Do the Gluteal Muscles Influence Dynamic Knee Valgus When Single-Leg Landing?

Ziyad Neamatallah

School of Health Science

University of Salford, Manchester, UK

Submitted in Partial Fulfilment of the Requirements of the Degree of Doctor of Philosophy

2018

List of Tables:

List of Figures

Acknowledgment

This thesis has been completed with the constant help from a number of people whom I will be glad to acknowledge and thank.

It is my pleasure to thank my principal supervisor, **Dr. Lee Herrington**, for his generous guidance, encouragement and support from the first day in my PhD. journey, which enabled me to develop an understanding of the subject. Furthermore, I wish to thank my supervisors **Prof. Richard Jones** and **Dr. Anmin Liu** for their help and guidance along this journey.

I would especially like to express my thanks to all the **participants** who gave of their precious time so that others might benefit. Without these volunteers, this project would not have been possible.

Special thanks to **my parents**; their love provided my inspiration and was my driving force. I owe them everything and wish I could show them just how much I love and appreciate them.

Sincere gratitude goes to my **lovely wife** and my soul **Meme** whose encouragement kept me going, and who supported me throughout my study journey. Without their support, patience, encouragement and sacrifice, this PhD would never have reached completion.

And last but not least, I would like to thank my country **Saudi Arabia**, and **Ministry of Higher Education** for funding my PhD, and I would also like to thank the **University of Salford** for its generous support.

Ziyad

Abstract

Background: The presence of dynamic knee valgus on landing has been found to be a significant risk factor in the development non-contact anterior cruciate ligament ACL injury. Gluteal muscles especially gluteus maximus and medius are believed to have a role in controlling hip motion that is associated with dynamic knee valgus. Landing onto one leg is a common scenario of ACL injury mechanism and would appear to require considerable Gluteal muscle activity to control the forces if the relationship were true.

Aim: the aim of the study was to investigate the relationship between the Gluteal muscles (strength and EMG activity) and the degree of dynamic knee valgus during Single Leg Squat (SLS) and multi-directional single leg landing.

Methods: Thirty-four active, healthy participants comprising of 17 males and 17 females participated in this study. Hip extension and abduction isokinetic (concentric / eccentric) strength was assessed, gluteus maximus and gluteus medius muscles Electromyography (EMG) activity was also assessed along with 3D motion lower limb biomechanics during SLS and multi-directional single leg landing tasks.

Findings: Moderate correlations were found between gluteus medius EMG activity and hip adduction angles during all landing tasks with R^2 ranging from 0.13 to 0.22. Gluteus medius EMG activity moderately correlated with knee abduction angle during right SLS and with internal hip rotation angle during left SLS. Significant moderate to strong correlations between hip abductors' and extensors' strength and knee abduction angle, hip adduction angle, knee abduction moment, hip adduction moment and internal hip rotation moment were found during landing tasks with R^2 ranging from 0.11 to 0.26.

Conclusion: There appears to be limited to moderate relationships existing between Gluteal muscles strength and EMG activity and lower limb biomechanical variables during SLS and multi-directional single leg landing tasks. Furthermore, the relationship appears also to be task, limb and gender dependent.

Key words: ACL, Landing, gluteal muscles, Hip, Strength, EMG, Biomechanics.

Chapter 1

Introduction:

1.1 Background:

Knee joint injuries are one of the commonest musculoskeletal complaints affecting youth and young adult athletes (Gage, Mcilvain, Collins, Fields and Comstock, 2012). Therefore, it is important to understand the contributory factors and causes that lead to these injuries, such as those to the anterior cruciate ligament (ACL), which occur predominantly from contact and non-contact mechanisms (Hootman et al., 2007). It has been reported that more than 70% of all ACL injuries are non-contact in nature (Quatman et al., 2014), are responsible for significant time loss in sports competitions (Hewett, Ford, Hoogenboom, and Myer, 2010) and could increase the likelihood of early knee osteoarthritis (Zabala, Favre, and Andriacchi, 2015). Most ACL injuries are reported to be due to non-contact reasons such as landing on a single leg, which is a common scenario in numerous sports and pastimes (Olsen, Myklebust, Engebretsen, and Bahr, 2004), as well as sudden deceleration while landing, due to a small knee flexion angle combined with frontal or transverse plane knee motion associated with loading in those planes (McLean, Lipfert, and van den Bogert, 2004; Quatman, Quatman-Yates, and Hewett, 2010; Shimokochi and Shultz, 2008). Boden et al. (2000) reported that 35% of ACL injuries were the result of sudden deceleration and 31% from landing. Female athletes run a greater risk of injury in this regard (two to five times more) than their male counterparts (Agel, Arendt, and Bershadsky, 2005). Nonetheless, the literature regarding risk factors is still controversial, because of the multifactorial nature of ACL injuries.

Several risk factors have been reported globally in the literature that may increase the chances of sustaining and ACL injury in both the male and the female groups. However, the main causes of these injuries can be divided into two main categories, namely extrinsic and intrinsic risk factors (Smith et al., 2012). The main intrinsic factors are biomechanical and neuromuscular, which have been focused on a great deal in the literature and are considered modifiable through intervention programmes (Sugimoto et al., 2015). One biomechanical risk factor that has been widely researched is the dynamic knee valgus – a combination of hip adduction, internal hip rotation, knee abduction and tibial external rotation (Claiborne, Armstrong, Gandhi, and Pincivero, 2006; Powers, 2010). In a study carried out on female participants who had an ACL injury, a higher peak knee abduction angle and an external knee abduction moment were found to be significant risks to sustaining an injury (Hewett et al., 2005). As a result, other researchers have investigated the factors that influence the biomechanical factors of the lower extremities (Cashman, 2012; Hollman, Hohl, Kraft, Strauss, and Traver, 2013; Willy and Davis, 2011). Hip motion during closed chain tasks such as landing and squatting is suggested to be a risk factor that can influence lower extremity biomechanics (Powers, 2010); in particular, the eccentric control of hip adduction and internal rotation has been identified as influencing dynamic knee valgus (Padua, Carcia, Arnold, and Granata, 2005; Powers, 2003). Furthermore, it has been theorised that greater external hip rotator and abductor strength may be able to resist excessive adduction and internal rotation, thus limiting knee abduction (Claiborne et al., 2006; Hollman et al., 2009). Conversely, weakness in the hip abductors and external rotators might lead to increased knee valgus motion and the potentially greater risk of ACL injury (Cashman, 2012).

The relationship between hip muscle function and dynamic knee valgus is potentially very important and controlled by two muscles: gluteus maximus (G Max) and gluteus medius (G Med). The G Max extends and externally rotates the hip, while the G Med abducts and assists in internal rotation, providing force in the opposite direction to counter valgus collapse (Hollman et al., 2009). Increases in the knee abduction angle correlate with increased hip adduction and internal rotation angles (Padua et al., 2005; Myer et al., 2010; Willson, Petrowitz, Butler, and Kernozek, 2012). Neumann (2010) claims that the G Max has the greater force for producing external rotation compared to other hip muscles, while the G Med has the greatest momentum to produce abduction compared to the gluteus minimus and tensor fascia latae (Neumann, 2010). Therefore, the hip tends to be internally rotated and adducted during landing (Powers, 2010); however, the G Max and G Med attempt to elongate, by putting the hip in a position that can improve their forcing capacity (Neumann, 2010).

In the literature, the relationship between gluteal muscle strength and dynamic knee valgus is inconclusive. Several studies have shown a correlation between the knee abduction angle and weaker G Max and G Med when investigating factors contributing to the risk of ACL injuries (Jacobs and Mattacola, 2005; Powers, 2010). However, one study found that muscle strength does not predict landing score error system values (qualitative outcome measures jump landing technique) (Beutler, de la Motte, Marshall, Padua, and Boden, 2009), noting that females are more likely to have a poor landing technique due to certain factors, one of which is landing with a higher knee abduction angle. That said, the relationship between muscle strength and specific kinematics patterns has not been investigated, and a simple qualitative assessment failed to predict a high-risk individual compared to 3D motion analysis (Ekegren, Miller, Celebrini, Eng, and Macintyre, 2009). Furthermore, a recent prospective study also failed to identify ACL injuries (Smith et al., 2012), which makes the sensitivity of qualitative assessment questionable. Claiborne et al. (2006) investigated the relationship between both concentric and eccentric hip muscle forces and knee valgus during single leg squats (SLS), concluding that there is a significant negative weak-to-moderate correlation between concentric hip abduction strength and knee abduction angle ($r= -0.37$, $R^2= 0.13$).

