (SLSEO, Single Limb Stance Eyes Open; SLSEC, Single Limb Stance Eyes Closed; SLSQ, Single Limb Squat).

The percentage difference between dominant and non-dominant lower limbs suggested no pattern pertaining to the dominant limb performing better (as demonstrated by lower path length scores), as some differences were positive and some negative. Values for percentage difference ranged from -8.6% to 4.8% with a mean percentage difference of -1.9%±6.7%. However, by calculating the mean using both positive and negative values, a lower mean value is likely as the positive and negative values will cancel each other out. Therefore, the absolute mean difference was calculated to be 5.1%±3.4%. These results demonstrate that on average there is approximately a 5% difference in balance performance between the dominant and non-dominant limb. Also, there is no significant difference between limbs expected across this range of tasks.

The Intra-class Correlation Coefficients (ICC) suggest excellent (>0.90) test-retest reliability (Koo & Li 2016) for accelerometry when measuring single limb stance (eyes open & closed) and single limb squat (Table 3). Furthermore, the SEM values were low for SLSEO and SLSQ where normalised as a percentage represent <10%. Converting this to minimal detectable change demonstrates that with 95% confidence a change of >26% represents real change. Despite excellent ICC values SLSEC had a much greater SEM (<35%) and therefore subsequently larger MDC (>93%).

Task	ICC (95% CI)	SEM (mg)	SEM as % of task median	MDC_{95} (mg)	MDC as % of task median
SLSEO	0.925 (0.852 – 0.969)	0.8	10.0	2.1	26.3
SLSEC	0.955 (0.874 – 0.991)	5.8	33.5	16.2	93.6
SLSQ	0.947 (0.896 – 0.978)	2.6	9.5	7.2	26.4

Table 3. Test-retest reliability scores for accelerometry tasks (ICC, Intraclass Correlation Coefficient; SEM, Standard Error of Measurement; MDC, Minimal Detectable Change; SLSEO, Single Limb Stand Eyes Open; SLSEC, Single Limb Stand Eyes Closed; SLSQ, Single Limb Squat).

Instrumented Wobbleboard

The instrumented wobbleboard reports the percentage time participants spent in different bandings. The bandings are created by separating the total tilt angle into thirds and labelled as inner, middle and outer; time spent with the edge in contact with the floor is also recorded. Examples of the recorded wobbleboard output for single leg stand (WBSLS) and single leg squat (WBSLSQ) tasks are shown in figures 5 & 6. Plots are coded so green is within the inner third of the maximal tilt angle, yellow the middle third and red the outer third. A weighting was applied to each banding so that an overall score could be created for each task.

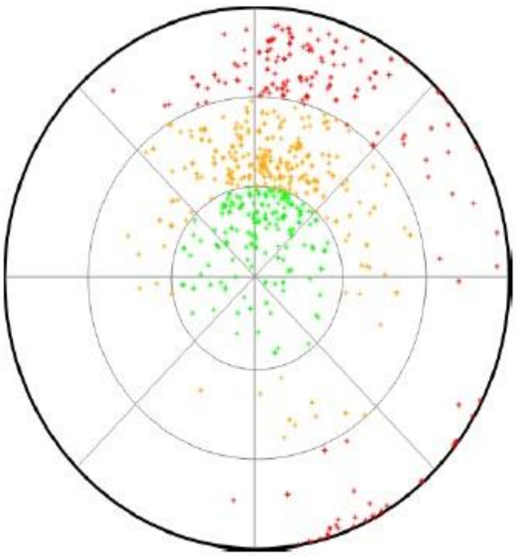


Figure 5. Feedback of tilt position for wobbleboard single limb stand.

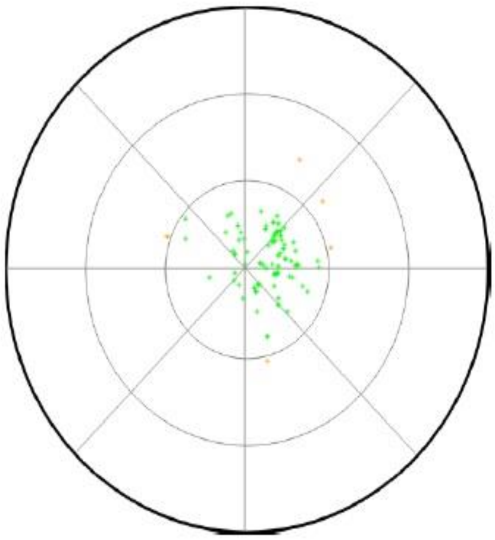


Figure 6. Feedback of tilt position for wobbleboard single limb squat.

As the data was not normally distributed, a Wilcoxon Signed Ranks (2-tailed significance) test was used to test for significance between dominant and non-dominant limbs, for their respective tasks. No significant differences were found between limb for any task. Figures 7 & 8 highlight the median percentage scores for dominant v non-dominant comparison for the wobbleboard tasks. The median score, interquartile range (IQR), percentage difference between dominant and non-dominant limbs and values for statistical significance are shown in Table 4.

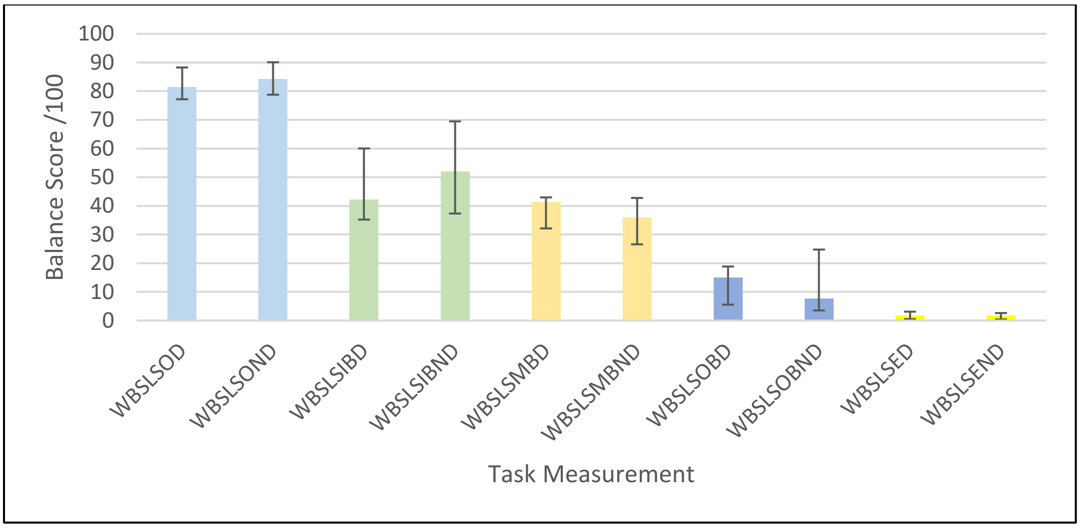


Figure 7. Wobbleboard single limb stand median and interquartile range dominant v non-dominant values compared (WBSLS, Wobbleboard Single Limb Stand; O, Overall; IB, Inner Band; MB, Middle Band; OB, Outer Band; E, Edge; D, Dominant; ND, Non-Dominant).

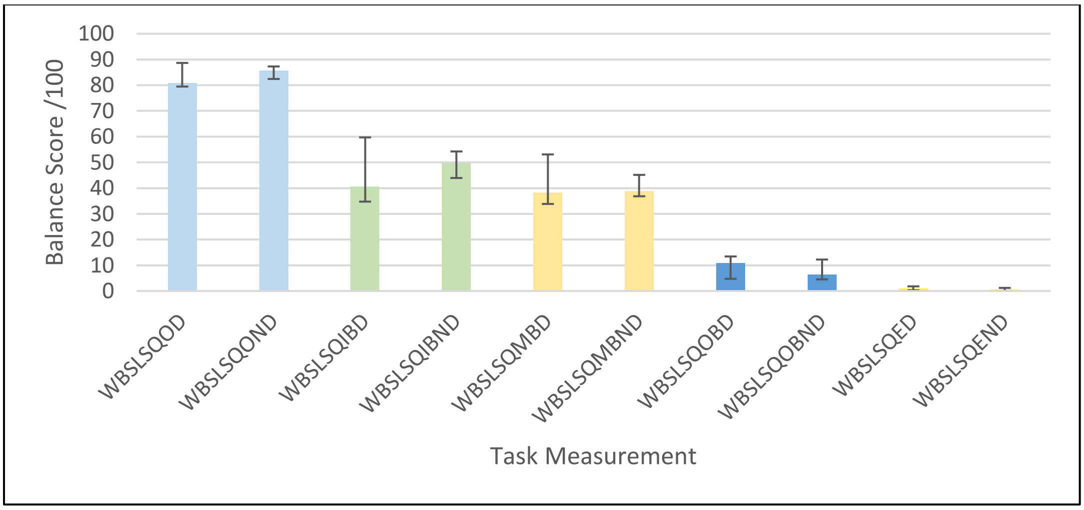


Figure 8. Wobbleboard single limb squat median and interquartile range dominant v non-dominant values compared (WBSLSQ, Wobbleboard Single Limb Squat; O, Overall; IB, Inner Band; MB, Middle Band; OB, Outer Band; E, Edge; D, Dominant; ND, Non-Dominant).

	Dominant		Non-Dominant		Percentage Difference (%)	Wilcoxon Signed Ranks (2-tailed significance)
Wobbleboard						
Single Leg Stand – Overall (WBSLSO)	Median	81.3	Median	84.1	-3.4	1.000
	IQR	12.0	IQR	16.4		
Single Leg Stand – Inner Band (WBSLSIB)	Median	42.2	Median	47.6	-12.8	0.907
	IQR	29.6	IQR	40.7		
Single Leg Stand – Middle Band (WBSLSMB)	Median	41.2	Median	35.9	12.9	0.305
	IQR	13.467	IQR	45.133		
Single Leg Stand – Outer Band (WBSLSOB)	Median	11.4	Median	8.0	29.8	1.000
	IQR	14.4	IQR	18.8		
Single Leg Stand – Edge (WBSLSE)	Median	1.8	Median	1.3	27.8	0.985
	IQR	3.0	IQR	2.3		
Single Leg Squat – Overall (WBSLSQO)	Median	81.9	Median	85.6	-4.5	0.243
	IQR	8.9	IQR	6.1		
Single Leg Squat – Inner Band (WBSLSQIB)	Median	40.5	Median	49.7	-22.7	0.263
	IQR	23.9	IQR	15.6		
Single Leg Squat – Middle Band (WBSLSQMB)	Median	38.7	Median	38.9	-0.5	0.459
	IQR	21.0	IQR	12.5		
Single Leg Squat – Outer Band (WBSLSQOB)	Median	10.1	Median	6.4	36.6	0.632
	IQR	11.2	IQR	7.4		
Single Leg Squat – Edge (WBSLSQE)	Median	0.9	Median	0.5	44.4	0.463
	IQR	2.2	IQR	1.8		

Table 4. Wobbleboard median and interquartile range data for dominant and non-dominant limb for each task (WBSLS, Wobbleboard Single Limb Stand; WBSLSQ, Wobbleboard Single Limb Squat; O, Overall; IB, Inner Band; MB, Middle Band; OB, Outer Band; E, Edge; D, Dominant; ND, Non-Dominant).

The percentage difference between dominant and non-dominant lower limbs suggested no pattern pertaining to the dominant limb performing better as some differences were positive and some negative. Values for percentage difference for the wobbleboard bandings ranged from -12.8 to 29.8% for single leg stance and -22.7 to 44.4% for single leg squat. The mean percentage difference is 14.4±24.3% whilst the absolute mean difference is 23.4%±14.3%. There is a grey area for wobbleboard scoring because of a variation in perceived performance i.e. increased time spent in the inner band is seen as positive, but in the outer band and edge is seen as negative. The wobbleboard software does not distinguish between bandings, other than to report time spent in each. For this reason, a weighting was added to each banding to create an overall score for each task. The values for percentage difference for the overall wobbleboard scores are -3.4% for single limb stance and -4.5% for single limb squat. The negative overall scores would suggest that the non-dominant limb should be expected to perform better with a mean difference of -4.0%±0.8%. However, no between limb significant difference was present for any wobbleboard measurement, for either task.

The ICC values for the weighted task scores suggest excellent (>0.90) test-retest reliability (Koo & Li 2016) for wobbleboard when measuring single limb stance and good (0.75-0.90) reliability (Koo & Li 2016) for single limb squat (Table 5). Furthermore, SEM values were low especially for the single leg stance task resulting in a minimal detectable change of <10% reflecting true detectable change in performance for wobbleboard single limb stance overall (WBSLSO) and <15% for wobbleboard single limb squat overall (WBSLSQO). The ICC, SEM and MDC values were reviewed for each banding (IB, Inner; MB Middle; OB, Outer; E, Edge) for each task (Table 6). For single limb stance the weighted result ICC was greater than the ICC for MB, OB & E, whilst for squat the weighted result ICC was greater than the ICC for OB & E. The SEM and MDC values were also generally much smaller for the overall scores than the individual bandings. As the weighted result proved more reliable than some of the individual bandings and accounted for findings across all bandings, it was deemed to be an appropriate method for scoring the wobbleboard tasks.

Task	ICC (95% CI)	SEM	SEM as % of task median	MDC_{95}	MDC as % of task median
WBSLSO	0.918 (0.730 – 0.990)	2.8	3.3	7.8	9.3
WBSLSQO	0.772 (0.513 – 0.919)	4.2	5.0	11.7	14.0

Table 5. Test-retest reliability scores for overall wobbleboard tasks (ICC, Intraclass Correlation Coefficient; SEM, Standard Error of Measurement; MDC, Minimal Detectable Change; WBSLSO, Wobbleboard Single Limb Stand Overall; WBSLSQO, Wobbleboard Single Limb Squat Overall).

Task	ICC	SEM	SEM as % of task median	MDC_{95}	MDC as % of task median
WBSLSIB	0.928	6.2	13.2	17.2	36.6
WBSLSMB	0.905	4.2	10.2	11.6	28.3
WBSLSOB	0.886	3.5	42.2	9.7	116.9
WBSLSE	0.534	3.5	233.3	9.7	646.7
WBSLSQIB	0.783	10.1	20.7	28.0	57.4
WBSLSQMB	0.804	7.0	18.0	19.4	50.0
WBSLSQOB	0.661	6.3	82.9	17.5	230.3
WBSLSQE	0.760	2.1	420.0	5.8	1160.0

Table 6. Test-retest reliability scores for wobbleboard tasks (ICC, Intraclass Correlation Coefficient; SEM, Standard Error of Measurement; MDC, Minimal Detectable Change; WBSLSIB/MB/OB/E, Wobbleboard Single Limb Stand Inner Band/Middle Band/Outer Band/Edge; WBSLSIB/MB/OB/E, Wobbleboard Single Limb Squat Inner Band/Middle Band/Outer Band/Edge).

Hop Landing

Hop landings were measured using accelerometry pathways for the first second post landing. The hop landing was recognised using the superoinferior acceleration data (Figure 9). Figures 10, 11, 12 show examples of the 1-second landing path length for hop forward (HF), hop lateral (HL) and hop medial (HM) tasks respectively.