Researchers have also started to focus on neuromuscular components. It has been theorised that subjects may indeed have enough strength to control their lower limbs, but without an appropriate level of activation, strength is ineffectual. This theory might explain why some athletes sustain ACL injuries while others do not do so. Furthermore, there are conflicting findings concerning the impact of gluteal muscle strength on dynamic knee valgus (Bell, Padua, and Clark, 2008; Claiborne et al., 2006; Jacobs, Uhl, Mattacola, Shapiro, and Rayens, 2007). Bell et al. (2008), for instance, claim that the level of muscle activation is more important than muscle strength in determining lower limb kinematics during dynamic tasks such as landing. Previous studies have demonstrated a relationship between gluteal muscle activation and dynamic knee valgus (Hollman et al., 2009; Patrek, Kernozek, Willson, Wright, and Doberstein, 2011; Zeller, McCrory, Kibler, and Uhl, 2003), while a study carried out by Hollman et al. (2009) reveals that decreased gluteal muscle activity is related to an increased knee abduction angle, which may lead to ACL injuries. However, others have stated there is no relationship between gluteal muscle activity and the knee abduction angle (Patrek et al., 2011; Zeller et al., 2003), albeit the tasks performed in these two studies were not challenging enough.

To date, there is limited literature studying how G Max and G Med strength and the electromyography (EMG) activity of these muscles influence dynamic knee valgus motion during functional tasks, such as multi-directional single-leg landings. It has been reported, though, that there is a greater peak knee abduction angle in single-leg landing in the diagonal and lateral than in the forward direction (Sinsurin, Vachalathiti, Jalayondeja, and Limroongreungrat, 2013); yet, it is important to study these two muscles, as they are vital in controlling the hip motion associated with dynamic knee valgus. The majority of previous research has investigated the relationship between gluteal strength or muscle activation in isolation, and just a few have evaluated these factors together in active, healthy subjects (Hollman et al., 2009; Hollman, Galardi, Lin, Voth, and Whitmarsh, 2014; Hollman et al., 2013; Homan, Norcross, Goerger, Prentice, and Blackburn, 2013). In addition, it is important to investigate which one has the greater effect on lower extremity biomechanics, especially dynamic knee valgus, though it is possible that a combination of both factors could have an effect in this regard – information that may be helpful in developing an injury prevention programme.

1.2 Study Aims:

Main Aim:

The current thesis's main aim is to investigate the role of gluteal muscles (strength and EMG activity data) on dynamic knee valgus during single-leg squats and multi-directional landing.

Specific Aims:

1) To establish the within and between-days reliability of isometric and isokinetic muscle strength testing of the hip abductors and extensors.
2) To determine the consistency of EMG activity data for G Max and G Med and biomechanical variables during SLS and multi-directional single-leg land tasks, using 3D motion analysis and EMG.
3) To investigate the differences in kinetics and kinematics during SLS and multi-directional single-leg landing tasks (between limbs and genders).
4) To investigate differences in G Max and G Med EMG activity during SLS and multi-directional single-leg landing tasks (between limbs and genders).
5) To investigate the differences in hip abductors and extensors in concentric and eccentric muscle strength (between limbs and genders).

6) To determine whether there is a relationship between gluteal muscle strength and lower limb biomechanical variables during SLS and multi-directional single-leg landing.

7) To determine whether there is a relationship between EMG activity data for the G Max and G Med and lower limb biomechanical variables during SLS and multi-directional single-leg landing.

1.3 Research Questions:

The study sets out to answer the following questions:

(RQ1) Is there a significant relationship between gluteal muscle strength and the following variables during SLS and multi-directional single-leg landing?

- Peak knee valgus angle.
- Peak knee valgus moment.
- Peak hip adduction angle.
- Peak hip adduction moment.
- Peak internal hip rotation angle.
- Peak internal hip rotation moment.

(RQ2) Is there a significant relationship between the EMG activity of the G Max and G Med and the following variables during SLS and multi-directional single-leg landing?

- Peak knee valgus angle.
- Peak knee valgus moment.
- Peak hip adduction angle.
- Peak hip adduction moment.
- Peak internal hip rotation angle.
- Peak internal hip rotation moment.

(RQ3) Are there any significant biomechanical differences in the biomechanical variables between limbs during SLS and multi-directional single-leg landing?

(RQ4) Are there any significant biomechanical differences between genders during SLS and multi-directional single-leg landing?

1.4 Research Hypothesis:

- Reliability: the testing procedure is reliable and no significant differences will occur during the testing sessions.
- Prospective study:

1. There are significant relationships between muscle strength and lower limb biomechanics during single-leg multi-directional landing and SLS tasks.
2. There are significant relationships between EMG activity and lower limb biomechanics during single-leg multi-directional landing and SLS.
3. There are significant differences in biomechanical variables between single-leg multi-directional landing and SLS tasks
4. There are differences between genders in biomechanical variables during SLS and single-leg multi-directional landing.
5. There are differences between genders in hip abductor and extensor strength measurements.

1.4 Thesis Structure:

Chapter 2

Literature Review:

This chapter reviews a number of studies on the work conducted herein and provides the rationale behind this thesis through the following sections:

- ACL: anatomy, function and mechanism of injury.
- Risk factors of ACL injuries.
- Interventions used to reduce ACL injury rates.
- Synthesised systematic review.

2.1 Anterior Cruciate Ligament (ACL) Anatomy and Function:

On average, the ACL is 3 cm in length and 1 cm in diameter (Zantop et al., 2005). The medial area of the lateral epicondyle of the femur is the origin of the ACL (Domnick, Raschke, and Herbort, 2016), and the centre of the eminencies of the tibial plateau, next to the anterior horn of the lateral meniscus, is the insertion (Domnick et al., 2016). The ACL in composed of two separate bundles, named "anteriomedial" and "posteriolateral" according to the location of their insertion (Domnick et al., 2016). The ACL has an important role in creating normal knee biomechanics by providing essential support to prevent anterior tibial translation and internal rotation (Domnick et al., 2016), especially when the knee is extended, as the ACL is elongated greatly during extension (Utturkar et al., 2013). In addition, it is also considered to be the secondary restraint to varus/valgus and internal/external rotation stress across the knee (Hewett et al., 2007). In another study, during in-vivo and sophisticated modelling, ACL strain at maximum knee extension during landing was examined (Taylor et al., 2011). It has been suggested that a lack of neuromuscular control of the lower extremity may play a role in ACL injury, because of the absence of or reduction in sensory feedback from the ACL to the neuromuscular system (Borsa, Lephart, Irrgang, Safran, and Fu, 1997; Dargel et al., 2007).

The ACL prevents the anterior translation of the tibia by absorbing 75% of the anterior translation load in full extension, and 85% of the load from 30 to 90 degrees of knee flexion (Kweon et al., 2013). The anteromedial fibres become taut as the knee is flexed, and the posterolateral fibres tighten as the knee is extended. These latter fibres tend to stabilise the joint near full extension, and protect particularly against rotatory loads (Petersen et al., 2007). In addition, Buoncristiani et al. (2006) stated that the ACL also stabilises the knee from internal rotation of the tibia and knee valgus motion. Knee valgus moment, the latter of which alone can increase the risk of ACL strain by 34%, though this can be increased further with internal rotation (Shin et al., 2011).

2.2 Anterior Cruciate Ligament (ACL) Injury:

Among knee joint injuries, ACL damage is one of the most common musculoskeletal complaints affecting those participating in sports (Myer, Ford, and Hewett, 2004). ACL injuries can occur as a result of contact or non-contact, though approximately 70% of all incidents are due to the latter (Quatman et al., 2014), with 75% of these injuries occurring during competition (Olsen et al., 2004). ACL injuries are responsible for significant sporting absence from competition (Hewett et al., 2010). In addition, this injury could increase the likelihood of early knee osteoarthritis (Zabala et al., 2015), the incidence of which has been reported in the literature in female basketball and football players, ranging from 0.1–0.3 per 1000 athletes (Gwinn, Wilckens, McDevitt, Ross, and Kao, 2000; Mihata, Beutler, and Boden, 2006; Myer, Ford, and Hewett, 2011). Due to the significant effects of these injuries, it is important to understand a range of contributing factors and causes (Culvenor, Cook, Collins, and Crossley, 2012). Moreover, as many of these ACL injuries require surgical intervention aimed at maximising stability and restoring normal function (Ardern, Webster, Taylor, and Feller, 2011), they create a sizeable financial burden, with 650 million dollars a year spent at both secondary and collegiate level in the USA alone (Myer et al., 2005), i.e. total costs of roughly $17,000 per athlete for both surgery and rehabilitation (Sugimoto et al., 2012). Again in the United States, the Centre for Disease Control and Prevention has stated that around 100,000 ACL reconstructions are carried out annually (Csintalan, Inacio, and Funahashi, 2008) in the country.

With regards to returning to sports, a previous systematic review, based on 69 studies, reported on 7,556 subjects in this regard (Ardern, Taylor, Feller, and Webster, 2014). The study noted that 56% return to pre-injury levels of sport and only 55% returned to competition. In a previous work done by Ardern et al. (2011), the study noted that 33% of subjects return to pre-injury level and competition 12 months post-operation, and males were more likely to return than females. More than half of Swedish female footballers were not able to return to full activity after ACL damage, and just 15% returned to their pre-injury level (Lohmander, Ostenberg, Englund, and Roos, 2004). Among Norwegian professional handball players, 42% either quit playing after ACL reconstruction or played at a lower level (Myklebust et al., 2003).

2.3 ACL Injury Mechanism (Non-Contact):

Non-contact ACL injuries can be defined as injuries that occur in the absence of body-to-body contact (Myklebust et al., 2003). It has been reported that more than 70% of ACL injuries happen during a non-contact scenario (Quatman et al., 2014), which means there is no direct contact with the knee during its collapse, with two-thirds happening when landing from a jump (Olsen et al., 2004; Zazulak et al., 2005) – the sudden deceleration during landing or changing direction on one leg is a common scenario in this regard (Olsen et al., 2004). Olsen and colleagues conducted research in 2004 and used a questionnaire to investigate ACL injury mechanisms. Most of the respondents stated that injuries occur while changing direction and landing. Consequently, sports like basketball, netball, handball and volleyball, which use frequent landing and directional changes, have a high incidence of ACL injuries (Olsen et al., 2004), and so understanding this point is important in preventing it from happening in the first instance.

The internal structure of the knee can be damaged by excessive loading of the knee on all planes, i.e. sagittal, frontal and transverse (McLean et al., 2004, Quatman et al., 2010, Myer et al., 2008). Several studies have reported, through video analysis, that female athletes land with the knee nearly extended, the hip adducted and internally rotated, the tibia externally rotated and the foot in an over-pronated position (Olsen et al., 2004, Boden et al., 2000a). This position has been called "dynamic knee valgus" (see Figure 2.1) (Hewett et al., 2005). In a systematic review that aimed to identify strengths and weaknesses in the literature regarding the ACL injury mechanism (Quatman et al., 2010), the study found 32% of the diagnostic studies supported the theory of the sagittal plane being the only cause; however, none of these studies investigated the influence of the other planes, which is an issue because ACL load is found to alter during multiplane loading, not only on a single plane (Markolf et al., 1995). Moreover, during landing, the ACL can be injured due to reduced knee flexion combined with both frontal and transverse motion, such as hip adduction, knee abduction and internal hip rotation (Quatman et al., 2010). To damage the ACL, forces of at least 1500-2000N are required (Chandrashekar et al., 2006). However, tensile properties are not uniform throughout the population, and so forces as low as 1200N may cause

ACL injury in women compared to 1700N in men (Chandrashekar et al., 2006). Anterior tibial shear causes the most strain on the ACL, albeit not with enough force to cause ligament rupture (Berns et al., 1992; McLean et al., 2004). Sagittal plane injury computer simulations suggest the resultant force on the ACL never exceeds 900N, not enough to disrupt the ACL (McLean et al. 2004).

While the sagittal plane is important in cases of ACL damage, the frontal plane is possibly more important (Hewett et al., 2010), because it has been found that while landing, an increase in the knee abduction angle is commonplace, as exemplified in a study of injuries in female handball players (Olsen et al., 2004). Moreover, bone bruising on the lateral femoral condyle has been found when imaging the knee after ACL injury (DeAngelis and Spindler, 2010). It is theorised to occur because of the lateral compression exerted during medial opening as the knee is abducted (Quatman et al., 2010). Bendjaballah, Shirazi-Adl and Zukor (1997) reported that the load on the ACL could be six times higher as a result of as little as 5° extra knee valgus.
In a cadaveric study, ACL rupture occurred due to excessive tensile loadingat loads similar to those in landing (Meyer and Haut 2005). This excessive tensile loading could induce anterior translation of the tibia and internal tibia rotation, which in turn could cause an ACL injury (Meyer and Haut, 2008). In another study, a custom dynamic knee loading frame was developed, in order to simulate ACL injury scenarios during landing (Withrow, Huston, Wojtys, & Ashton-Miller, 2006b). The model was designed and built to hold knee specimens at a 15° knee flexion angle, using physiological levels of trans-knee muscle tension (Withrow, et al., 2006b). Peak ACL strain appeared to depend more on anteriorly directed tibial forces and knee abduction moments, with peak internal tibial rotation occurring much later in the simulated landing (Kiapour et al., 2014). However, the study found an increase in both anterior tibial translation and ACL strain, due to anterior posterior imbalances in the simulated knee muscle loads prior to impact. This is supported by previous findings suggesting that anterior translation of the tibia with respect to the femur and increased levels of ACL strain or risk of ACL injury occur under aggressive quadriceps force (Berns et al., 1992, Li et al., 1999, DeMorat et al., 2004, Wall et al., 2012, Quatman et al., 2012).

Simulated landings in Kiapour et al.'s (2014) study successfully resulted in increased anterior tibial translation, knee abduction, ACL strain and internal tibial rotation. This supported previous clinical video analysis and in vivo biomechanical studies indicating that these issues are associated with landing (Ford et al., 2003, Hewett et al., 2005, Koga et al., 2010). However, by the time that peak ACL strain occurs, it has been found that internal tibial rotation reaches up to 63% of its maximum value. At this time, anterior tibial translation and knee abduction have already reached their peaks (Kiapour et al., 2014). This finding suggests that although internal tibial rotation contributes to increased ACL strain, it seems to be secondary to anterior tibial translation and knee abduction in affecting ACL and the potential risk of injury, as noted by the knee joint kinematics timing sequence (Kiapour et al., 2014). Furthermore, previous studies have stated higher peak ACL strains and rates of damage under anterior shear force and abduction moment when compared to internal tibial rotation moment (Levine et al., 2013, Quatman et al., 2013).

These cadaveric studies have clarified how the ACL is loaded and injured and form the basis for assessing biomechanical variables thought to represent movement biomechanics. However, the usefulness of a cadaveric study is questionable due to the fact that specimens have been destroyed structurally and may have lower bone density (Wall et al., 2012). While ACL injuries occur in multiplane motion, it is important to understand the contributing factors that lead to this motion, and what can be done to prevent it. While ACL injuries occur in multiplane motion, it is important to understand the contributing factors that lead to this motion, and what can be done to prevent it.

2.4 ACL Risk Factors:

In identifying the risk factors for non-contact ACL injuries, researchers have implicated a number of reasons, such as biomechanical, environmental, hormonal and neuromuscular issues (Griffin, 2006). Reporting such factors might be beneficial to rehabilitation programmes and screening tasks, which in turn would help in preventing ACL injuries. The biomechanical and neuromuscular aspects are considered modifiable factors, whereas the others are not thought of in the same vein (Smith et al., 2012).