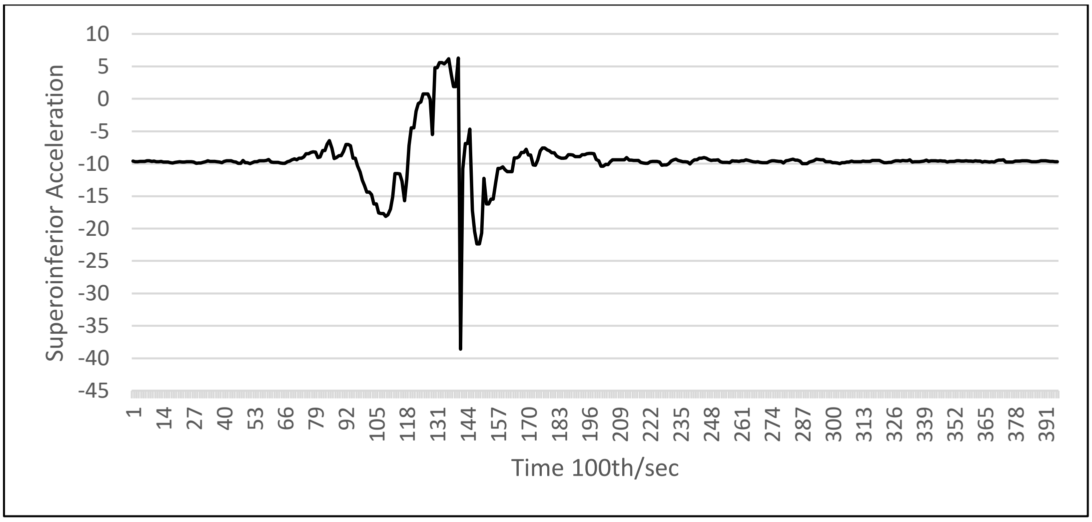


Figure 9. Superoinferior acceleration for a typical hop task

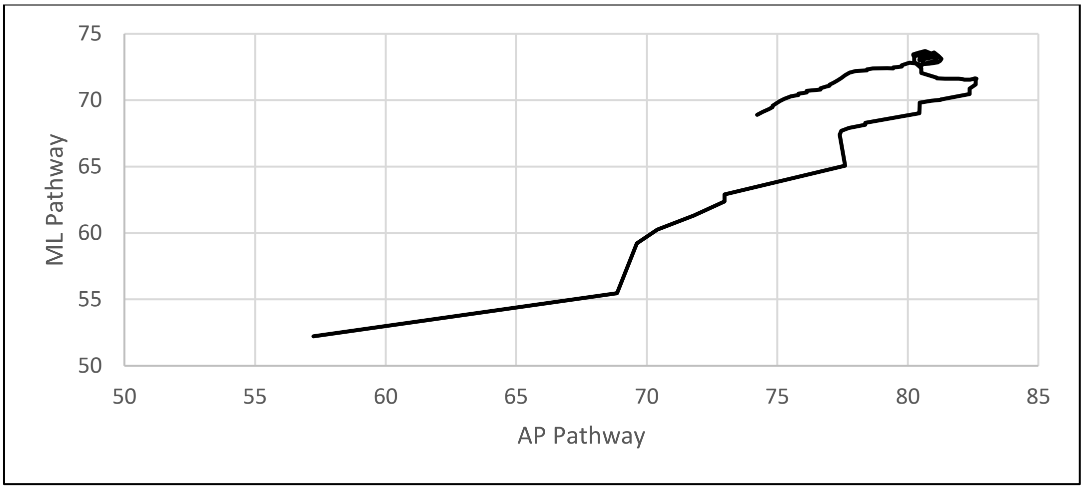


Figure 10. Anteroposterior pathway v Mediolateral pathway for Hop Forward.

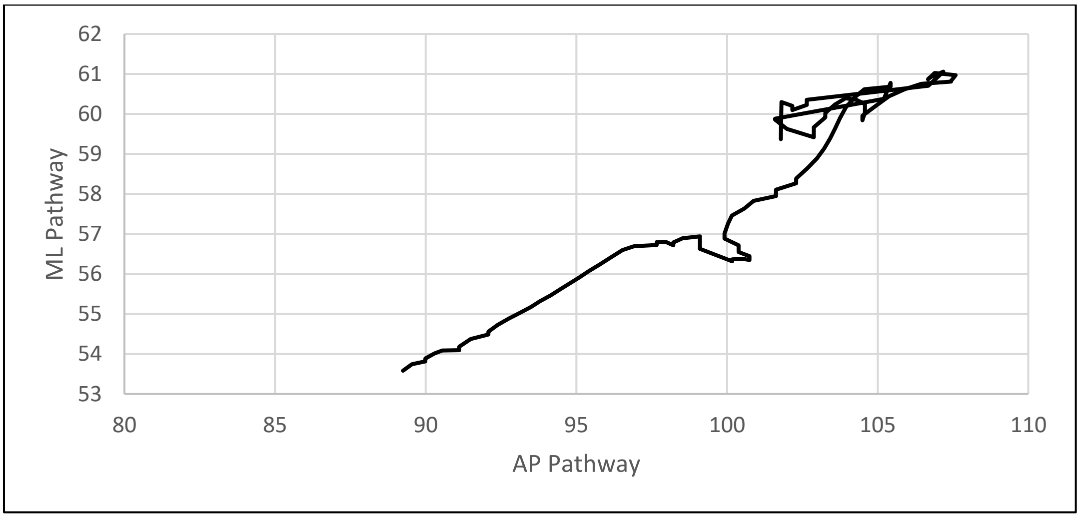


Figure 11. Anteroposterior pathway v Mediolateral pathway for Hop Lateral.

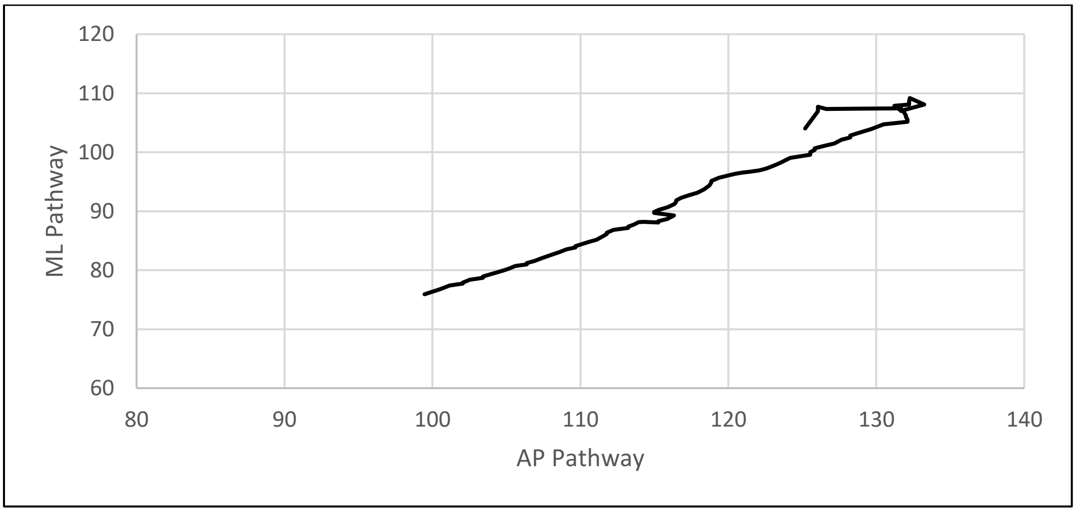


Figure 12. Anteroposterior pathway v Mediolateral pathway for Hop Medial.

As the data were not normally distributed, a Wilcoxon Signed Ranks (2-tailed significance) test was used to test for significance between dominant and non-dominant limbs, for their respective tasks. No significant differences were found between limb for any task. Figure 13 highlights the median normalised path length scores for dominant v non-dominant comparison for the balance sensor hop landing tasks. The median score, interquartile range (IQR), percentage difference between dominant and non-dominant limbs and values for statistical significance are shown in Table 7.

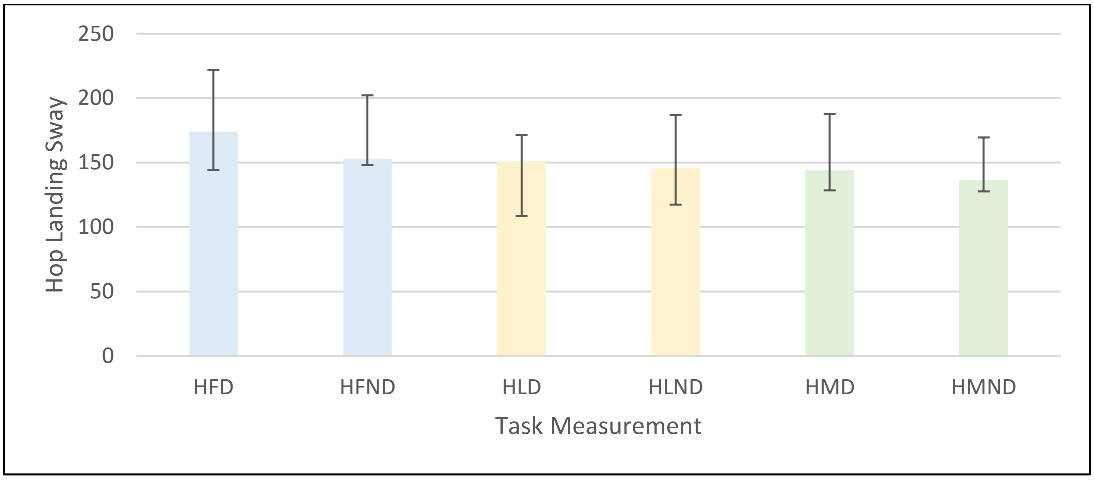


Figure 13. Hop landing median and interquartile range for dominant v non-dominant values compared (HFD/ND, Hop Forward Dominant/Non-Dominant; HLD/ND, Hop Lateral Dominant/Non-Dominant; HMD/ND, Hop Medial Dominant/Non-Dominant).

	Dominant		**Non-Dominant**		**Percentage Difference (%)**	**Wilcoxon Signed Ranks (2-tailed significance)**
Accelerometry						
Hop Forward (HF)	Median	174.1	Median	153.1	12.1	1.000
	IQR	74.2	IQR	52.0		
Hop Lateral (HL)	Median	144.7	Median	146.0	-0.9	0.348
	IQR	60.2	IQR	75.6		
Hop Medial (HM)	Median	140.5	Median	135.9	3.3	0.528
	IQR	56.2	IQR	47.3		

Table 7. Hop landing median and interquartile range data for dominant v non-dominant limb for each task (HF, Hop Forward; HL, Hop Lateral; HM, Hop Medial).

The percentage difference between dominant and non-dominant lower limbs range from -0.9% to 12.1% when compared to the dominant limb. The mean percentage difference is 4.8%±6.6%. The absolute mean percentage difference was calculated as 5.4%±5.9%. The mixture of positive and negative scores makes it difficult to be sure whether the dominant or non-dominant limb should be expected to perform better. As only the hop lateral score is negative and by a relatively small amount when compared with the other hop tasks, it may be appropriate to consider the non-dominant limb to consistently perform better. Although, no limb should be expected to perform significantly better, for any task.

The ICC values for the tasks suggest excellent (>0.90) test-retest reliability (Koo & Li 2016) for balance sensor hop landing when measuring hop medial and moderate (0.5 – 0.75) reliability (Koo & Li 2016) for both hop forward and hop lateral (Table 8). Furthermore, SEM values were low for medial hop landing, where normalised as a percentage represent <16%. Converting this to minimal detectable change demonstrates that with 95% confidence a change of >31% represents real change. Despite moderate ICC values, hop landing lateral and forward had a much greater SEM and therefore subsequently larger MDC, suggesting greater task variability.

Task	ICC (95% CI)	SEM (mg)	SEM as % of task median	MDC_{95} (mg)	MDC as % of task median
HF	0.734 (0.406 – 0.912)	42.0	24.3	116.4	67.2
HL	0.737 (0.395 – 0.923)	31.4	21.5	86.9	59.5
HM	0.960 (0.895 – 0.991)	15.4	11.3	42.7	31.3

Table 8. Test-retest reliability scores for hop landing tasks (HF, Hop Forward; HL, Hop Lateral; HM, Hop Medial).

Task Difficulty

To ensure that limb symmetry was being measured across a variety of tasks, it was important to review the different task's relationship to each other. The values for task specific cumulative (dominant and non-dominant results) medians (accelerometry path lengths, wobbleboard balance scores and hop landing sway scores), IQR's, percentage differences between tasks and Mann-Whitney U tests for statistical significance can be seen in table 9.

Tasks	Task 1		Task 2		Percentage difference between tasks (%)	Task 1 as % of Task 2	Mann-Whitney U (2-tailed significance)
Accelerometry							
SLSEO v SLSEC	Median	8.0	Median	17.3	-116.3	46.2	0.000
	IQR	2.4	IQR	13.6			
SLSEO v SLSQ	Median	8.0	Median	27.3	-241.3	29.3	0.000
	IQR	2.4	IQR	16.1			
SLSEC v SLSQ	Median	17.3	Median	27.3	-57.8	63.4	0.002
	IQR	13.6	IQR	16.1			
HF v HL	Median	173.1	Median	146.0	15.7	118.6	0.008
	IQR	61.3	IQR	63.8			
HF v HM	Median	173.1	Median	136.5	21.1	126.8	0.004
	IQR	61.3	IQR	44.5			
HL v HM	Median	146.0	Median	136.5	6.5	107.0	0.962
	IQR	63.8	IQR	44.5			
Wobbleboard							
WBSLSO v WBSLSQO	Median	84.1	Median	83.5	0.652	100.7	0.750
	IQR	12.4	IQR	7.3			
WBSLSIB v WBSLSQIB	Median	47.0	Median	48.8	-3.938	96.3	0.923
	IQR	30.4	IQR	20.8			
WBSLSMB v WBSLSQMB	Median	41.0	Median	38.8	5.331	105.7	0.273
	IQR	15.6	IQR	14.3			
WBSLSOB v WBSLSQOB	Median	8.3	Median	7.6	9.105	109.2	0.360
	IQR	14.7	IQR	9.5			
WBSLSE v WBSLSQE	Median	1.5	Median	0.5	66.472	300.0	0.058
	IQR	2.4	IQR	2.0			

Table 9. Cumulative task scores for each task highlighting task difficulty (SLSEO, Single Limb Stance Eyes Open; SLSEC, Single Limb Stance Eyes Closed; SLSQ, Single Limb Squat; HF, Hop Forward; HL, Hop Lateral; HM, Hop Medial; WBSLS, Wobbleboard Single Limb Stance; WBSLSQ, Wobbleboard Single Limb Squat; O, Overall; IB, Inner Band; MB, Middle Band; OB, Outer Band; E, Edge).

Statistically significant differences ($p<0.05$) were found between all tasks, except between lateral and medial hop, and between wobbleboard single leg stand and wobbleboard squat. The increase in sway area, from quiet standing to squatting, supports the theory of increasing difficulty across these tasks. The significant difference between HF and HL/HM suggests the different planes of hop landing represent different balance challenges. The lack of statistical significance during wobble board tasks suggests that these wobbleboard challenges do not differ in terms of difficulty. Figures 14, 15 & 16 highlight between-task comparisons for their respective measurement devices.