2.4.1 Biomechanical Risk Factors:

The knee joint moves in three planes, namely the frontal (abduction/adduction), the transverse (internal/ external rotation) and the sagittal (flexion/extension) (Woo, Abramowitch, Kilger, and Liang, 2006). Alterations in the alignment of the hip, as well as knee frontal and transverse motion during functional tasks such as landing, is often called “dynamic knee valgus,” which has been studied because of its potential role as a factor in non-contact injuries ACL injuries (Hewett et al., 2005; Homan et al., 2013). This motion (see Figure 2.1) is a combination of internal hip rotation and adduction, knee abduction, external rotation of the tibia and foot pronation. Each plane will be reviewed in detail in this chapter.

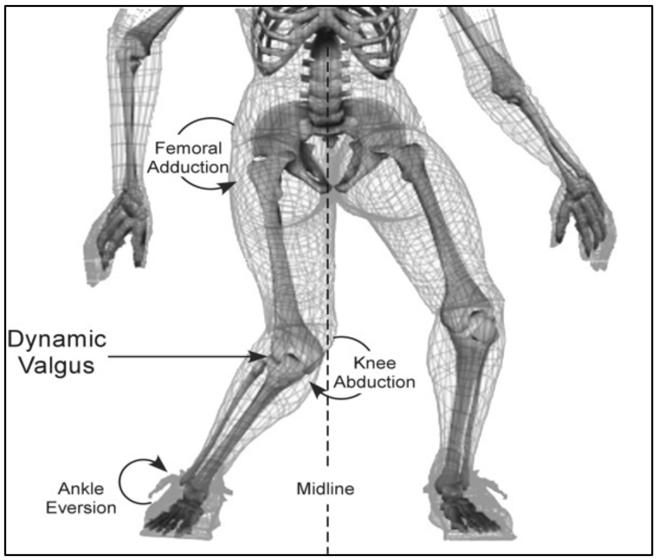


Figure 2.1 **Non-contact ACL injury mechanism, adopted from (Hewett et al., 2006)**

Frontal Plane:

The link between dynamic knee valgus and non-contact ACL injuries has been reported in the literature, using computer modelling experiments (Fukuda et al., 2003). Powers (2003) suggested that an increase in hip adduction motion might lead to increased knee abduction, while Hewett et al. (2005) carried out a prospective study among 205 female athletes from different sporting activities, aiming to report on the relationship between ACL injury and frontal plane motion, finding that hip adduction moment was indeed strongly correlated with knee abduction moment in those who suffered from an ACL injury. However, limited information was reported regarding this link, as this finding was secondary because the main aim of the study was different; nonetheless, the excessive hip adduction angle was clearly reported in the literature during ACL injury episodes (Olsen et al., 2004; Shin, Chaudhari, and Andriacchi, 2009). Additionally, greater external hip adduction moment has been found to predict ACL injuries in females, possibly because of a link between controlling these forces and difficulties in controlling hip motion, thus increasing the dynamic knee valgus angle (Hewett et al., 2006). It has been suggested that an excessive hip adduction angle is due to an ipsilateral trunk lean while standing, because of a decrease in hip abductor muscle strength (Dierks, Manal, Hamill, and Davis, 2008). However, during unilateral and bilateral landing, hip adduction was found in the absence of trunk lean (Power, 2010), which might explain that the problem may lie in the hip and not the trunk.

Another motion reported in the literature is knee abduction, also referred to as "knee valgus." Whilst Bendjaballah, Shirazi-Adl, and Zukor (1997) report that the load on the ACL could be six times higher with as little as 5° extra knee abduction, McLean et al. (2004) state that a 2° change in this regard could lead to a 100% increase in abduction moment at the knee. The forces that can be generated with knee abduction alone have been shown to produce ACL rupture in both modelling (McLean et al., 2004) and in vitro (Withrow et al., 2006) studies. Limited Cadaveric and in vivo studies demonstrate that the frontal plane movement is a risk factor in ACL strain. One in vivo study, for example, showed that 10 newton metres of abduction moment, or adduction moment at 20° of knee flexion, can produce a 10° rotation in the frontal plane (Shultz et al., 2007). Cadaveric specimens exposed to adduction torque show increases in ACL tension throughout a range of knee flexion angles, with greatest tension between zero and 30° of knee flexion (Markolf et al., 1995; Miyasaka et al., 2002). Greater axial forces on the lateral side of the knee than on the medial side result from increased knee abduction, which in turn increases the lateral compressive forces that may contribute to greater internal rotation. In addition, with knee abduction, the ligaments on the lateral side of the knee may experience a reduction in tensile force, while the medial ligaments increased tensile force. This may allow the lateral tibial plateau to shift anteriorly with internal rotation, which can increase strain on the ACL (Markolf et al., 1995). Furthermore, a robotic arm to examine ACL tensile forces has shown that the combined loading of a valgus knee, with either internal knee rotation or external rotation increases ACL tensile force (Gabriel et al., 2004; Kanamori et al., 2002). However, ACL tensile force almost doubled with combined valgus and internal knee rotation loads than with combined valgus and knee external rotation loads at 15° knee flexion and was greater than an isolated valgus load at 30° knee flexion (Kanamori et al., 2002). These studies provide evidence that when excessive valgus and internal knee rotation loads are combined near full knee extension, the ACL may be at greater risk of strain and injury.

In the prospective study of Hewett et al. (2005), the authors reported nine players had an ACL injury, and those players had a markedly excessive knee abduction angle and moments during jump landing manoeuvres. Injured females exhibited a 5° knee abduction angle on initial contact and a 9° peak knee abduction angle, which is greater than uninjured females by 8.4° at an initial contact and 7.6° at peak. Moreover, the study reported that knee abduction moment was a strong predictor of non-contact ACL injuries, with 78% sensitivity and 73% specificity. However, the study found the relationship only in adolescent females and not in older or male subjects.

The relationship between hip adduction angles and knee abduction angles has been reported previously in the literature, explaining the importance of hip motion as a driver for knee motion (Hewett et al., 2005; Olsen et al., 2004). As the hip adducts, the distal femur will move the knee joint medially, which leads to greater dynamic knee valgus if the foot is fixed and will make any ground reaction force move laterally to the knee joint and result in greater knee abduction moment. During a step-down task, Hollman et al. (2009) found that the hip adduction angle was correlated with the knee abduction angle ($r = .75$, $p = 0.001$) and stated that hip adduction accounted for 57% of variance in the knee abduction angle, the author thereby suggesting that a reduced excessive knee valgus angle can be prevented by controlling the excessive hip adduction angle. However, the study used 2D to analyse kinematics, so further investigation is needed using a 3D motion analysis model. Powers (2010) claimed that during weight-bearing exercise, excessive hip adduction combined with internal rotation plays a vital role in affecting the knee joint by moving it medially in relation to the foot position (Powers, 2010).

Transverse Plane:

It has been suggested that internal hip rotation may play a significant role in forming dynamic knee valgus (Powers, 2003). Fung et al. (2007), for instance, stated that it leads to external rotation of the knee, which could lead to ACL impingement against the lateral femoral condyle in the trochlear notch, thus increasing the risk of injury. It has been suggested that tibial external rotation has a potential role in ACL injuries (Ireland, 1999) and has been seen in injury episodes through video analysis (Fung et al., 2007; Markolf et al., 1995; Olsen et al., 2004). Oh et al. (2012), for example, state that tibial internal rotation could cause ACL strain, even more so than external rotation. It has also been reported that in the case of knee extension with internal tibia rotation, the femur must also be internally rotated, which leads to an increase in the hip adduction angle (Tiberio, 1987). Moreover, ACL impingement can be caused by external rotation and lead to a disproportionate increase in ACL load, though this might not happen during internal rotation (Fung et al., 2007). Cadaveric studies demonstrate that high tensile or internal torsional tibial loads can cause ACL damage along with limited damage to other knee ligaments (Meyer et al., 2008). Similarly, one study showed that internal tibial torque generates significantly higher ACL forces than the application of 100N of anterior tibial force during shallow knee flexion angles (Markolf et al., 1995) In contrast, external tibial torque applied to cadaveric knees demonstrated little difference in ACL strain and tension over a wide range of flexion angles (Markolf et al., 1995). Skiing results in a high rate of ACL injury. A common mechanism described in this regard is internal tibial rotation, or a combination of high axial loading with transverse plane rotations (Jarvinen et al., 1994). However, comparisons between ACL injuries that result from skiing and those that occur during landing tasks are questionable, because skiers have different movements and mechanical constraints, since their feet are fixed in ski bindings and they have the added extensions of the skis. These may increase the moment arms for applying external multi-planar loads to the distal end of the lower extremity.

During landing, several studies have found a correlation between internal hip rotation and the knee abduction angle (McLean et al., 2004; Sigward, Ota, and Powers, 2008). Internal hip rotation represents 56-60% of the change in peak knee abduction angle during side cutting manoeuvres, according to McLean et al. (2004), though the authors did not investigate this relationship during other tasks such as single-leg landing. It is not clear if internal rotation of the hip alone is actually a risk factor, but internal hip rotation, combined with hip adduction and knee abduction, might increase this possibility.

Excessive foot pronation could alter lower limb biomechanics and lead to proximal joint musculoskeletal injuries (Gross et al., 2007; Resende, Deluzio, Kirkwood, Hassan, and Fonseca, 2015). A previous study has revealed a correlation between hip adduction and subtalar eversion during gait, and it is suggested that foot kinematics may lead to the development of PFP syndrome (PFP) (Barton et al., 2012). This relationship between subtalar eversion and hip adduction might be important in sustaining ACL injuries, as hip adduction contributes in dynamic knee valgus. Piva et al. (2005) stated that decreased flexibility of the calf muscles (gastrocnemius and soleus) could be another factor influencing excessive foot pronation in order to achieve the required movement in ankle dorsiflexion. Moreover, it has been suggested that limited ankle dorsiflexion may lead to increased dynamic knee valgus in functional tasks that require simultaneous knee flexion and ankle dorsiflexion. Among females, an increase in frontal knee plane excursion (differences between right and left markers from initial contact to maximum knee flexion) is related to less ankle dorsiflexion during drop landing tasks (Sigward et al., 2008). Moreover, limited ankle dorsiflexion has been suggested to increase medial knee displacement during lateral step-down tasks (Rabin and Kozol, 2010). However, in the latter study, lower limb motion was assessed visually, and so it would have been better if the study had used a motion analysis system instead.

Sagittal Plane:

A change in sagittal plane load can play a role in altering ACL load. In both vitro and vivo studies, it has been found that quadriceps contraction peaks between 15 and 30° can cause tibial anterior translation and ACL strain (Pandy and Shelburne, 1997). Mclean et al. (2004), for instance, stated that ground reaction force and muscle contraction synergy can protect against ACL injuries, and it has been found recently that greater ACL sprain often occurs in a relatively extended knee position (Markolf et al., 1995, Berns et al., 1992). Females exhibit a shallower knee flexion angle (less than 25°) with lesser force absorption at the hip during landing, which increases knee loading (Chappell et al., 2005, Decker et al., 2003). Two studies have investigated the relationship between lower extremity kinematics and kinetics and quadriceps activation in relation to two sagittal plane trunk flexion positions (flexed and preferred) (Blackburn and Padua, 2009, Blackburn and Padua, 2008). The authors' 2009 study reported that during landing in a flexed position, knee flexion increases in line with a decrease in quadriceps activation, and a smaller peak ground reaction force is produced. This suggests that leaning forward might protect from ACL injuries by reducing the knee extensor moment and increasing the hip extensor moment during single-leg landing (Shimokochi et al., 2009). Knee frontal plane loading has been found to be more important in ACL injuries (McLean et al., 2004), and the knee's frontal and transverse planes can increase the risk of strain (Markolf et al., 1995), which suggests that dynamic knee valgus is important during dynamic screening tasks.

While isolated sagittal, frontal and transverse plane factors are suggested to increase ACL injury risk, combined knee loading across multiple planes results in the largest ACL loads (Markolf et al., 1995; Shin et al., 2011). Consequently, ACL injuries are thought to occur via a multi-planar mechanism (Quatman et al. 2010). A focus on this multi-planar mechanism is required for the development of optimal injury prevention strategies (Kiapour et al., 2015). It is also important to acknowledge that while certain biomechanical risk factors have been implicated in ACL injury, the neuromuscular system is fundamental to controlling lower limb biomechanics (Hewett et al., 2012), whereby the muscles provide support against external loads and contribute to knee joint stability during dynamic tasks such as landing (Griffin et al., 2000; Zhang et al., 2000). The presence of altered or poor neuromuscular control is likely to contribute to increased ACL loads or injury risk during high-risk sporting tasks. These resultant lower limb biomechanics should not be considered as the underlying cause of ACL injury but as a consequence of neuromuscular dysfunction. Linking neuromuscular control to lower limb biomechanics known to increase ACL loads or injury can aid in identifying neuromuscular contributions to ACL injury and provide a practical target for injury prevention strategies.

In conclusion, excessive hip adduction, internal hip rotation, knee abduction might lead to an increase in dynamic knee valgus, which in turn can increase the likelihood of an ACL injury. Dynamic knee valgus and peak knee abduction moments during landing have been found to predict ACL injury in adolescent female athletes (Hewett et al., 2005), while dynamic knee valgus with lesser hip and knee flexion have been found to reduce the capacity to absorb force on the knee joint (Brown et al., 2009). In this case, during landing, ACL as a passive joint restraint requires greater effort to stabilise the knee, as peak ACL strain occurs with lower knee flexion (Tylor et al., 2011). In cadaveric models, external moment with lower knee flexion angle has been linked to greater ACL load (Markolf et al., 1995; Shin et al., 2011; Kiapour et al., 2014), might lead to subsequent ACL damage. External moment resulted from ground reaction force can be reduced by internal moment achieved through a number of active muscular and passive ligament controls (Powers, 2010). However, anterior tibial shear with combined knee valgus and/or rotational moments cause significantly greater strain on the ACL, thereby increasing the injury potential (Berns et al., 1992; Markolf et al., 1995; McLean et al., 2004).

Differences between females and males:

The literature revealed differences between genders in knee abduction angle when performing landing tasks (Brown, McLean, and Palmieri-Smith, 2014; Ford et al., 2003; Herrington and Munro, 2010; Hewett et al., 2004; Jacobs et al., 2007; Pappas, Hagins, Sheikhzadeh, Nordin, and Rose, 2007; Schmitz, Kulas, Perrin, Riemann, and Shultz, 2007) and single-leg squat tasks (Willson, Ireland, and Davis, 2006; Zeller et al., 2003). It has been stated that the knee abduction angle and moment might predict ACL injuries (Hewett et al., 2005). Females were examined while landing with a knee abduction angle 5° greater than males, with a maximum of 11Nm higher moments (Ford et al., 2010). This was reported by Pappas et al. (2007), who noted similar knee abduction angle differences between females and males, though another study reported only a 2.4° greater knee abduction angle in females (Kernozek, Torry, and Iwasaki, 2007). This contrast might be explained by the differences in step height used in the studies, in that 31 cm, 40 cm and 50 cm were used in the Ford et al., Pappas et al. and Kernozek et al. studies, respectively. Furthermore, a number of studies have shown that women demonstrate increases in hip adduction, internal hip rotation and knee abduction during landing tasks (Brown et al., 2014; Herrington and Munro, 2010; Jacobs et al., 2007; Pappas et al., 2007), all of which might explain why females have higher injury rates in ACL than males. Furthermore, it is also important to acknowledge that the neuromuscular system is fundamental in controlling lower limb biomechanics (Hewett et al., 2012), since changes in neuromuscular characteristics are evident between male and female post-puberty along with subsequent differences in ACL injury rates. As they mature, female demonstrate significantly greater valgus motion (Ford et al., 2010; Hewett et al., 2004) and no changes in strength and power in 14 to 17 aged group, albeit strength and powers have been found to increase in male, with knee valgus remaining the same even after maturity (Barber-Westin et al., 2006). Furthermore, post-pubertal female exhibit greater valgus motion and lower strength and power than men (Barber-Westin et al., 2006; Ford et al., 2010; Hewett et al., 2004). The growth spurt associated with puberty increases lever lengths of the lower limb. The corresponding increase in strength in males during puberty enables them to counteract changes in biomechanics and maintain or improve neuromuscular control of the lower limb. In contrast, females do not make the same adaptations in terms of strength in line with decreased neuromuscular control. These changes between genders post-puberty may be responsible partly for the differences that exist in injury rates.