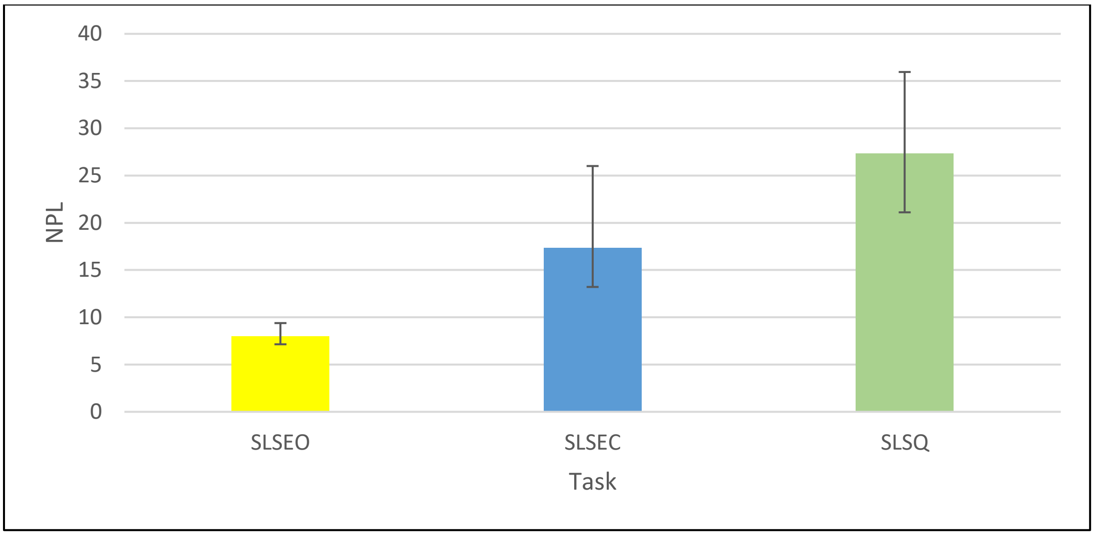


Figure 14. Median and interquartile range for accelerometry between task comparison.

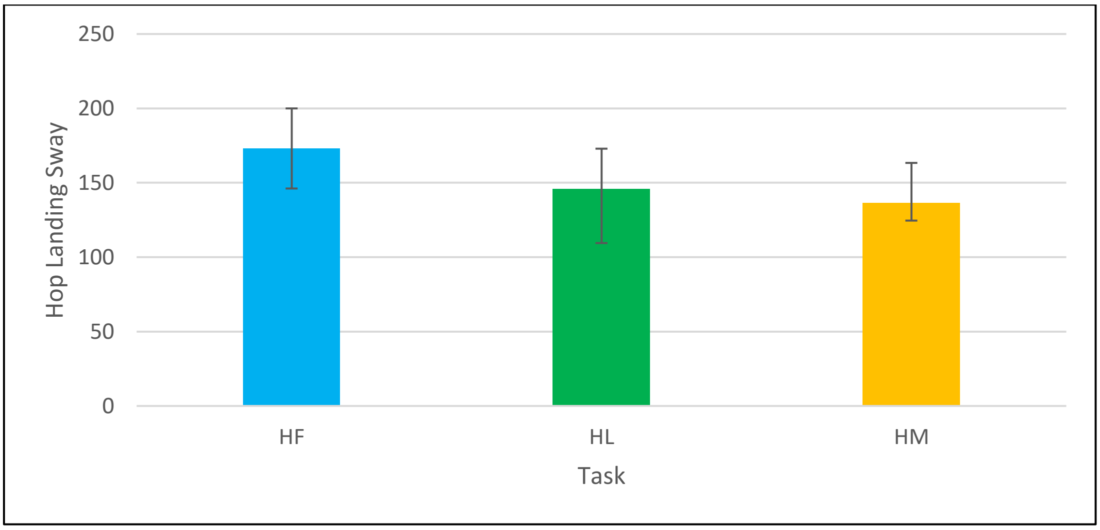


Figure 15. Median and interquartile range for hop landing between task comparison.

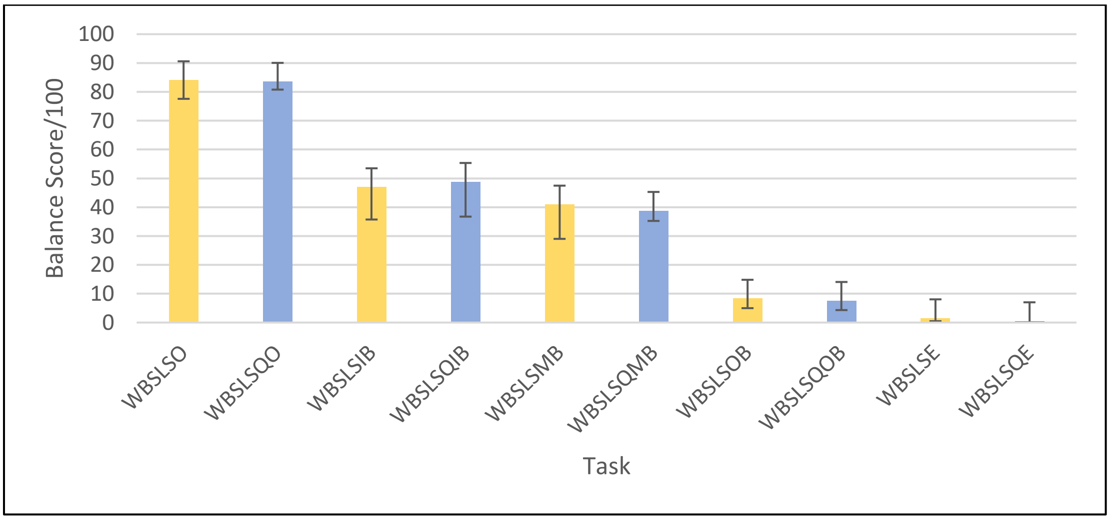


Figure 16. Median and interquartile range for wobbleboard between task comparison.

The accelerometry data demonstrates that the single leg stand eyes open is the easiest task to perform; with single leg stand eyes closed being approximately 2.2 times harder to perform. The single leg squat is approximately 3.4 times harder than SLSEO and 1.6 times harder than SLSEC. The hop forward is the hardest of the hop tests being approximately 1.2 times harder than the HL and 1.3 times harder than the HM. The hop lateral and hop medial provide similar levels of difficulty to each other with no significant difference between tasks. The wobbleboard tasks show no significant difference in difficulty at any level.

Task Correlations

A total of twenty-eight task correlations were explored for 2-tailed significance using Spearman's non-parametric correlation coefficients. Six were found to be significantly correlated. Table 10 shows the significantly correlated tasks along with their r, r^2 and p values.

Task	**Spearman's Coefficient of Correlation (r) (2-tailed)**	**Coefficient of Determination (r^2)**	**% Variance**	**Statistical Significance (p)**
WBSLSO v SLSQ	-0.675	0.456	46	0.000
SLSEO v SLSEC	0.665	0.442	44	0.000
SLSEC v HF	-0.531	0.282	28	0.004
HL v HM	0.479	0.229	23	0.006
SLSQ v HM	-0.379	0.144	14	0.030
HF v HM	0.359	0.129	13	0.040

Table 10. Significant 2- tailed correlations between tasks (WBSLSO, Wobbleboard Single Limb Stance Overall; SLSQ, Single Limb Squat; SLSEO, Single Limb Stance Eyes Open; SLSEC, Single Limb Stance Eyes Closed; HF, Hop Forward; HM, Hop Medial).

The strengths of correlations were determined using ≤0.35= weak/low, 0.36-0.67=moderate and >0.67=strong (Taylor 1990 cited by Clark et al 2016). The r^2 value allows for an assumption regarding the level of variance in task two that can be accounted for by task one. For example, the WBSLSO is related to SLSQ with 46% of the variance in SLSQ explained by wobbleboard performance. The only correlation between wobbleboard performance and balance performance was for WBSLSO and SLSQ. No other correlations were significant between wobbleboard and accelerometer scores.

Bland-Altman Plots

To further review the use of the wobbleboard as a balance measurement device, Bland-Altman plots were created against the accelerometer. Bland-Altman plots are a method for reviewing the level of agreement between measurement devices when measuring the same variable (Giavarina 2015). As the accelerometer effectively measures a person's deficit from perfect balance i.e. postural sway of 0, the wobbleboard deficit was calculated for comparison i.e. 100% minus participant score, thus allowing for an absolute agreement difference of 0. Bland-Altman plots were created for SLSEO v WBSLSO, SLSQ v WBSLSQO and SLSQ v WBSLSO. The SLSQ v WBSLSO was created because that was the only significant correlation found between accelerometry and wobbleboard scores. Figures 17, 18, & 19 show the Bland-Altman plots for SLSEO v WBSLSO, SLSQ v WBSLSQO & SLSQ v WBSLSO respectively.

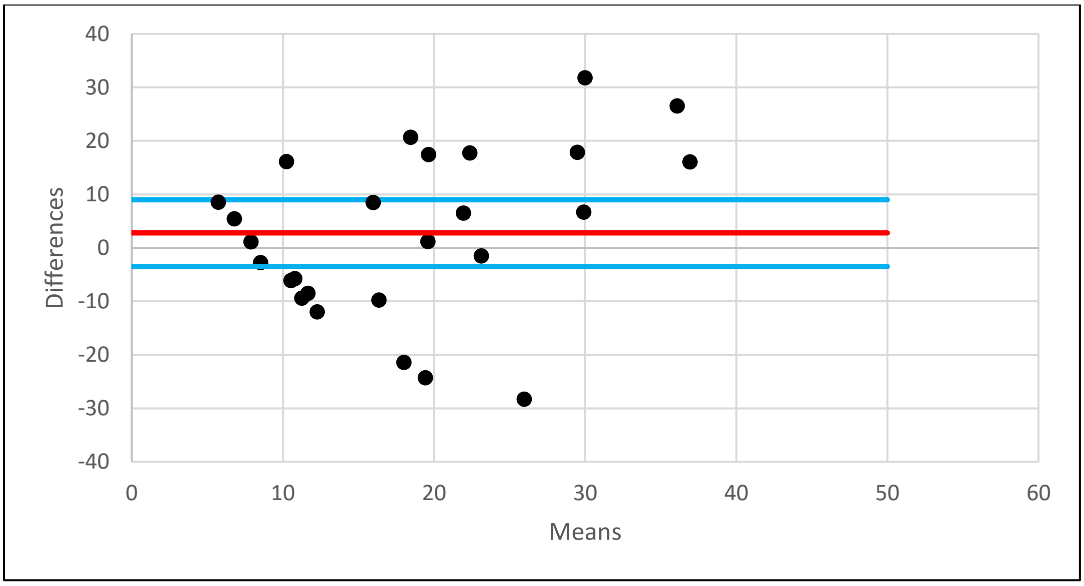


Figure 17. Bland-Altman plot comparing accelerometry single leg stand score with wobbleboard single leg stand overall score.

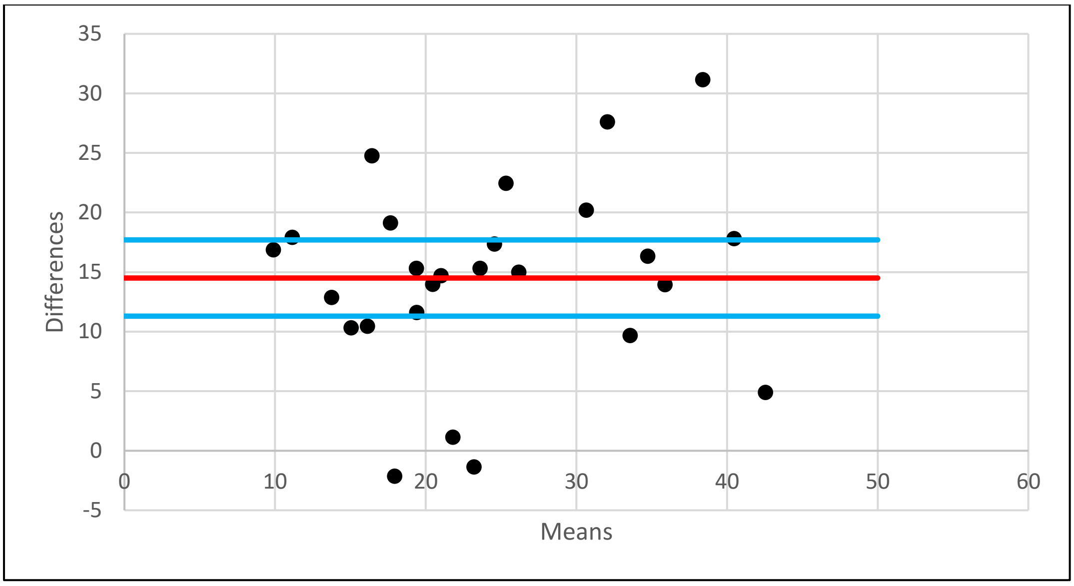


Figure 18. Bland-Altman plot comparing accelerometry single leg squat score with wobbleboard single leg squat overall score.

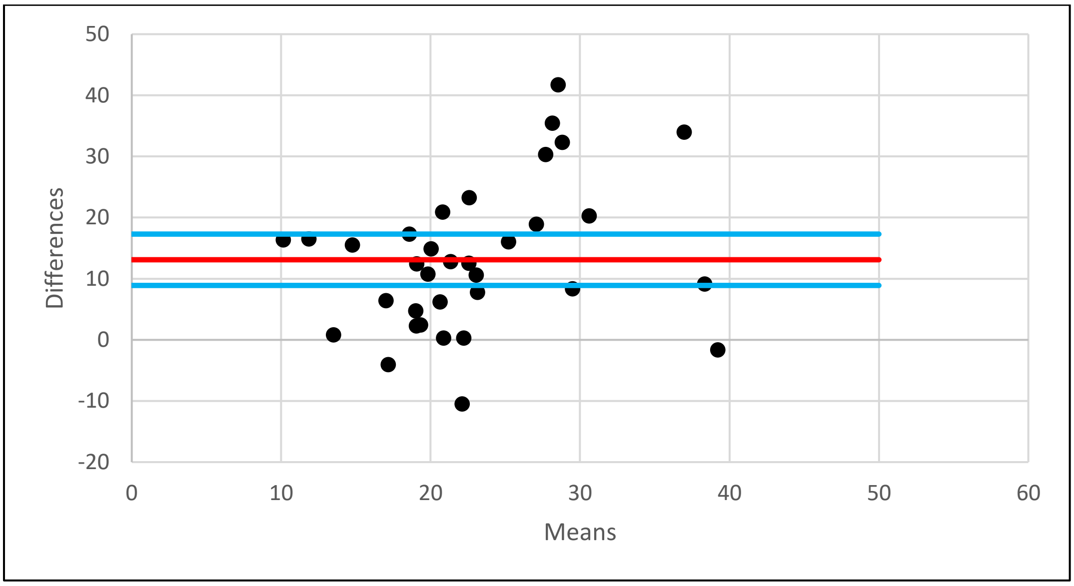


Figure 19. Bland-Altman plot comparing accelerometry single leg squat score with wobbleboard single leg stand overall score

The red line shows the mean value of the difference whilst the blue lines highlight the 95% confidence interval (used as the agreement value). Across all the plots, multiple points lie outside the confidence interval, suggesting that there is low agreement between the two balance measurement devices.

Functional Difficulties Questionnaire (FDQ-9)

All participants completed the FDQ-9 prior to balance assessment. The FDQ-9 scores ranged from 12-20 points. Figure 20 highlights the FDQ-9 scores.