Differences between legs:

Kinematic differences between injured and uninjured limbs have been reported in the literature (Yamazaki, Muneta, Ju, and Sekiya, 2009a). The differences between dominant and non-dominant legs have also been investigated, noting that the dominant leg has a higher rate of ACL injuries (Faude, Junge, Kindermann, and Dvorak, 2006). The relationship between leg dominancy and non-contact ACL injuries was reported by Negrete et al. (2007), who found that females sustained more ACL injuries in their right side, which was the dominant leg in this particular cohort. Nonetheless, in a study carried out on football players, females sustained more ACL injuries in the non-dominant leg, while male players had ACL injuries in their dominant limb (Brophy, Silvers, Gonzales, and Mandelbaum, 2010). The relationship between leg dominancy and ACL injuries is therefore still unclear. In a study of recreational skiers who sustained ACL injuries during practice and competition, it was reported that female athletes demonstrated a three times higher risk of sustaining an ACL injury on their non-dominant leg (Ruedl et al., 2012). Female skiers had a 2.4 times greater ACL injury risk than males, and ACL injuries happened 85% more frequently to the left knee joint as opposed to the right side, the latter of which was reported to be the preferred kicking leg for all participants. The study suggested that if the non-dominant leg acts as a support limb with low motor control in the non-dominant leg, that leg may have high knee valgus loads and therefore a higher risk of sustaining an ACL injury. Hewett et al. (2010) noted that female athletes with increased abduction loads and high dynamic knee valgus were at greater risk of ACL damage.

Furthermore, knee injuries in football are associated with the dominant leg (Ross et al., 2004); however, the right leg is more prone to damage when participants land bilaterally and take a step back from the net (Zahradnik, Jandacka, Uchytil, Farana, and Hamill, 2015). Another study, carried out on 21 female volleyball athletes, found that dominant and non-dominant legs have different strategies when landing from a jump, especially in relation to peak ground reaction force (Sinsurin, Srisangboriboon, and Vachalathiti, 2017). The differences in performance between the right and left or the dominant and dominant leg during functional tasks might affect lower limb alignment, especially dynamic knee valgus. Lower limb neuromuscular asymmetry has been linked to ACL injuries (McEleveen et al., 2010). As a result, asymmetry in the gluteal muscles between legs might affect dynamic knee valgus and result in the inability of a weaker limb to produce or absorb the same amount of force that the stronger leg can manage to do. Reporting the differences between limbs is important for injury screen tests and intervention.

It has been suggested that movement in multiplanes will increase the load on the ACL and lead to injury (Markolf et al., 1995; Quatman et al., 2014). Therefore, focusing on controlling motions and loads in these planes is important. Hewett et al. (2012) stated that lower limb biomechanics can be controlled by the neuromuscular system. During dynamic tasks, poor neuromuscular control might reveal poor lower limb biomechanics and lead to an increase in ACL load. Therefore, understanding the relationship between the neuromuscular system and lower limbs is important. However, the performance differences between limbs or genders might also be related to non-biomechanical risk factors such as anatomical and hormonal aspects.

2.4.2 Non-Biomechanical Risk Factors:

Non-biomechanical risk factors can be extrinsic or intrinsic. Extrinsic factors include shoe-surface or contact with objects; however, these factors are hard to modify or control. Intrinsic factors include anatomical, hormonal, previous ACL injuries, neuromuscular and psychological elements (Hewett et al., 2006, Boden et al., 2000b). Hewett et al. (2006) reported several anatomical risk factors: a smaller intercondylar notch, ACL size, a greater Q angle, greater joint laxity and genu recurvatum. Differences in the hormonal profile of males and females may play a role in ACL injury rates (Hewett et al., 2004), although neither anatomical nor hormonal factors can be modified easily.

The literature reports two anatomical factors that may influence ACL injuries: femoral intercondylar notch width (Harner, Paulos, Greenwald, Rosenberg, and Cooley, 1994) and joint laxity (Myer et al., 2008). Conflicting results have been reported in studies investigating the relationship between intercondylar notch width and ACL injury (Harner et al., 1994; Uhorchak et al., 2003). This conflict may be because some studies used the intercondylar notch width and others used the notch width index (the ratio of notch width to the width of distal femur) (see figure 2.2) (Shelbourne, Davis, and Klootwyk, 1998). It is recommended to use the intercondylar width rather than the width index, because it is influenced by the subject's height (Shelbourne et al., 1998).

Figure 2.2 The notch width index (NWI) is the ratio of the width of the intercondylar notch (a) to the width of the distal femur (b) at the level of the popliteal groove (white arrow): NWI = a/b., as adapted from Sonnery-Cottet et al. (2011)

Uhorchak et al. (2003) examined the influence of the smaller intercondylar notch and ACL injury. ACL impingement on the intercondylar notch wall and smaller ACL size are the theories hypothesised, which means the reasons behind the relationship remain unclear. During knee valgus and tibial external rotation, 3D modelling has shown that the ACL may impinge against the intercondylar notch wall (Fung et al., 2007). It has also been reported that a combination of 8° of knee valgus and 13° of external rotation at approximately 40-45° knee flexion impinge the ACL against the lateral femoral intercondylar notch wall, with an increase of 1% in ACL strain (Fung et al., 2007). However, this small percentage alone is unlikely to tear the ACL. In addition, the pattern of motion is based on previous studies that measured the manual manipulation of cadaveric knees.

It has been hypothesised that the intercondylar notch alone is not related to ACL injury rates (Shelbourne et al., 1998). Furthermore, correlation between ACL size and notch width has been reported in males (Chandrashekar et al., 2005). As a result, it seems that both intercondylar notch width and ACL size may play a role in increasing the risk of ACL injuries.

Joint laxity has been linked to a high risk of ACL injury in both genders (Myer et al., 2008), although it has been stated that it is only females who differ in joint laxity between injured and uninjured limbs (Uhorchak et al., 2003), since they tend to have greater joint laxity and reduced proprioception compared to males (Myer et al., 2008). Neuromuscular control can be altered by increasing the anterior laxity of the knee (Shultz, Carcia and Perrin, 2004), and greater laxity in frontal and transverse planes demonstrates greater hip adduction, internal rotation and knee valgus angles (Shultz and Schmitz, 2009), which could lead to increased joint instability and greater anterior tibial translation, thus increasing the risk of injury.

Previous injuries and incomplete rehabilitation have been reported in studies as risk factors for repeated injuries, due to either not physically being able to return to pre-injury level or insufficient rehabilitation (Ekstrand et al., 1983, Chomiak et al., 2000); however, previous injuries cannot easily be modified or controlled. For this reason, most of the current literature tends to focus on neuromuscular and biomechanical risk factors.

2.4.3 Neuromuscular Control:

It has been reported that abnormal neuromuscular control of the lower extremities, especially the knee joint, may play a role in the non-contact element of ACL injury in females (Hewett et al., 2005). Neuromuscular factors associated with ACL injury include muscle strength, muscle activation level and patterns which might change knee joint loading (Myer et al., 2004). Neuromuscular imbalance is reported to be the main contributory factor leading to ACL injury, due to incomplete active muscular control to compensate for and reduce joint loading during dynamic tasks (Beynnon and Fleming, 1998). Neuromuscular imbalance such as quadriceps dominance, ligament dominance and leg dominance have been detected in female athletes (Hewett et al., 2005). Hewett et al. (2001) define quadriceps dominance as an "imbalance between quadriceps and hamstring recruitment patterns in which the quadriceps is activated over the hamstring in an attempt to stabilise the knee". It has been theorised that quadriceps dominance increases the anterior tibial shear force that leads to a greater risk of ACL injury (Hewett et al., 2001). Therefore, co-contraction of the hamstring muscles may help in reducing the risk of ACL injuries by reducing anterior translation (Li et al., 1999). Theoretically, all of these factors lead to ACL injury by influencing lower limb biomechanics, especially dynamic knee valgus. While anatomical and hormonal factors are not modifiable, neuromuscular factors can be modified by intervention programmes (Hewett et al., 1999; Barendrecht et al., 2011).