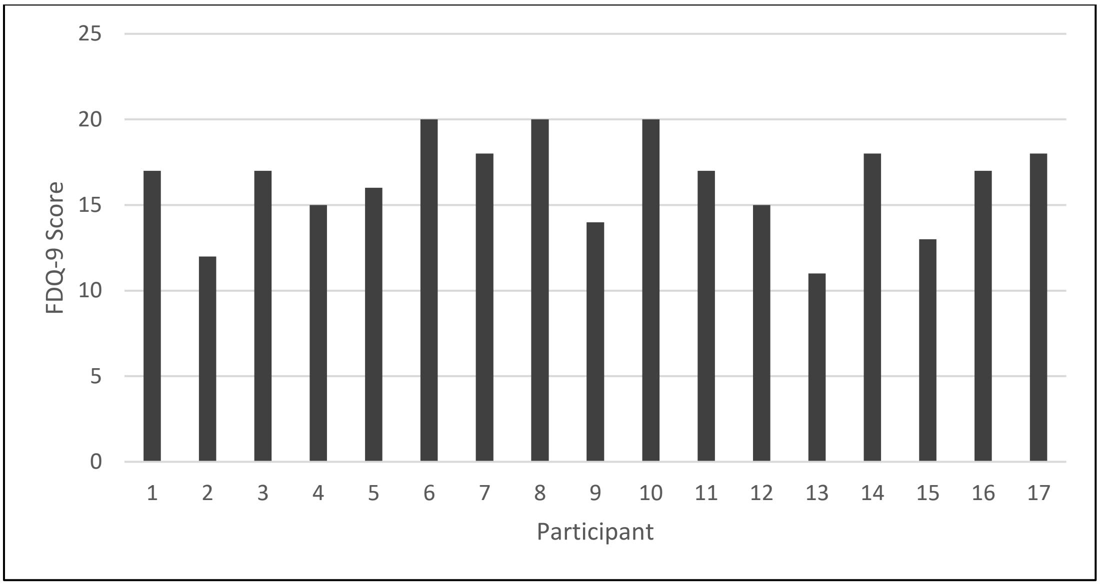


Figure 20. Functional Difficulties Questionnaire (FDQ-9) scores per participant.

Clark et al (2013) suggested a score of >21 was representative of functional difficulties. The maximum score for participants of this study was 20 suggesting that none of the participants demonstrated functional difficulties.

A total of eight tasks were tested for significant 1-tailed correlations against the FDQ-9 score. Two significant correlations were recorded and are shown in Table 11.

Task	**Spearman's Coefficient of Correlation (r) (1-tailed)**	**Coefficient of Determination (r^2)**	**% Variance**	**Statistical Significance (p)**
FDQ-9 v SLSQ	0.340	0.116	12	0.024
FDQ-9 v HF	0.308	0.095	10	0.038

Table 11. Significant 1-tailed correlations between FDQ-9 and balance tasks (FDQ-9, Functional Difficulties Questionnaire 9; SLSQ, Single Limb Squat; HF, Hop Forward).

The demographic data obtained as part of the study (self-reported by participants as part of the FDQ-9 questionnaire) was also reviewed. A total of twenty-four 1-tailed correlations were reviewed for significance for participant age, height and weight respectively. Five significant correlations were found for age (3), height (1) & weight (1) and can be seen in tables 12, 13 & 14 respectively.

Task	Spearman's Coefficient of Correlation (r) (1-tailed)	Coefficient of Determination (r^2)	% Variance	Statistical Significance (p)
Age v HL	0.570	0.325	33	0.000
Age v HF	0.447	0.200	20	0.004
Age v HM	0.379	0.144	14	0.015

Table 12. Significant 1-tailed correlations between age and balance tasks (HF, Hop Forward; HL, Hop Lateral; HM, Hop). Medial

Task	Spearman's Coefficient of Correlation (r) (1-tailed)	Coefficient of Determination (r^2)	% Variance	Statistical Significance (p)
Height v HM	-0.330	0.109	11	0.030

Table 13. Significant 1-tailed correlations between height and balance tasks (HM, Hop Medial).

Task	Spearman's Coefficient of Correlation (r) (1-tailed)	Coefficient of Determination (r^2)	% Variance	Statistical Significance (p)
Weight v SLSQ	-0.293	0.086	9	0.047

Table 14. Significant 1-tailed correlations between weight and balance tasks (SLSQ, Single Limb Squat).

The demographic correlations all relate to the balance sensor data, where lower scores represent better performance. Therefore, the positive correlations between age and hop landing suggest that the older a participant the worse their hop landing ability; whilst the negative correlations between height & HM and weight & SLSQ would infer that increased height and weight lead to better balance scores.

<u>Sample Size</u>

A post-hoc sample size calculation was carried out using a mean percentage difference between dominant and non-dominant limb of 4.9%±3.7%. The suggested sample size given an α of 0.05 with 80% power was calculated to be 4. Further review with an α of 0.01 with 99% power suggested a sample size of 12. This would suggest that this study was appropriately powered despite failing to reach the primary sample size of 30.

Summary of results pertaining specifically to study aims

Dominant v Non-Dominant Limb Symmetry

No statistically significant differences were identified across any of the balance tasks. The percentage differences were calculated as the difference between dominant and non-dominant limb, against the dominant limb. The mean percentage difference score across all tasks (balance sensor scores, wobbleboard overall scores, hop landing scores) was 0.1%±6.4% (95% CI -5.3% - 5.5%). The absolute mean percentage (ignoring positive and negative figures) was 4.9%±3.7% (95% CI 1.8% - 8.0%). The suggestion being that the LSI for balance, lies at approximately 5% between limb difference, with no specification on which limb should be expected to perform better.

Comparison of accelerometry with the wobbleboard

Bland-Altman plots suggested a low level of agreement between wobbleboard and accelerometry. This was supported by only one significant correlation being found between wobbleboard and accelerometry. However, the symmetry scores from the wobbleboard tasks are similar to those from the accelerometer; suggesting that the wobbleboard is a valid limb symmetry measurement device but perhaps measures a different variable to accelerometry.

Reliability of accelerometry in hop measurement

Accelerometry demonstrated moderate to excellent (0.73-0.96) test-retest reliability. This would suggest that accelerometry provides a reliable measurement device for measuring hop landing ability. The limb symmetry scores for hop landing were similar to the accelerometry and wobbleboard tasks.

FDQ-9

Only two weak significant correlations were highlighted between the FDQ-9 and accelerometry. The significant correlations were found for two of the more complex balance tasks. No participants in the study met the criterion for functional difficulties, suggesting that the FDQ-9 may be a suitable subjective method for predicting balance dysfunction even in individuals who lack obvious functional disorders.

5.Discussion

This study aimed to enhance the understanding of limb symmetry index (LSI) for balance by expanding the test parameters past the most popularly explored single limb stance. The study employed sacral mounted accelerometry and instrumented wobbleboard technology to review clinically accessible, validated balance measurement devices; whilst challenging single limb balance in increasingly complex balance challenges. To this extent this study also examined the efficacy of using trunk mounted accelerometry to measure single limb hop landing ability. The accelerometry tasks were compared with the wobbleboard tasks for concurrent validity; whilst all the tasks were also reviewed against a previously validated subjective predictor of balance (FDQ-9), to further review the ability of subjective measurements to predict balance dysfunction.

A total of eight tasks were completed for dominant v non-dominant limb comparison. Single limb stance was reviewed for both devices to review similarity to previous evidence and provide a recognised baseline for comparing the range of tasks. Single limb squat was tested as it represents a peculiar balance challenge. The base of support is stationary, but the centre of gravity is deliberately moved distal to the starting position in the transverse plane, before it is returned to the starting position. Therefore, as a task it combines elements of both static and dynamic balance by oscillating between the two definitions. The study used accelerometry to measure single limb hop in three different directions (HF, hop forward; HM, hop medial; HL, hop lateral) and test the reliability of using such novel methodology. Tasks were compared with each other to ensure a range of balance challenges were being measured; significant differences were found between all tasks, except wobbleboard single limb stand and squat and between hop lateral and hop medial, suggesting that for the most part a range of different tasks had been explored.

Limb Symmetry Index for Balance

No significant differences were found between limb for any task suggesting that limb symmetry exists irrespective of task. The overall value for LSI was established using the absolute mean of all the challenges. The LSI was determined to be 4.9%±3.7%, suggesting that the expected difference between dominant and non-dominant limb should be approximately 5%, with no specification as to which limb is expected to perform better. The overall LSI absolute mean value, falls in line with the previously established mean single leg stand LSI value for adults, from the associated literature review (7.7%±17.5%) (Figure 21). The overall LSI value, however, extends the understanding of LSI by including a variety of tasks.

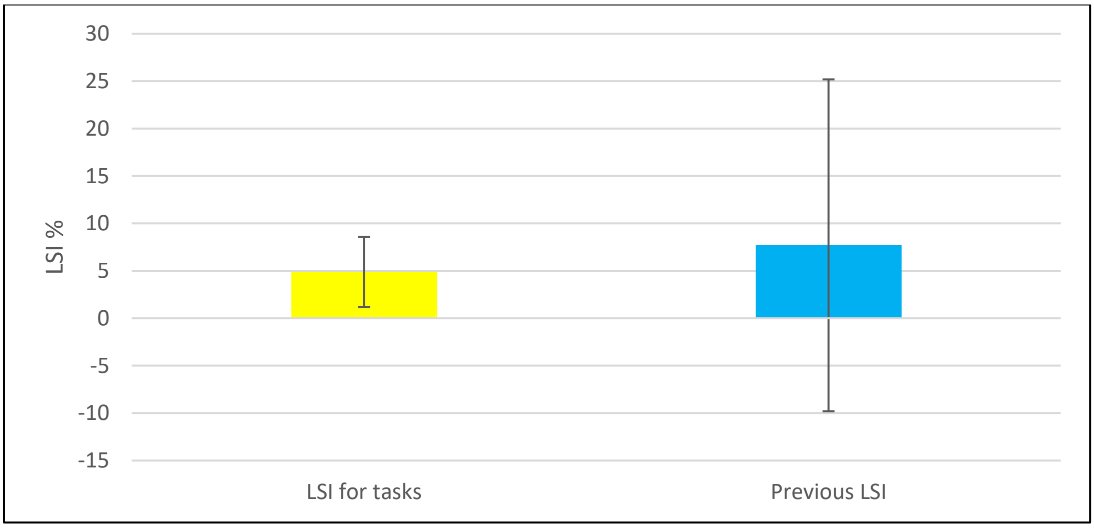


Figure 21. Limb symmetry index (LSI) for balance across tasks compared with previously established LSI.

The single limb balance LSI for accelerometry was 4.8% which is not dissimilar to the same test score for accelerometry found by Eguchi & Takada (2014) of 5.6%. This suggests that the findings of this study are in agreement with previous literature and strengthens the use of accelerometry as a reliable balance assessment tool. The lack of significant differences between limbs was also highlighted in previous literature (Chew-Bullock et al 2012, Clifford & Holder-Powell 2010, Cug et al 2014, Eguchi & Takada 2014, Johnson & Leck 2010, Kilroy et al 2016, Masu et al 2014, Matsuda et al 2008, Teranishi et al 2011). The findings of this study support the lack of significant difference between limbs; whilst also suggesting that task difficulty has no bearing on LSI for balance.

The explanation for the lack of significant difference likely lies within the intrinsic link between balance and normal everyday function. Normal ambulation involves reciprocity of gait so that an individual can transfer and support their weight from one leg to the other (Uustal & Baerga 2004). Eguchi & Takada (2014) demonstrated age related increases in postural stability with adults demonstrating the best performance in single limb stand; whilst the accompanying literature review also highlighted a significant difference in balance performance between over and under 18's. Pau et al (2012) did find a significant difference between lower limbs during single limb stance testing in participants with a mean age of 13. Perhaps as a person enters adolescence, fundamental body changes may affect single limb balance ability which are corrected by adulthood. Shim (2015) highlighted the rapid increases in growth as a person enters puberty accompanied by the commencement of epiphyseal plate fusion. These sudden changes in an individual's biomechanics perhaps resulting in a fluctuating centre of gravity, which combined with changes in joint proprioception, provide an ever-evolving challenge to their balance strategy. Ates (2017) investigated the effect of training on balance in participants aged 12-14 and found significant differences in performance between the trained and untrained groups. The idea that training improves balance is not new, but strengthens the theory that balance improves with age, because of composite functional repetition. By adulthood, the expectation should be that normal healthy individuals will have extensively practiced a large variety of functional tasks, to the point where limb asymmetry is minimised. This study also noted significant correlations between increasing age and poorer hop landing performance. Although no significant differences were found between limbs, it may suggest that age related changes may affect the more complex balance tasks first. From this it may be possible to infer that balance is normally distributed with age, where age represents time spent practicing balance, before age related physical changes increase the difficulty of balance and reduce the individual's skill. However, to confirm what normal age-related balance performance targets are, further research would need to be done to test different age groups, with these different tasks.

Comparison of Accelerometry with Wobbleboard

Accelerometry was used to measure single limb stand eyes open (SLSEO), single limb stand eyes closed (SLSEC) and single limb squat (SLSQ). All were found to have excellent reliability. The percentage between limb difference was found to be 4.8%, -8.6% and -1.8% respectively. The absolute mean for these tasks was calculated to be 5.1%±3.4%. The wobbleboard was also used to measure single limb stand (WBSLSO) and single limb squat (WBSLSQO), where the weighted scores of all the tilt angles were used to provide an overall score. The wobbleboard tasks demonstrated good to excellent test-retest reliability. The percentage between limb difference was found to be -3.4% (WBSLSO) and -4.5%(WBSLSQO); giving an absolute mean difference of 4.0%±0.8%. Comparing the wobbleboard tasks with accelerometry highlights similar findings for between limb symmetry (Figure 22).

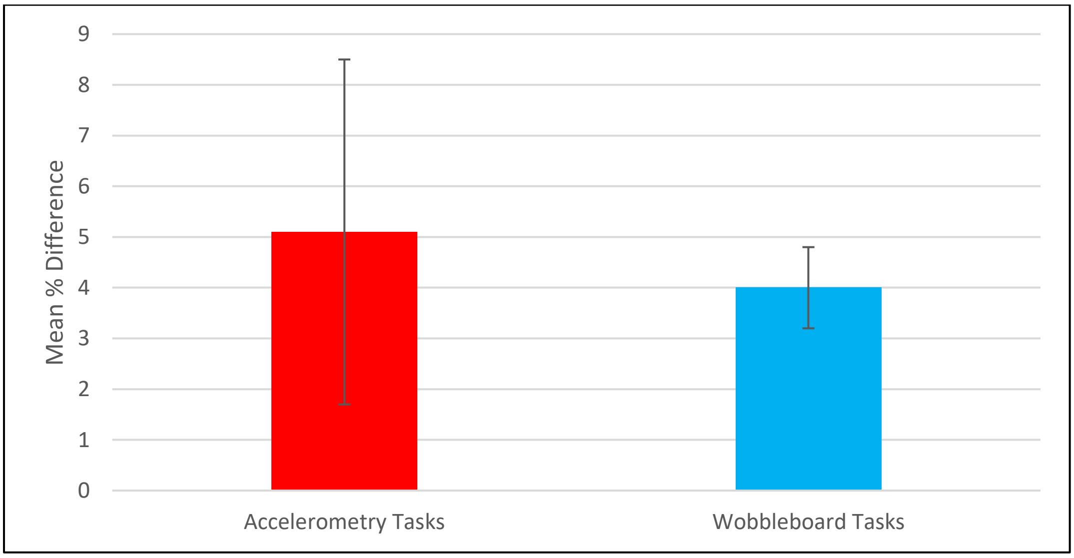


Figure 22. Comparison of absolute means for accelerometry and wobbleboard.

Given that this is the first study to review limb symmetry using an instrumented wobbleboard, the comparability of results is positive and supports the evidence that instrumented wobbleboards are capable of measuring balance. However, the wobbleboard didn't seem to be as proficient at differentiating between tasks. The accelerometry tasks all displayed significantly different levels of difficulty, allowing the confident suggestion that the accelerometer can identify different task types. The accelerometry results also support the theory that moving from SLSEO to SLSEC to SLSQ represents increasing difficulty. The median normalised path length (NPL) values indicate that tasks get progressively more difficult, with single leg stand eyes open (median NPL = 8.0) being the easiest task followed by single leg stand eyes closed (median NPL = 17.3) and single leg squat (median NPL = 27.3). However, the wobbleboard data reported no statistically significant differences between single leg stand and single leg squat at any level. The weighted wobbleboard scores were 84.1 for single leg stand and 83.5 for single leg squat. With higher scores representing better balance performance, the suggestion being that wobbleboard squat may be harder than single leg stand, but this does not represent a statistically significant finding. Alternatively, it may be considered that the addition of a perturbed surface during single leg stand increases the difficulty experienced to such a level, that adding a squat challenge has minimal effect in escalating the difficulty further.

The perturbed balancing surface elicited by the wobbleboard may provide an increased workload on the lower limb joints, particularly the ankle joint. The ankle joint would likely be the most affected joint in wobbleboard stability because of its proximity to the unstable surface. This is supported by the findings of Linens et al (2016) and Wester et al (1996) who both documented the effective use of wobbleboard training for ankle instability and injury. The increased bias towards ankle stability may decrease the involvement of the knee and hip to such an extent, that when a squat is performed the increased need for knee and hip stability is almost unregistered. Furthermore, Salavati et al (2007) highlighted how fatigue of the ankle stabilisers decreased postural stability by around 25%. A variable of the wobbleboard task was that participants were required to balance for thirty seconds at a time, for the single limb stand; whilst the squat only required balance for a few seconds. Potentially fatigue may have reduced a participant’s balance score, in single limb standing, causing it to be similar to that of the squat. As a result, it may be considered that attempting to increase the difficulty of the wobbleboard by performing potentially more complex tasks, is unnecessary, as single limb standing provides a high enough difficulty level to begin with. Furthermore, it would stand to reason that using a wobbleboard for postural stability rehabilitation following knee or hip injury may be inappropriate due to the bias placed on ankle stability.

The theory that WBSLSO provides a complex balance challenge is supported by the correlation testing between accelerometry and wobbleboard. A negative significant ($p=0.000$) two-tailed correlation ($r=-0.675$) was noted between wobbleboard single limb performance and accelerometry squat performance. Higher wobbleboard scores represent better performance whilst lower accelerometry scores equal better performance. Therefore, a negative r value suggests that better wobbleboard performance strongly ($r>0.67$) relates to better accelerometry squat performance. This may help to explain the difficulty in differentiating between WBSLSO and WBSLSQO, as it suggests wobbleboard single limb stand relates to a more complex balance challenge, when compared with accelerometry.

The two-tailed correlation indicates that the relationship between the two tasks may work both ways; potentially meaning that increasing performance in one task may see an increase in ability for the related task. This may provide a training pathway for rehabilitation, following lower limb injury, where access to a wobbleboard may be limited. The strong correlation may potentially mean that practising single limb balance in a squat position may give a similar ankle stability bias; thus, targeting ankle stability in the absence of a wobbleboard. Given that the failure rate for the wobbleboard tasks is higher, there is also the possibility that single limb squat may also provide a safer alternative to the wobbleboard. However, further research would be needed to fully test the correlation between the two tasks and confirm the relationship along with how one task may affect the other; especially as this study has not reviewed training effects for participants relating to the tasks.

As the study only found one significant correlation between wobbleboard and accelerometry, Bland-Altman plots were created to review if there may be a level of agreement between the two devices. The Bland-Altman plots (Figure 17, 18, 19) would suggest a low level of agreement between the accelerometer and wobbleboard, as a large number of data points were distributed outside the acceptable levels of agreement, suggesting that perhaps accelerometry and wobbleboard measure differing variables. Fusco et al (2018) also tested the wobbleboard for concurrent validity against the Y-balance test (YBT) and similar to this study struggled to fully support the concurrent validity of the wobbleboard against validated balance measures.

However, the Fusco et al (2018) study also used an instrumented wobbleboard to measure single limb stand. Although no formal comparison was made between limbs as part of their study, reviewing their published results would suggest that their study also found no significant difference between limbs, with a mean percentage difference of 3.9%. A key difference in the study by Fusco et al (2018) is that they used real time feedback to support participants in keeping the wobbleboard inside a target zone; from the literature, this has been assumed to be similar to the inner band measurements of the instrumented wobbleboard used in this study. This would represent a different single limb stance task to this study, as participants were not party to any form of responsive assistance to maintain the wobbleboard equilibrium. The suggestion that the wobbleboard maintains similar symmetry scores between studies would support its use a valid measure of limb symmetry. Fusco et al (2018) demonstrated good wobbleboard reliability (ICC's >0.85) but their study only reviewed time spent at one tilt angle; whilst the weighted score used by this study demonstrated excellent (ICC = 0.92) test-retest reliability and allows for consideration of all the available tilt angles. Therefore, the use of an instrumented wobbleboard may be appropriate as a valid and reliable measure of limb symmetry, but potentially only allows for balance comparisons with other wobbleboard performances. The findings of this study would also advise the use of weighted wobbleboard scoring to account for all areas of tilt during testing.

Also, whilst the wobbleboard may represent a unique balance measurement and training device, particularly where ankle stability is a factor, it may not be the most reliable device for measuring functional task-related balance. The excellent reliability of the accelerometry data, along with the apparent ability of the accelerometer to discriminate between tasks, makes it a more suitable alternative to the wobbleboard for functional balance measurement.

The ability of accelerometry to differentiate between multiple tasks may potentially provide another diagnostic tool for highlighting balance discrepancies. The single leg stand with eyes closed would appear to be approximately 2.2 times as hard as with eyes open. This corresponds closely to the findings of Slavoljub et al (2015), who suggest approximately a 2.3 times difference between the two tasks, with eyes closed providing a poorer balance performance than eyes open. Given that visual feedback is the only variable between these two tasks, it is conceivable to assume that visual feedback is responsible for more than 50% of balance ability, at least in quiet standing. This knowledge may be useful as a diagnostic tool – if a person's increase in postural sway is significantly different to the 2.2 times when they remove visual feedback, this may be an identifier of problems in another area of balance input. Single leg standing with eyes open and closed demonstrated a significant moderate to high positive correlation (r= 0.665), which further supports the use of balance measured with visual impairment as a diagnostic tool. As such this method may provide a simple test for deficits from areas such as proprioception, weakness or neuromuscular control. However, as the minimum detectable change for the single limb standing eyes closed task is 16.2 and the median difference for single leg stand eyes closed is 17.3; it may be difficult to use this test as an outcome measure, post rehabilitation, to see if any difference has been made. Also, as single leg standing was the only task tested with eyes closed, it is impossible to say whether or not visual impairment makes all tasks twice as hard.

The single limb squat is an important everyday task when considered for its role in ascending and descending stairs. Given that around a quarter of accidental home deaths are a result of falls involving stairs (Scott 2005), it is an important consideration for rehabilitation or as a predictor of risk from accidental injury. The path length data from the single limb squat would suggest it is approximately 3.4 times harder than the single leg stand. This suggests that adding a dynamic element to balance automatically increases the difficulty of the task, at least threefold. When it is considered that climbing stairs not only increases the stress on the individual due to the amplified workload, as well as the increased difficulty of the task, it is paramount that rehabilitation programmes aim to restore or maintain squatting balance; at approximately 3.4 times that of single limb stance.

Crossley et al (2011) concluded that single limb squatting could highlight hip muscle dysfunction. The study used electromyography (EMG) to record muscle function and found reduced hip muscle activation in participants with poor single limb squat control. Crossley et al (2011) used subjective reviewing of squat performance to determine whether a participant was rated good, fair or poor. The use of accelerometry squat sway data may provide a valid and reliable objective determinant of SLSQ ability. Thus, poor SLSQ balance performance may be a predictor of hip muscle dysfunction.

Therefore, as well as using LSI to predict balance deficits, a task difficulty ratio should also be considered. Deviation from a task difficulty ratio of approximately 1:2:3, representing SLSEO:SLSEC:SLSQ, may provide further evidence of balance deficits. The particular type of balance deficit may also provide clues as to the cause of the problem, i.e. SLSEC may highlight reduced proprioception or weakness, whilst SLSQ may indicate hip muscle dysfunction. These tasks may provide the same clues as to the problem, where limb symmetry for balance is not present, in the respective tasks. For this reason, it is important to review limb symmetry for balance across multiple tasks; as a result, accelerometry may provide a powerful yet accessible clinical resource in identifying and treating balance dysfunction. Instrumented wobbleboards may also provide diagnostic use due to their likely bias on ankle stability. When LSI is outside normal limits, for wobbleboard performance, the cause should be considered to be ankle related. However, further research would be needed to review balance performance with previously diagnosed impairments to support this theory.

Reliability of Accelerometry for Hop Landing Performance

This study also attempted to take measurement of lower limb functional tasks further still, by using sacral mounted accelerometry to measure single limb hop landing. The current study is the first to use accelerometry to measure hop landing using a single limb take off and standardised hop distance. Heebner et al (2014) used accelerometry to examine dynamic postural control through several tasks, including single limb landing from a two-footed take-off. They reported ICC's of 0.732 to 0.899, which fall in line with this study. This study found excellent test re-test reliability for HM and moderate reliability for HF & HL. HM also reported the lowest minimum detectable change (MDC) of all the hop tasks suggesting this would be the most sensitive to change; therefore, suggesting that such methodology could be a reliable way to quantify hop landing performance in future clinical and research studies.

As with all the other tasks no significant differences were found between limbs for any of the hop tasks. The mean percentage difference was 4.8%±6.6%; whilst the absolute mean percentage difference was calculated as 5.4%±5.9%. This value falls in line with the previously established values for LSI from accelerometry and wobbleboard (Figure 23), suggesting that trunk mounted accelerometry may be an appropriate method for measuring a single limb hop landing.

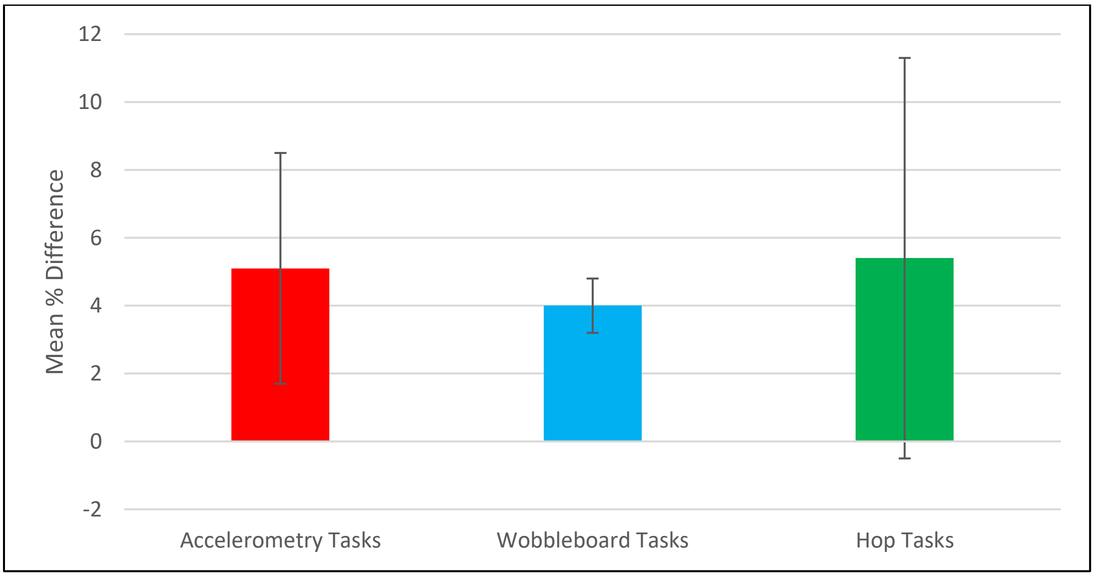


Figure 23. Comparison of absolute means for all measurements.

Single limb hopping represents an important high level functional challenge. The single limb hop signifies a transition from dynamic balance into static balance, with only the path length for the first second from landing being reported. This aims to demonstrate the difficulty experienced in single leg stability during a rapid deceleration in movement. Deceleration is necessary after any acceleration regardless of the relative velocity, to slow the body's centre of mass and stabilise posture (Hewit et al 2011). Many sports require the participant to rapidly decelerate and stabilise posture, in preparation for directional changes. The intention of decelerating whilst moving over ground is to decrease the body's momentum (mass x velocity), by generating an impulse (force x time), facilitating a complete stop or directional change to occur i.e. impulse ≥ momentum (Hewit et al 2011). The individual's efficacy when performing this task is likely to have a direct impact on their skill level in their respective sport. The involvement of leg kinematics is paramount in deceleration because of the role in force generation during impact absorption (Hewit et al 2011). Measurement of the sway of centre of mass during rapid deceleration therefore likely provides an indicator of the efficiency of lower limb kinematics – both as a predictor of injury or an indicator of return to sport.

The HF demonstrated the largest postural sway of the three hop tasks and was statistically significantly different to both the HL and HM ($p<0.01$). The HF is indicated to be 1.2 times harder than the HL and 1.3 times more difficult than the HM. The key difference in hop challenge between the HF and HM/HL being the distance participants were required to jump. The HF distance was 1.67 times further than the sideways hops. Given that the sway difference between HF and the sideways hops is not as large as 1.67; it would be reasonable to assume that HF is the easier of the three hop tasks as opposed to the more difficult. This would then suggest that movement in the sagittal plane is more stable than movement in the coronal plane. Heebner et al (2014) also reported smaller postural sway for sagittal plane hopping when compared to coronal plane. Given the previously discussed need for lower limb kinematics in decelerating the centre of mass on landing; then comparison between the muscle groups responsible may hold the key to determining directional hop landing efficiency. The expectation being that muscle groups acting predominantly in the directional plane of landing, are responsible for the deceleration of centre of mass that occurs.

Hop landing requires rapid changes between eccentric and concentric muscle contraction allowing a rapid transfer of ground reaction force, which provides the impulse required to decrease the momentum of the centre of mass (Hewit et al 2011). In the sagittal plane the ground reaction force transfer requires stability from ankle, knee and hip muscles; whilst the decreased range of movement at the ankle and knee in the coronal plane likely lead to a primary reliance on hip muscles to stabilise a sideways hop landing.

The muscles responsible for joint control in the sagittal plane include gastrocnemius (ankle), quadriceps (knee) and gluteus maximus (hip). In the coronal plane the hip muscles can be classified as either hip abductors or hip adductors. The multiple muscle groups in the sagittal plane potentially combine, allowing for greater shock absorption; whilst the limited range of lateral and medial movement across the ankle and knee joints assist the channelling of the ground reaction force towards the centre of mass. This may explain why HF appears to be easier than either sideways hop.