Hip musculature is complex, containing approximately 22 different muscles. The groups referred to as the gluteal (gluteus maximus (G Max) and gluteus medius (G Med)) are the major and strongest hip musculature (Byrne, Mulhall and Baker, 2010). G Max is the largest hip muscle, accounting for about 16% of the total cross-sectional area. About 80% of G Max inserts into the iliotibial band, while the remaining portion inserts into a distal portion of the femur's gluteal tuberosity (Reiman, Bolgla, and Loudon, 2012). The G Med, on the other hand, is a broad, fan-shaped hip muscle attached to the superior ilium and inserting laterally into the greater trochanter. Its musculature comprises anterior 2) middle and 3) posterior groups that are separated by branches from the superior gluteal nerve. The gluteus minimus is a secondary hip muscle deep-rooted anteriorly into the anterolateral aspect of the greater trochanter and forms the middle portion of the G Med (Reiman et al., 2012; Selkowitz, Beneck, and Powers, 2013). The hip musculature is shown in Figure 2.3.

G Max is a powerful hip extensor and external rotator, while G Med and gluteus minimus are the principal hip abductors accounting for about 60% of cross-sectional area of the total hip abductor musculature (Byrne et al., 2010; Selkowitz et al., 2013). The gluteal hip musculature functions to allow hip extension and rotation required during running, jumping and landing, climbing and many other dynamic athletic activities. G Max allows upward and forward body movement of the body while the hip is in a flexed position, ranging from 45° to 60°, especially during squatting and climbing steep inclines (Reiman et al., 2012). G Med, on the other hand, stabilises the femur and pelvis during weight-bearing activities by assisting in-load transfer through the hip joint, while maintaining alignment of the lower extremity relative to hip and knee joints (Presswood, Cronin, Keogh, and Whatman, 2008). Its anterior portion (gluteus minimus) abducts, allows medial rotation and assists hip flexion (Byrne et al., 2010). Hip muscle dysfunction, especially weak gluteal musculature, is widely associated with reduced athletic performance and increased risk of lower extremity injuries (Presswood, Cronin, Keogh, and Whatman, 2008) such as ACL injury.

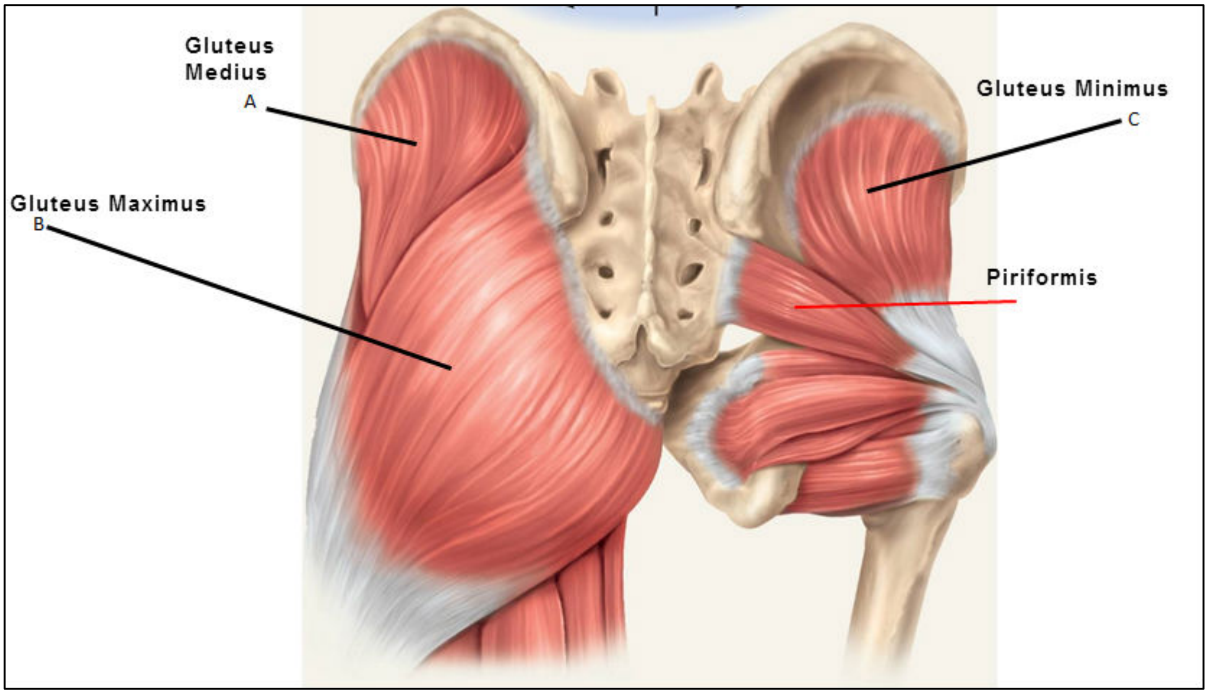


Figure 2.3 Hip joint musculature, showing the gluteus maximus, gluteus medius, gluteus minimus and the piriformis.

Muscle Strength:

Inferior hip and knee joint muscle strength may contribute to poor biomechanics and lead to ACL injuries (Cashman, 2012). The quadriceps and hamstrings mainly control movements in the sagittal plane; however, it is questionable whether the sagittal plane alone can cause an ACL injury. Increased hip adduction and internal rotation lead to an increase in dynamic knee valgus, thus risking ACL injuries (Powers, 2003, Hewett et al., 2005). As the gluteal muscles are the group responsible for working eccentrically to control excessive adduction and internal rotation during landing (Neumann, 2010), it has been suggested that weak hip musculature, including the gluteal muscles, may affect dynamic knee valgus during dynamic tasks through the failure to control hip motion (Cashman, 2012).

Claiborne et al. (2006) investigated the relationship between both the concentric and the eccentric strength of hip muscles and the knee abduction angle during SLS. The study concluded that there is significant correlation between concentric hip abduction strength and the knee abduction angle ($r= -0.37$, $R^2= 0.13$). However, Wilson et al. (2006) and Lawrence et al. (2008) produced different results, with Wilson et al. (2006) finding that during single-leg squats, isometric external hip rotation strength significantly correlates with knee valgus ($r=0.4$).

Moreover, Lawrence et al. (2008) found during single-leg drop landings that women with strong external hip rotation strength saw a decrease in the knee abduction angle and vertical ground reaction force compared to the weaker group. In previous studies, conflicting results might relate to differences in muscle strength testing tools, since Claiborne et al. (2006) used concentric and eccentric force, whereas Willson et al. (2006) and Lawrence et al. (2008) used isometric strength; therefore, it is difficult to correlate isometric strength with dynamic movement – a concept supported by several studies (Sigward et al., 2008, Willson and Davis, 2008, Jacobs et al., 2007). Moreover, the level of physical activity of the population is not stated in the studies, which makes it difficult to evaluate the impact of the population sample on the results.

The relationship between hip muscle strength and landing tasks has been investigated in the literature (Jacobs and Mattacola, 2005, Sell et al., 2010, Yeow et al., 2009). A lower peak knee abduction angle during landing has been related to greater eccentric hip abductor strength (Jacobs and Mattacola, 2005), while another study has evaluated the effect of landing height on energy dissipation in the lower limbs (Yeow et al., 2009), finding greater eccentric work on the knee and hip than the ankle, with an increase in the hip joint in response to increasing jumping height. However, the study cannot be generalised to single-leg landing, because only double-leg landings were used on two different force plates. Therefore, eccentric strength would appear to be an important element during the control of landing tasks, and the assessment of isokinetic muscle strength may provide better information about the role of the hip muscles in relation to dynamic knee valgus motion during dynamic movements such as landing or squatting.