The theory of muscle strength being a determinant of impulse transfer during hop landing is strengthened by the sideways hop landings. There was no significant difference between HM and HL landing (p=0.96), although HM showed better sway performance than the HL. The HM sway pathway is approximately 93.5% of the HL sway pathway. The muscle patterning needed for HM & HL are in opposition to each other. HM requires hip adductor muscles to activate a medial movement of the centre of mass, but uses eccentric and concentric contraction of hip abductors to stabilise the landing and stop medial momentum; whilst HL uses abductors to direct and adductors to stabilise. Studies measuring hip abductor and adductor strength demonstrate that adductors are weaker than the abductors (Gerodimos et al 2015, Sheng et al 2016, Jung et al 2017). The studies suggest mean adductor strength is approximately 85.3% the strength of abductors (range 78.3% - 96%) (Gerodimos et al 2015, Sheng et al 2016, Jung et al 2017).

Although the mean value for strength is approximately 8% different to the sway value, it must be considered that the strength testing all used isokinetic dynamometry, with varying angular velocities and different hip positions. Therefore, the strength testing protocol is perhaps not strictly relatable to the strength needed for hop landing; however, as the sway pathway difference falls inside the range of values established for strength differences, it may be considered that hip abductor or adductor strength, is largely responsible for sideways hop landing success. Also, it is not expected that the sway difference will exactly match the difference in strength as the other balance affecting variables must be considered to have influence.

If the mean value for strength is taken to be 85.5% and the sway difference is 93.5% then it would suggest that muscle strength is responsible for around 91% of sideways hop landing ability. If muscle strength plays such a large role in forward hop landing as well, it would support the assumption that more muscle groups acting in the same plane, are responsible for making rapid sagittal deceleration more manageable.

As previously discussed, hop testing is often used as a marker of return to sport following anterior cruciate ligament (ACL) injury. The re-injury rate suggesting that testing hop height or distance, may not be the most appropriate measurement to establish readiness for return to competitive function. This may be because approximately 50% of ACL injuries were found to occur during side stepping (Olsen et al 2004, Cochrane et al 2007). Side-stepping puts more emphasis on movement in the coronal plane than hop height or hop forward. The increased difficulty exhibited by medio-lateral landings may therefore mean that hopping in the sagittal plane doesn't fully represent the level of postural stability needed for return to sport. It should also be considered, that ACL rupture may be more easily prevented by increasing adductor strength, to enhance postural control in lateral landing. Furthermore, where LSI for hop landing falls outside normal limits, it may predict muscle imbalance between limbs; with sideways hop landing potentially able to narrow down the primary antagonist to either abductors or adductors depending on hop direction.

Functional Difficulties Questionnaire (FDQ-9)

The FDQ-9 scores ranged from 12-20, suggesting that none of the study participants demonstrated functional difficulties (Score >21 Clark et al 2013). Despite this, positive correlations were recorded between FDQ-9 score and SLSQ & HF ($r=0.340$ $p=0.024$, $r=0.308$ $p=0.038$ respectively). Although the correlations are weak they support the idea that poor balance is representative of poor function (Clark et al 2016). The positive correlation suggests that as FDQ-9 score increases, the balance sensor scores also increase. As higher balance sensor scores represent poorer balance performance; higher FDQ-9 scores represent lower balance performance.

The FDQ-9 scores were reviewed against the inner band wobbleboard scores for both SLS & SLSQ. No significant correlations were recorded for these tasks, which would appear to contradict the findings of Clark et al (2016), who suggested moderate correlations between FDQ-9 and WBSLS inner band scores. However, the Clark et al (2016) study had a larger number of participants exhibiting functional difficulties (n=11), which may explain the evidence of correlations between poorer functional scores and poorer balance. Clark et al (2016) also used a lower cut off score as representative of functional difficulties (≥19/36) rather than the cut off established in the FDQ-9 creation study (Clark et al 2013) of >21/36. Despite this, the findings of this study may still indicate that the FDQ-9 is sensitive enough to predict expected balance deficits for higher level balance function, in people who don’t meet the criteria for true functional disorders via the FDQ-9.

Summary

The aims of this study were all achieved. It is evident that the sacral mounted accelerometer and SMART wobbleboard offer a reliable method of quantifying limb symmetry performance and this has been extended to include single leg squatting and hop landing. The quantification of SEM and MDC values will assist the interpretation of true metrics of change and offer real insights in quantifying performance variations. No significant differences were determined between the dominant and non-dominant limbs regardless of task or measurement method. In addition, this study has quantified the LSI for a range of tasks and across a range of measurement methods. Overall, it appears that LSI around 5% is the expected difference between the dominant and non-dominant limbs. This simplified value will aid in the interpretation of balance impairments, either as part of screening and assessment or following rehabilitation, by assisting to determine whether ‘normality’ has been restored. The ability of the accelerometer to distinguish between tasks may assist the prediction of balance deficits, through a task difficulty ratio, as well as providing clues as to the cause. Accelerometry would also appear to be a novel yet reliable way, to easily quantify hop landing performance. This provides a useful alternative to vertical and distance hop testing when assessing achievement of rehabilitation goals. The FDQ-9 may also prove a useful subjective predictor of balance performance in healthy individuals with no functional disorders. The final aim of this study was to explore the relationship between balance performance and wobbleboard performance. The findings of this study infer a very limited relationship and suggest a lack of support for using wobbleboards to measure balance, or to help compensate for every balance disorder. However, wobbleboards may still be appropriate for identifying limb asymmetry, particularly where ankle dysfunction is suspected.

Study Limitations

This study failed to meet its sample size requirement. The expected number of participants needed was thirty. Post hoc sample size calculations would suggest that the study is adequately powered for the aim of studying the LSI for balance across differing challenges. However, a study with more participants would perhaps yield stronger results and may lead to stronger assumptions regarding the concurrent validity of the wobbleboard and the use of the FDQ-9.

A secondary limitation is that the study only provides a snapshot of information regarding ‘normal’ participants; therefore, inferences regarding those with injury or disease cannot be made without further research. Furthermore, a young sample was achieved and as such extrapolation to older individuals may not be possible.

Implications for future research

This study would appear to have laid a foundation regarding single limb balance and its potential uses in clinical practice – particularly for screening lower limb injuries and planning their rehabilitation.

Further projects that may help build on this foundation and support these theories should be considered. Suggested proposals for future research would include:

- Testing the more advanced balance tasks with eyes closed where possible. This would help to establish the effect of visual impairment on single limb balance.
- Testing participants with a known lower limb injury to review the effect on LSI.
- Re-test the hop landings using standardised distances for all hop landing tasks.
- Randomised controlled trials to review training effects of the correlated tasks.
- Randomised controlled trial to review effect of adductor strengthening on lateral hop landing sway.

6.Conclusion

The suggestion of this study is that LSI for lower limb balance performance should be around a 5% difference between limbs, with no specification as to which limb should be expected to perform better. No significant differences were found between limbs for any of the tasks performed indicating that task difficulty does not affect LSI for balance. Therefore, LSI for balance may provide a way of highlighting deficits in between limb ability, thus acting to predict the likelihood of physical dysfunction or performance related injury. The progression of difficulty of the task may also have provided a further predictor of injury risk, by highlighting the ratio of balance performance across the tasks. Potentially, deviating from the ratio may indicate abnormal function and an increased risk of harm during balance performance. As a result, LSI and task balance ratio may provide a solid indicator of rehabilitation post injury, when measured by accelerometry. The wobbleboard has been shown to potentially provide an advanced balance challenge; perhaps similar to the oscillating balance of the single limb squat. It may also provide a reliable measurement for LSI with a possible emphasis on ankle instability. Sacral mounted accelerometry would also appear to be a valid and reliable measurement of hop landing performance. This may assist with clinical assessment of advanced balance challenges, to aid rehabilitation and provide a predictor of injury, particularly for more complex functional tasks.

The information gained as part of this study has potentially yielded important information relating to the relationships between tasks and balance types. However, as previously discussed the limitations of the study would suggest that a large amount of further research needs to be conducted to provide further evidence to support these findings; although, this study may have provided an indication of the expected levels for some balance affecting factors, such as eyesight and muscle strength. The correlations between tasks may also enable speculation regarding balance training and the use of wobbleboard training in clinical practice.

References

Alnahdi, A.H., Alderaa, A.A., Aldali, A.Z., Alsobayel, H., 2015. Reference values for the Y Balance Test and the lower extremity functional scale in young healthy adults. *Journal of Physical Therapy Science*, 27;3917-3921.

Armstrong, C.A., Oldham, J.A., 1999. A comparison of dominant and nondominant hand strengths. *Journal of Hand Surgery*, 24(4):421-5.

Aslan, U.B., Caylak, U., Yagci, N., Akdag, B., 2008. Balance performance, aging and falling: A comparative study based on a Turkish sample. *Archives of Gerontology and Geriatrics*, 46:283-292.

Ates, B., 2017. Investigation of Static and Dynamic Balance Performances of Female Student-Athletes and Non-Athletes Between 12 and 14 Years of Age. *Niğde University Physical Education and Sport Sciences Magazine*, 11;1-8.

Baierle, T., Kromer, T., Petermann, C., Magosch, P., Luomajoki, H., 2013. Balance ability and postural stability among patients with painful shoulder disorders and healthy controls*. BMC Muskuloskeletal Disorders* [Online] 1-9 Available from http://www.biomedcentral.com/1471-2474/14/282 [Accessed 22/06/2016]

Barber, S., Noyes F., Mangine, R., McCloskey, J., Hartman, W., 1990. Quantitative assessment of functional limitations in normal and anterior cruciate ligament deficient knees. *Clinical Orthopaedics and Related Research* 255:204-214.

Barbieri, F.A., Gobbi, L.T.B., Santiago, P.R.P., 2015. Dominant-non-dominant asymmetry of kicking a stationary and rolling ball in a futsal context. *Journal of Sports Sciences*, 33, 13;1411-1419.

Beauchet, O., Fantino, B., Allali, G., Muir, S.W., Montero-Odasso, M., Annweiler, C., 2011. Timed up and go test and risk of falls in older adults: A systematic review. *The Journal of Nutrition, Health & Aging: Geriatric Science*, 15(10); 933-938.

Bird, S.P., Markwick, W.J., 2016. Musculoskeletal Screening and Functional Testing: Considerations for Basketball Athletes. *International Journal of Sports Physical Therapy*, 11(5);784-802.

Bouillon, L. E., Baker, J.L., 2011. Dynamic Balance Differences as Measured by the Star Excursion Balance Test Between Adult-aged and Middle-aged Women. *Sports Physical Therapy*, 3(5);466-469.

Bryant, E.C., Trew, M.E., Bruce, A.M., Kuisma, R.M.E., Smith, A.W., 2005. Gender differences in balance performance at the time of retirement. *Clinical Biomechanics*, 20;330-335.

Burnett, D.R., Campbell-Kyureghyan, N.H., Topp, R.V., Quesada, P.M., 2015. Biomechanics of lower limbs during walking among candidates for total knee arthroplasty with and without low back pain. *BioMed Research International* [Online]. Article ID 142562 p 1-8. Available from: http://dx.doi.org/10.1155/2015/142562 [Accessed 22/06/2016]

Cannon, J., Tarpenning, K., Kay, D., Marino, F.E., 2001. Ageing is not associated with a decline in neuromuscular innervation or reduced specific force in men aged 20 and 50 years. *Clinical Physiology*, 23(3);350-357.

Chew-Bullock, T.S.Y., Anderson, D.I., Hamel, K.A., Gorelick, M.L., Wallace, S.A., Sidaway, B., 2012. Kicking performance in relation to balance ability over the support leg. *Human Movement Science*, 31;1615-1623.

Clark, C., Clark, S., Dorey, C., Williams, J., 2016. Correlation of the functional difficulties questionnaire (FDQ-9) with dynamic balance using the SMART instrumented wobbleboard. *Physical Therapy in Sport*, 21;68-74.

Clark, C.J., Thomas, S., Khattab, A.D., Carr, E.C., 2013. Development and Psychometric Properties of a Screening Tool for Assessing Developmental Coordination Disorder in Adults. *International Journal of Physical Medicine & Rehabilitation*, 1(5);1-9.

Clifford, A.M., Holder-Powell, H., 2010. Postural control in healthy individuals. *Clinical Biomechanics*, 25;546-551.

Cochrane, J.L., Lloyd, D.G., Buttfield, A., Seward, H., McGivern, J., 2007. Characteristics of anterior cruciate ligament injuries in Australian football. *Journal of Science and Medicine in Sport*, 10;96-104.

Crossley, K.M., Zhang, W.J., Schache, A.G., Bryant, A., Cowan, S., 2011. Performance on the Single-Leg Squat Task Indicates Hip Abductor Muscle Function. *The American Journal of Sports Medicine*, 39(4);866-873.

Cug, M., Ozdemir, R.A., Ak, E., 2014. Influence of Leg Dominance on Single-Leg Stance Performance During Dynamic Conditions: An Investigation into the Validity of Symmetry Hypothesis for Dynamic Postural Control in Healthy Individuals. *Turkish Journal of Physical Medicine Rehabilitation*, 60;22-26.

Daly, R.M., Rosengren, B.E., Alwis, G., Ahlborg, H.G., Sernbo, I., Karlsson, M.K., 2013. Gender specific age-related changes in bone density, muscle strength and functional performance in the elderly: a 10-year prospective population-based study. *BMC Geriatrics*, 13(71);2-9.

Daniel D., Malcolm L., Stone M., Perth H., Morgan J., Riehl B., 1982. Quantification of knee stability and function. *Contemporary Orthopaedics* 5 (1);83-91.

Darter, B.J., Rodriguez, K.M., Wilken, J.M., 2013. Test–Retest Reliability and Minimum Detectable Change Using the K4b2: Oxygen Consumption, Gait Efficiency, and Heart Rate for Healthy Adults During Submaximal Walking. *Research Quarterly for Exercise and Sport*, 84(2);223-231.

Downs, S., 2015. The Berg Balance Scale. *Journal of Physiotherapy*, 61;46.

Downs, S.H., Black, N., 1998. The feasibility of creating a checklist for the assessment of the methodological quality of both of randomised and non-randomised studies of health care interventions. *Journal of Epidemiology and Community Health*, 52;377-384.

Eguchi, R., Takada, S., 2014. Usefulness of the tri-axial accelerometer for assessing balance function in children. *Paediatrics International*, 56;753-758.

Eliks, M., Ostiak-Tomaszewska, W., Lisinski, P., Koczewski, P., 2017. Does structural leg-length discrepancy affect postural control? Preliminary study. *BMC Musculoskeletal Disorders*, 18(346);1-7.

Elosua, M.R., Ciudad, M.J., Contreras, M.J., 2017. Gender Differences in Verbal and Visuospatial Working Memory Tasks in Patients with Mild Cognitive Impairment and Alzheimer Disease. *Dementia and Geriatric Cognitive Disorders Extra*, 7;101-108.

Emery, C.A., Cassidy, J.D., Klassen, T.P., Rosychuk, R.J., Rowe, B.H., 2005. Effectiveness of a home-based balance-training program in reducing sports-related injuries among healthy adolescents: a cluster randomized controlled trial. *Canadian Medical Association Journal*, 172(6);749-754.

Fusco, A., Giancotty, G.F., Fuchs, P.X., Wagner, H., Varalda, C., Capranica, L., Cortis, C., 2018. Dynamic Balance Evaluation: Reliability and Validity of a Computerized Wobble Board. *The Journal of Strength and Conditioning Research* [Online]. In Press. Available from https://www.researchgate.net/publication/323389056_Dynamic_Balance_Evaluation_Reliability_and_Validity_of_a_Computerized_Wobble_Board [Accessed 08/05/2018].

Gabbett, T.J., 2010. The Development and Application of an Injury Prediciton Model for Noncontact, Soft-Tissue Injuries in Elite Collision Sport Athletes. *Journal of Strength and Conditioning Research*, 24(10);2593-2603.

Gerodimos, V., Karantrantou, K., Paschalis, V., Zafeiridis, A., 2015. Reliability of concentric and eccentric strength of hip abductor and adductor muscles in young soccer players. *Biology of Sport*, 32;351-356.

Giavarina, D., 2015. Understanding Bland Altman analysis. *Biochemia Medica*, 25(2);141-151.

Gokeler, A., Welling, W., Benjaminse, A., Lemmink, K., Seil, R., Zaffagnini, S., 2017. A critical analysis of limb symmetry indices of hop tests in athletes after anterior cruciate ligament reconstruction: A case control study. *Orthopaedics & Traumatology: Surgery & Research*, 103;947-951.

Golriz, S., Hebert, J., Foreman, K.B., Walker, B., F. 2012. The reliability of a portable clinical force plate used for the assessment of static postural control: repeated measures reliability study. *Chiropractic & Manual Therapies* [Online], 20(14);1-6. Available from http://chiromt.com/content/20/1/14 [Accessed 10/01/2018].

Gribble, P. A., Hertel, J., Plisky, P., 2012. Using the Star Excursion Balance Test to Assess Dynamic Postural-Control Deficits and Outcomes in Lower Extremity Injury: A Literature and Systematic Review. *Journal of Athletic Training*, 47(3);339-357.

Haley, S.M., Fragala-Pinkham, M.A., 2006. Interpreting Change Scores of Tests and Measures Used in Physical Therapy. *Physical Therapy*, 86(5);735-743.

Handrigan, G., Hue, O., Simoneau, M., Corbeil, P., Marceau, P., Marceau, S., Tremblay, A., Teasdale, N., 2010. Weight loss and muscular strength affect static balance control. *International Journal of Obesity*, 34;936-942.

Heebner, N.R., Akins, J.S., Lephart, S.M., Sell, T.C., 2014. Reliability and validity of an accelerometry based measure of static and dynamic postural stability in healthy and active individuals. *Gait & Posture* [Online]. In Press. Available from: https://www.gaitposture.com/article/S0966-6362(14)00792-9/fulltext [Accessed 08/05/2018].

Hewit, J., Cronin, J., Button, C., Hume, P., 2011. Understanding Deceleration in Sport. *Strength and Conditioning Journal*, 33(1);47-52.

Hinman, M.R., 2000. Factors affecting reliability of the Biodex Balance System: a summary of four studies. *Sport Rehabilitation*, 9;240-250.

Hoffman, M., Schrader, J., Applegate, T., Koceja, D., 1998. Unilateral postural control of the functionally dominant and nondominant extremities of healthy subjects. *Journal of Athletic Training*, 33(4);319-322

Jackson, D.W., Jarrett, H., Bailey, D., Kausek, J., Swanson-Major, J., Powell, J.W., 1978. Injury prediction in the young athlete: a preliminary report. *The American Journal of Sports Medicine*, 6(1);6-14.

Johnson, S., Leck, K., 2010. The effects of dietary fasting on physical balance among healthy young women. *Nutrition Journal*, 9;1-7.

Jung, H.C., Lee, S., Seo, M.W., Song, J.K., 2017. Isokinetic assessment of agonist and antagonist strength ratios in collegiate taekwondo athletes: a preliminary study. *Sport Sciences for Health*, 13;175-181.

Kilroy, E.A., Crabtree, O.M., Crosby, B., Parker, A., Barfield, W.R., 2016. The effect of single-leg stance on dancer and control group static balance. *International Journal of Exercise Science*, 9(2);110-120.

Kim, Y.W., Moon, S.J., 2015. Effects of treadmill training with the eyes closed on gait and balance ability of chronic stroke patients. *Journal of Physical Therapy Science*, 27(9);2935-2938.

Kinzey, S.L., Armstrong, C.W., 1998. The Reliability of the Star-Excursion Test in Assessing Dynamic Balance. *The Journal of Orthopaedic & Sports Physical Therapy*, 27(5);356-360.

Knapik, J.J., Bauman, C.L., Jones, B.H., Harris, J.M., Vaughan, L., 1991. Preseason strength and flexibility imbalances associated with athletic injuries in female collegiate athletes. *American Journal of Sports Medicine*, 19(1);76-81.

Koo, T.K., Li, M.Y., 2016. A Guideline of Selecting and Reporting Intraclass Correlation Coefficients for Reliability Research. *Journal of Chiropractic Medicine*, 15;155-163.

Lanshammer, K., Ribom, E.L., 2011. Differences in muscle strength in dominant and non-dominant leg in females aged 20-39 years - A population-based study. *Physical Therapy in Sport*, 12;76-79.

Lawrence, S., Killian, C., Rundquist, P., Jenkins, W., 2017. Measures of limb symmetry used for injury risk identification: What is normal? *British Journal of Sports Medicine*, 51;347.

Linens, S.W., Ross, S.E., Arnold, B.L., 2016. Wobble Board Rehabilitation for Improving Balance in Ankle with Chronic Instability. *Clinical Journal of Sports Medicine*, 26;76-82.

Logerstedt, D., Grindem, H., Lynch, A., Eitzen, I., Engebretsen, L., Risberg, M.A., Axe, M., Snyder-Mackler, L., 2012. Single-legged Hop Tests as Predictors of Self-Reported Knee Function After Anterior Cruciate Ligament Reconstruction: The Delaware-Oslo ACL Cohort Study. *American Journal of Sports Medicine*, 40(10);2348-2356.

Low-Choy, N.L., Brauer, S.G., Nitz, J.C., 2007. Age-Related Changes in Strength and Somatosensation during Midlife: Rationale for Targeted Preventive Intervention Programs. *Annals of the New York Academy of Sciences*, 180-193.

Lu, T.W., Chang, C.F., 2012. Biomechanics of human movement and its clinical applications. *Kaohsiung Journal of Medical Sciences*, 28;13-25.

Mackey, D.C., Robinovitch, S.N., 2006. Mechanisms underlying age-related differences in ability to recover balance with the ankle strategy. *Gait & Posture*, 23;59-68.

Masu, Y., Muramatsu, K., Hayashi, N., 2014. Characteristics of sway in the center of gravity of badminton players. *Journal of Physical Therapy Science*, 26:1671-1674.

Matsuda, S., Demura, S., Uchiyama, M., 2008. Centre of pressure sway characteristics during static one-legged stance of athletes from different sports. *Journal of Sports Sciences*, 26(7);775-779.

Maulder, P.S., 2013. Dominant limb asymmetry associated with prospective injury occurrence. *South African Journal for Research in Sport, Physical Education and Recreation*, 35(1);121-131.

Meyer, G., Ayalon, M., 2006. Biomechanical aspects of dynamic stability. *Eurpean review of aging and physical activity*, 3;29-33.

Moe-Nilssen, R., Helbostad, J.L., 2002. Trunk accelerometry as a measure of balance control during quiet standing. *Gait & Posture*, 16;60-68.

Moran, R.W., Schneiders, A.G., Mason, J., Sullivan, S.J., 2017. Do Functional Movement Screen (FMS) composite scores predict subsequent injury? A systematic review with meta-analysis. *British Journal of Sports Medicine*, 0;1-10.

Munro, A.G., Herrington, L.C., 2011. Between-Session Reliability of Four Hop Tests and the Agility T-Test. *Journal of Strength Conditioning Research*, 25(5);1470-1477.

Nakagawa, H.B., Ferraresi, J.R., Prata, M.G., Scheicher, M.E., 2017. Postural balance and functional independence of elderly people according to gender and age: cross-sectional study. *Sao Paulo Medical Journal*, 135(3);260-265.

NICE, 2013. *Falls in older people: assessing risk and prevention* [Online]. Manchester. National Institute for Health and Care Excellence. Cg161.

Noohu, M.M., Dey, A.B., Hussain, M.E., 2014. Relevance of balance measurement tools and balance training for fall prevention in older adults. *Journal of Clinical Gerontology & Geriatrics*, 5;31-35.

Ogaya, S., Ikezoe, T., Soda, N., Ichihashi, N., 2011. Effects of Balance Training Using Wobble Boards in the Elderly. *Journal of Strength Conditioning Research*, 25(9);2616-2622.

Olsen, O.E., Myklebust, G., Engebretsen, L., Bahr, R., 2004. Injury Mechanisms for Anterior Cruciate Ligament Injuries in Team Handball: A Systematic Video Analysis. *The American Journal of Sports Medicine*, 32(4);1002-1012.

Owen, J.L., Campbell, S., Falkner, S.J., Bialkowski, C., Ward, T., 2006. Is there evidence that proprioception or balance training can prevent anterior cruciate ligament (ACL) injuries in athletes without previous ACL injury? *Physical Therapy,* 86(10);1436 – 1440.

Oxford Dictionary, 2018. *Balance* [Online]. Oxford. Oxford University Press. Available from: https://en.oxforddictionaries.com/definition/balance [Accessed 08/05/18]

Painter, J.A., Elliott, S.J., 2009. Influence of gender on falls. *Physical & Occupational Therapy in Geriatrics*, 27(6);387-404.

Pau, M., Loi, A., Pezzotta, M.C., 2012. Does sensorimotor training improve the static balance of young volleyball players? *Sports Biomechanics*, 11(1);97-107.

Petschnig, R., Baron, R., Albrecht, M., 1998. The relationship between isokinetic quadriceps strength test and hop tests for distance and one-legged vertical jump test following anterior cruciate ligament reconstruction. *Journal of Orthopaedic and Sports Physiotherapy* 28(1);23-31.

Pijnappels, M., Reeves, N.D., Maganaris, C.N., van Dieen, J.H., 2008a. Tripping without falling; lower limb strength, a limitation for balance recovery and a target for training in the elderly. *Journal of Electromyography and Kinesiology*, 18;188-196.

Pijnappels, M., van der Burg, J.C.E., Reeves, N.D., van Dieen, J.H., 2008b. Identification of elderly fallers by muscle strength measures. *European Journal of Applied Physiology*, 102(5);585-592.

Resnik, L., Borgia, M., 2011. Reliability of Outcome Measures for People with Lower-Limb Amputations: Distinguishing True Change from Statistical Error. *Physical Therapy*, 91;555-565.

Rohman, E., Steubs, J.T., Tompkins, M., 2015. Changes in Involved and Uninvolved Limb Function During Rehabilitation After Anterior Cruciate Ligament Reconstruction. *The American Journal of Sports Medicine*, 43(6);1391-1398.

Sadeghi, H., Allard, P., Prince, F., Labelle, H., 2000. Symmetry and limb dominance in able-bodied gait: a review. *Gait and Posture* 12;34-45

Salavati, M., Moghadam, M., Ebrahimi, I., Arab, A.M., 2007. Changes in postural stability with fatigue of lower extremity frontal and sagittal plane movers. *Gait & Posture,* 26;214-218.

Sapega A., 1990. Muscle performance evaluation in orthopaedic practice. *Journal of Bone and Joint Surgery* 72A(10);1562-1574

Saunders, N.W., Koutakis, P., Kloos, A.D., Kegelmeyer, D.A., Dicke, J.D., Devor, S.T., 2015. Reliability and Validity of a Wireless Accelerometer for the Assessment of Postural Sway. *Journal of Applied Biomechanics*, 31;159-163.

Scott, A., 2005. *Falls on Stairways – Literature Review*. Derbyshire. Health & Safety Laboratory. HSL/2005/10.

Seimetz, C., Tan, D., Katayama, R., Lockhart, T., 2012. A comparison between methods of measuring postural stability: Force plates versus accelerometers. *Biomedical Science Instruments*, 48;386-392.

Sheng, S.J., Tandjung, F.A., Prananta, M.S., 2016. Comparison of Hip Abductor and Adductor Muscle Strength in Taekwondo and Non-Taekwondo Practitioners. *Althea Medical Journal*, 3(3);392-395.

Shim, K.S., 2015. Pubertal growth and epiphyseal fusion. *Annals of Pediatric Endocrinology & Metabolism*, 20;8-12.

Shumway-Cook, A., Brauer, S., Woollacott, M., 2000. Predicting the Probability for Falls in Community-Dwelling Older Adults Using the Timed Up & Go Test. *Physical Therapy*, 80;896-903.

Slavoljub, U., Goran, Z., Radmila, K., Sasa, P., Zoran, M., Bojan, J., Marko, A., 2015. Comparison of the static balance of children with and without visual impairment. *Research in Physical Education, Sport and Health*, 4(2);95-99.

Smith, B.S., Burton, B., Johnson, D., Kendrick, S., Meyer, E., Yuan, W., 2015. Effects of wearing athletic shoes, five-toed shoes, and standing barefoot on balance performance in young adults. *The International Journal of Sports Physical Therapy*, 10(1);69-74.

Sullivan, S.B.O., Portnry, L.G., 2014. *Physical Rehabilitation*. 6th Edition. Philadelphia. FA Davis.

Teranishi, T., Kondo, I., Sonoda, S., Wada, Y., Miyasaka, H., Tanino, G., Narita, W., Sakurai, H., Okada, M., Saitoh, E., 2011. Validity study of the standing test for imbalance and disequilibrium (SIDE): Is the amount of body sway in adopted postures consistent with item order? *Gait & Posture*, 34;295-299.

Uustal, H., Baerga, E., 2004. Chapter 6: Prosthetics and orthotics. In: Cuccurullo, S.J., *Physical Medicine and Rehabilitation Board Review.* New York. Demos Medical Publishing, 409-489.

Vaisman, A., Guiloff, R., Rojas, J., Delgado, I., Figueroa, D., Calvo, R., 2017. Comparison of Muscular Power Between Dominant and Nondominant Legs in Healthy Young Adults Associated with Single-Leg-Dominant Sports. *The Orthopaedic Journal of Sports Medicine*, 5(12);1-6.

Watkins J., 1999. *Structure and function of the musculoskeletal system*. 2nd Edition. Champaign. Human Kinetics.

Wellsandt, E., Failla, M.J., Snyder-Mackler, L., 2017. Limb Symmetry Indexes Can Overestimate Knee Function After Anterior Cruciate Ligament Injury. *Journal of Orthopaedic and Sports Physical Therapy*, 47(5);334-338.

Wester, J.U., Jespersen, S.M., Nielsen, K.D., Neumann, L., 1996. Wobble Board Training After Partial Sprains of the Lateral Ligaments of the Ankle: A Prospective Randomized Study. *Journal of Orthopaedic and Sports Physical Therapy*, 23(5);332-336.

Whitney, S.L., Roche, J.L., Marchetti, G.F., Lin, C. C., Steed, D.P., Furman, G.R., Musolino, M.C., Redfern, M.S., 2011. A comparison of accelerometry and center of pressure measures during computerized dynamic posturography: A measure of balance. *Gait & Posture*, 33;594-599.

Wilder, M.R., Cannon, J., 2009. Effect of age on muscle activation and twitch properties during static and dynamic actions. *Muscle & Nerve*, 5;683-691.

Williams, J., Bentman, S., 2014. An investigation into the reliability and variability of wobble board performance in a healthy population using the SMARTwobble instrumented wobble board. *Physical Therapy in Sport*, 15;143-147.

Williams, J.M., Dorey, C., Clark, S., Clark, C., 2016. The within-day and between-day reliability of using sacral accelerations to quantify balance performance. *Physical Therapy in Sport*, 17;45-50.

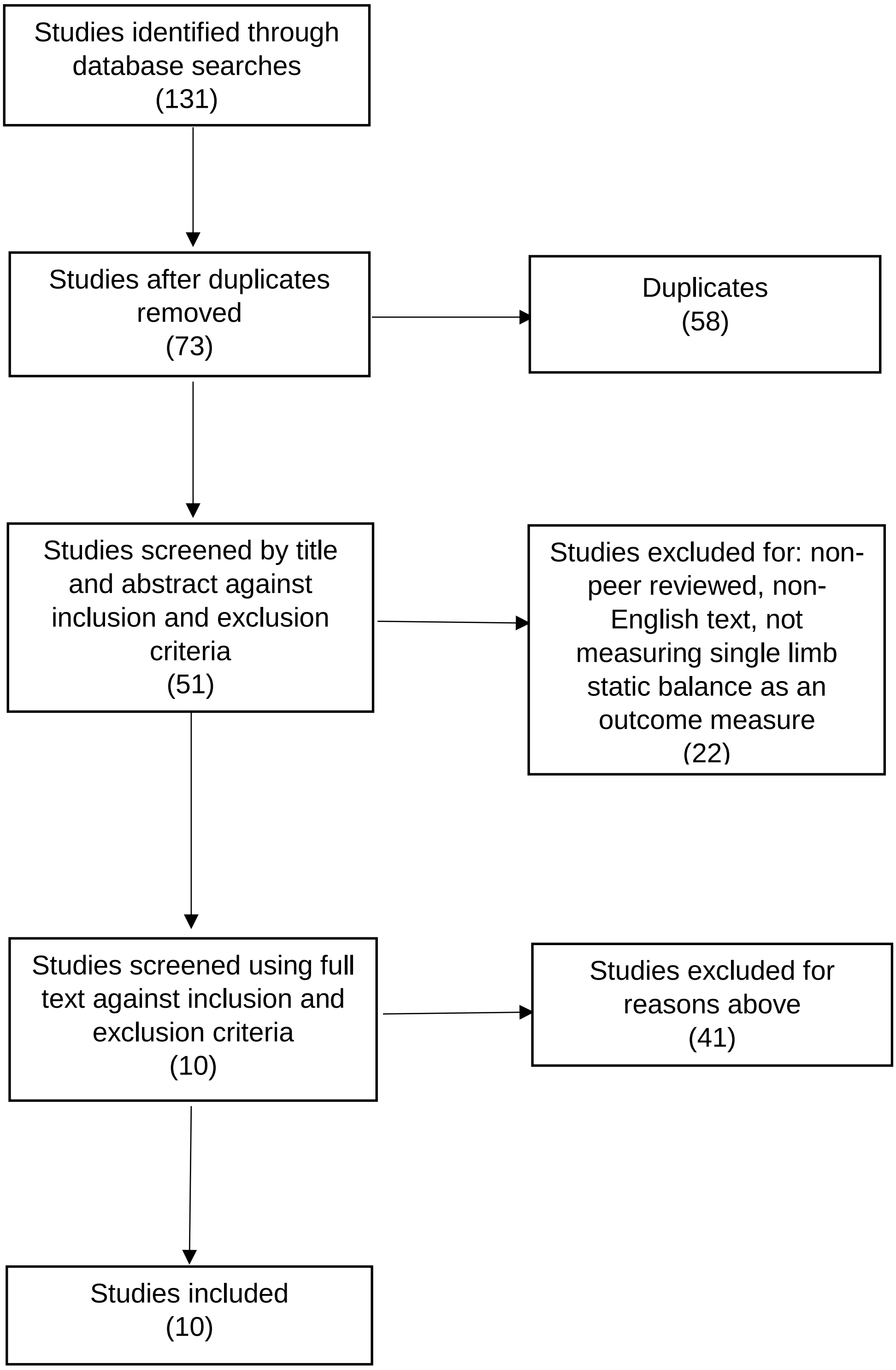
Winter, D.A., 1995. Human balance and posture control during standing and walking. *Gait & Posture*, 3;193-214.

Zvijac, J.E., Toriscelli, T.A., Merrick, W.S., Papp, D.F., Kiebzak, G.M., 2014. Isokinetic concentric quadriceps and hamstring normative data for elite collegiate American football players participating in the NFL scouting combine. *Journal of Strength and Conditioning Research*, 28(4);875-883.

Appendices

Key Concepts	Search Terms
Balance Dominant v Non-Dominant Leg Single Leg Stance Limb Symmetry	Search 1 - Balance and ‘Dominant and non-dominant leg’ Search 2 – Single Leg Stance and Limb Symmetry

Appendix 1. Search terms and associated Boolean logic.



Appendix 2. Literature review article selection flow diagram.

Author	Study Design	Participants	Eligibility Criteria	Task	Outcome Measure	Results	Quality of Evidence
Chew-Bullock et al 2012	Cross Sectional Observational	16 male 22 female	Over 18 Not reported	30 second single leg stand on both dominant and non-dominant leg	Centre of pressure data recorded via AMTI Force Plate	No significant difference between dominant and non-dominant legs. No results for gender differences recorded in dominant and non-dominant leg.	34.6%
Clifford & Holder-Powell 2010	Cross Sectional Observational	10 male 10 female	Over 18 No history of lower limb/back problems. No neurological or sensory dysfunctions.	15 second single leg stand on both dominant and non-dominant leg	Wobble data was recorded via CODA MPX 3D motion analysis system	Data not normally distributed. No significant difference between dominant and non-dominant legs. No results for gender differences recorded.	50%
Cug et al 2014	Cross Sectional Observational	21 male 24 female	Over 18 No regular exercise background for 6 months pre-testing. No diagnosis of any cardiovascular, metabolic, orthopaedic and vestibular disorders.	20 second single leg stand on both dominant and non-dominant leg. Repeated 6 times with 1 min rest in between.	Centre of mass displacement measured via Biodex Balance System SD	No significant difference between dominant and non-dominant legs. No results for gender differences recorded in dominant and non-dominant leg.	50%
Eguchi & Takada 2014	Cross Sectional Observational	198 total 87 adolescent boys 85 adolescent girls 7 male 19 female	No information provided	30 second single leg stand on both dominant and non-dominant leg	Postural sway measured via a tri-axial accelerometer	No significant difference between dominant and non-dominant legs. Significant differences between gender recorded in several age groups.	51.9%
Johnson & Leck 2010	Within subject cross-sectional cross over experimental design	22 female	Over 18. No pre-existing medical condition that might effect the outcomes measured	Single leg stand on dominant and non-dominant leg for as long as possible. Also with eyes closed.	Maximum time participants could maintain single leg balance	No significant difference between dominant and non-dominant legs with either eyes open or eyes closed. Significant differences ($p<0.01$) recorded between eyes open and eyes closed.	67.3%
Kilroy et al 2016	Cross sectional Observational	14 female 7 dancers 7 non-dancers	Completion of a pre-participation questionnaire & a score of >70% on the Lower Extemity Functional Scale.	30 second single leg stand on both dominant and non-dominant leg whilst barefoot and with an athletic shoe.	Centre of pressure measured via a Bertec Type 4060 force platform.	Significant difference in medio-lateral directions for dancers with athletic shoes. No other significant differences reported.	63.5%

Appendix 3. Data extraction table for literature review.

Author	Study Design	Participants	Eligibility Criteria	Task	Outcome Measure	Results	Quality of Evidence
Masu et al 2014	Cross sectional observational controlled study	16 male 8 high level badminton 8 low level badminton	No inclusion/ exclusion criteria provided	30 second single leg stand on both dominant and non-dominant leg with eyes both open and closed.	Centre of gravity (CoG) and sway data was recorded via a stabilometer (WBS-INK, UNIMEX Inc.) & analysed via CoG Samp Version 2.00	No significant difference between dominant and non-dominant legs with either eyes open or eyes closed. Significant differences ($p<0.01$) recorded between eyes open and eyes closed.	63.5%
Matsuda et al 2008	Cross sectional observational	40 males 10 football 10 basketball 10 swimmers 10 non-athletic	The athletes must have at least 6 years technical training in their chosen sport. The non-athletes had no such training.	60 second single leg stand on both dominant and non-dominant leg.	Postural sway was measured via a stabilometer (Gravicorder G5500).	No significant difference between dominant and non-dominant legs.	50.0%
Pau et al 2012	Experimental design with cross sectional observational controlled study and follow up post intervention	Under 18 26 female volleyball players	All athletes must have participated in the 2010-2011 regional under-14 championship of the Italian Volleyball federation	20 second single leg stand on both dominant and non-dominant leg with both eyes open and closed.	Centre of pressure was measured via force plate (Footscan 0.5 system).	Significant differences present in medio-lateral direction between dominant and non-dominant limb in all groups. Significant differences present in anterior-posterior direction for all but follow up control group. Significant difference in postural sway in the post training programme group and the starting control group. Significant difference in the centre of pressure path length in the post intervention group.	53.8%
Teranishi et al 2011	Cross sectional observational	30 men and 30 women	No known motor impairments or movement related disorders affecting balance control abilities.	20 second single leg stand on both dominant and non-dominant leg.	Centre of pressure was measured via force plate (Twin-gravicoder G6100)	No significant difference between dominant and non-dominant legs. No results for gender differences recorded in dominant and non-dominant leg.	46.2%

Appendix 3. Data extraction table for literature review.

Paper	Total Study Participants	Weighted % difference	Male weighted % difference	Female weighted % difference	Athletic weighted % difference	Non-athletic weighted % difference	Over 18 weighted % difference	Under 18 weighted % difference
Kilroy et al 2016	14	3.04±2.34		6.91±5.31	18.62±9.93	-0.41±0.66	4.92±3.79	
Cug et al 2014	45	0.32	-0.40	1.00		0.39	0.52	
Masu et al 2014	16	-0.14±0.54	-0.47±1.83		-0.76±2.96		-0.22±0.87	
Clifford & Powell 2010	20	1.44±2.80				1.75±3.42	2.32±4.53	
Chew-Bullock et al 2012	38	-0.50±2.70			-2.76±14.89		-0.81±4.37	
Eguchi & Takada 2014	198	6.32±7.71	9.08±11.06	8.31±10.14		7.72±9.41	0.46±2.97	15.45±17.16
Pau et al 2012	26	0.51±1.11		1.17±2.53	2.83±6.14			1.35±2.92
Teranishi et al 2011	60	0				0	0	
Johnson & Leck 2010	22	0.32±0.62		0.74±1.41		0.40±0.76	0.52±1.01	
Totals	439	11.31±17.78	8.21±12.89	18.13±19.39	17.93±33.92	9.85±14.25	7.71±17.54	16.80±20.08

Appendix 4. Meta-analysis data for limb symmetry index for balance.

BALANCE Project Questionnaire

All data from the project is kept confidential

Date:

Participant Code:

Age .... Height....................... Weight ....

Sex .... Leg Length………… Shoe Size…………

Which foot would you normally use to kick a football ....

Please tick the relevant answer:

A1. What is your highest educational achievement?

GCSE/CSE... A/AS/A2 level ..... Baccalaureate ... Certificate ..., BTech ...., Diploma ..., Degree ..., Masters ..., Doctorate ..., Other ....

Please tick the relevant answer:

A2. Have you ever suffered from a condition which has affected your brain or your nerves. For example cerebral palsy, head or spinal cord injury, stroke or multiple sclerosis? Please tick the relevant answer.

Yes..... No........

A3. Have you had any lower limb or back injury that has required medical intervention in the last 24 months?

Yes ...... No .....

A4. Have you had any lower limb or spinal surgery in the last 24 months?

Yes ...... No ......

A5. Do you have a current rheumatological condition (pain in your muscles and/or joints)?

Yes .... No ......

A6. Do you have a visual impairment which is not corrected by glasses/contact lens'?

Yes .... No ......

A7. Do you have any problem that currently affects your balance? i.e. Vertigo

Yes ..... No ......

Functional Difficulties Questionnaire

Please tick the box which most closely resembles your abilities.	**Very good (1)**	**Good (2)**	**Poor (3)**	**Very poor (4)**
1 **AS A CHILD,** how good was your hand writing?	☐	☐	☐	☐
2. **AS A CHILD,** were you good at team games that involved balls? i.e. football, netball, basketball,	☐	☐	☐	☐
3**. AS A CHILD,** how did others rate your coordination	☐	☐	☐	☐
4. **AS AN ADULT,** how good are you at avoiding obstacles, like bumping into doors?	☐	☐	☐	☐
5. **AS AN ADULT,** how good are you at organizing yourself? i.e. getting ready for work or for a meeting	☐	☐	☐	☐
6**. AS AN ADULT,** how good were you at catching a ball one handed?	☐	☐	☐	☐
7. **AS AN ADULT,** how good are you at balancing on a bike, in a bus or train, or on skis?	☐	☐	☐	☐
8. **AS AN ADULT,** how good are you at using your hands i.e. to do jobs around the home, DIY, sewing or using scissors?	☐	☐	☐	☐
9. **AS AN ADULT,** how good is your hand writing now?	☐	☐	☐	☐
Total Score				

Thank you for completing the questionnaire

Appendix 5. Demographic and Functional Difficulties Questionnaire.

GLOSSARY

α - Alpha

ACL – Anterior Cruciate Ligament

AP – Anterioposterior

CI – Confidence Interval

COM – Centre of Mass

COP – Centre of Pressure

DCD – Developmental Coordination Disorder

E – Edge

EMG – Electromyography

FDQ-9 – Functional Difficulties Questionnaire

HF – Hop Forward

HL – Hop Lateral

HM – Hop Medial

IB – Inner Band

ICC –Intraclass Correlation Coefficient

IQR – Interquartile Range

LSI – Limb Symmetry Index

MB – Middle Band

MDC – Minimum Detectable Change

MG – m = milli, g = units of gravity

ML – Mediolateral

MPSA - Midot Posture Scale Analyzer

NHS – National Health Service

NICE – National Institute for Health and Care Excellence

NPL – Normalised Path Length

OB – Outer Band

PC – Personal Computer

PSFS – Patient-Specific Functional Scale

PSIS – Posterior Superior Iliac Spine

RMS – Root Mean Squared

ROM – Range of Movement

SD – Standard Deviation

SEBT – Star Excursion Balance Test

SEM – Standard Error of Measurement

SF-36 – 36-Item Short Form Health Questionnaire

SLSEC – Single Leg Stand Eyes Closed

SLSEO – Single Leg Stand Eyes Open

SLSQ – Single Leg Squat

SPSS v23 – Statistical Package for the Social Sciences version 23

TUG – Timed Up and Go

UTD – Unable to be Determined

WBSLS – Wobbleboard Single Leg Stand

WBSLSO - Wobbleboard Single Leg Stand Overall

WBSLSQ – Wobbleboard Single Leg Squat

WBSLSQO - Wobbleboard Single Leg Squat Overall

Windows OS – Windows Computer Operating System

YBT – Y-balance test