Isokinetic muscle testing:

In clinic and research, muscle strength measurement is important, as it gives a better understanding of the influence of muscle over movement. Isokinetic testing machines are considered the gold standard for strength measurement, and tests can be performed in isometric, concentric and eccentric contraction modes (Martin et al., 2006). Mechanical isokinetic machines such as Biodex have the ability to test muscle group strength at a constant angular joint velocity, starting at zero and moving up to 500° per second. According to the Biodex manual, testing speeds of 60, 120, 180 and 300 have been recommended for hip and knee joints (Biodex, 1990). Slower isokinetic testing velocity produces greater force during eccentric contraction, and as velocity increases, the force producing capability stays the same or increases slightly (Perrin, 1993; Boling et al., 2009). However, testing below 60° per second may affect the results because of fatigue. For hip extension and abduction, it has been stated that 60° per second is a good representation of both the concentric and the eccentric capabilities of each muscle group (Boling et al., 2009). On the other hand, the risk of missing some resistance and force might increase when testing speed is more than mentioned above.

Several studies have focused on testing the reliability of knee strength assessment (Gagnon et al., 2005, Saenz et al., 2010, Pincivero et al., 2003); however, very few have focused on hip joint reliability measures (Claiborne, Timmons, and Pincivero, 2009; Julia et al., 2010; Meyer et al., 2013). A study carried out by Meyer et al. (2013) aimed at standardising a method for assessing hip joint strength using the Biodex system. Eighteen participants performed isometric and isokinetic hip muscle contractions. These subjects had to perform at isokinetic peak torque with a low speed of 60˚/sec and a high speed of 120˚/sec, and intra-class correlation coefficient (ICC) values were between moderate to high ($0.68 \leq ICC \leq 0.97$) and the standard error of measurement (SEM) from 9.48% to 23.79%. This research found higher values in hip flexion at 60°/s than a study carried out by Claiborne et al. (2009), who sought to establish the test-retest reliability of isokinetic hip torque, using the Biodex isokinetic dynamometer. Thirteen healthy adult subjects participated in two experimental tests over the course of a week. Isokinetic hip torque speed was 60˚/sec. High torque reliability was found in concentric hip flexion (right and left), extension (right) and eccentric hip flexion (right), and the extension (right and left) ICC ranged from 0.81 to 0.91 and SEM ranged from 7.80 to 14.68 Nm. Also, moderate torque reliability ICC (0.49–0.79) was found in concentric hip extension (left) and eccentric hip flexion (left) (Claiborne et al., 2009). Moreover, a study carried out by Julia et al. (2010) reported ICC values of 0.94 for concentric hip flexion at 60°/s, tested in a supine position, though ICC values of only 0.7 were found by Arokoski et al. (2002b). These two studies also showed different findings for hip extensions, which can be explained by variance in the range of tested motions. Both Julia et al. (2010) and Arokoski et al.'s (2002b) studies did not report SEM. On the other hand, although Arokoski et al. (2002b) and Claiborne et al. (2009) used different methodologies for testing isometric hip abduction, for instance standing versus supine, they still produced comparable results. A study carried out by Widler et al. (2009) stated that a side lying position is the most reliable and valid method for measuring isometric hip strength and comparing with different positions.

Overall, the link between gluteal musculature strength and lower limb motion has been reported in the literature during several functional tasks, but most of them used isometric strength in their studies. Isokinetic muscle testing has been found to be a reliable method, using different angular speeds, and may prove more representative of muscle action during functional movement.

Muscle Activation:

It has been stated that a high level of muscle strength may not be reflected in a high level of muscle activity during dynamic tasks such as landing or squatting (Homan et al., 2013), and so the level of muscle activation may be more important to lower limb kinematics (Bell et al., 2008). With this concept in mind, several research studies have investigated the quadriceps and hamstring activation and amplitude during functional tasks (Lloyd, Buchanan, and Besier, 2005; Padua et al., 2005; Shultz, Nguyen, Leonard, and Schmitz, 2009). It has been found that quadriceps contraction can apply an anterior shear force to the tibia that leads to the greater risk of ACL injury between 15 and 30° of knee flexion (Hewett et al., 2001). Greater quadriceps amplitude is significantly correlated with greater anterior shear force (Shultz et al., 2009); however, quadriceps and hamstring activation do not predict knee or hip motion in the frontal plane. Hanson et al. (2008) found greater quadriceps activation compared to hamstring muscles during the preloading and loading phase in sidestep cutting tasks. In addition, in a study carried out on 55 elite female football and handball athletes during side cutting, five athletes who suffered from an ACL tear had an increase in pre-activity levels of vastus lateralis in comparison to the level of medial hamstring (Zebis, Andersen, Bencke, Kjær, and Aagaard, 2009). These results may suggest that enhancing hamstring muscle activity may be relevant to ACL injury intervention programmes. All previous studies conclude that muscle activity might contribute to ACL injuries.

It has been stated that decreased activation of the hip muscles may lead to altered lower limb motion (Zazulak et al., 2005, Hewett et al., 2005). Several studies have investigated the effect of gluteal muscle activity and how it relates to lower limb injuries. Two studies found no significant differences in gluteus medius activation between genders (Zazulak, et al. 2005; Russell, et al. 2006), but Zazulak et al. (2005) found differences in G Max activity in females compared to males during landing tasks. However, neither study used 3D motion analysis to examine kinetics and kinematics, and thus it would be difficult to state the effect of muscle activity on the knee joint. Another study also measured knee kinematics with females during single-leg step-downs (Hollman et al., 2009). The study concluded that G Max recruitment might have a greater association with a reduced knee abduction angle in women than external rotation strength during step-down tasks. Furthermore, Homan et al. (2013) investigated the influence of gluteal activation and knee kinematics on 82 healthy participants during double leg jump landing tasks. The study stated that there were no differences between weak and strong muscles on knee valgus motion; however, the weaker group showed greater muscle activation during the task (Homan et al., 2013). Nonetheless, these studies were conducted during controlled double leg landings, step-downs and single-leg squats, and so it is difficult to predict whether these tasks are representative of those during which ACL injury actually occurs and if they can be compared to the data taken from more challenging tasks such as single-leg landings.

Electromyography (EMG):

EMG, used to assess muscle activation, is described in the literature (Ayotte et al., 2007, Bolgla et al., 2010, Bolgla and Uhl, 2005, Distefano et al., 2009) and provides an indication of the neural drive sent from the central nervous system to the muscles while an amplifier magnifies the muscle action potential and smooths out ambient noise (Pease and Lew, 2007). Action potential is a response that occurs when muscle fibres are activated by motor neurons and electrical impulses are conducted along the axon (Marieb, 2004). A surface EMG uses electrodes applied to the skin to detect these action potentials. It is important to apply these electrodes parallel to the muscle fibres and in the middle of the muscle belly (Ayotte et al., 2007).

A problem with EMG is that the level of activity detected can vary greatly between subjects, which makes it difficult to compare raw data between participants; therefore, to compare muscle activation among different subjects, EMG normalisation is required. Previous research studies have used maximum voluntary isometric contraction (MVIC), while others have normalised muscle activity through mean dynamics and peak dynamics (Benoit et al., 2003, Bolgla and Uhl, 2005, Morris et al., 1998, Neumann et al., 1992). The implementation of MVIC normalisation procedures has been criticised, but it is used in most existing studies (Frigo and Crenna, 2009). A study carried out by Bolgla and Uhl (2005) aimed to determine the reliability of three normalising methods for G Med during different rehabilitation exercises. This study found that

ICC values exceeded 0.93 for all exercises in MVIC, and 0.85 in mean dynamics and peak dynamics, except for side-lying abduction exercises. Another study by Bolgla et al. (2010) also aimed to determine the reliability of EMG methods to assess timing differences in the G Med, vastus medialis and vastus lateralis. Most of the EMG measures had ICCs > 0.7; however, others had ICCs < 0.7. Moreover, an experimental laboratory study performed a comparison between EMG signals of the G Max and G Med during different therapeutic exercises (Distefano et al., 2009). ICC values ranged from (0.93-0.98) across four repetitions for the G Med, and (0.85-0.98) for the G Max. However, forward hop and sideways hop tasks were less reliable, with ICCs ranging from (0.37-0.55) and (0.21-0.44) for G Med and G Max, respectively (Distefano et al., 2009). The standard error of measurement ranged between 30% and 41% during these tasks, which might be due to the dynamic nature of the hop activity.

Gender Differences:

It has been revealed that females demonstrate lower strength in the hip abductors, external rotators and extensors compared to males (Claiborne et al., 2006, Beutler et al., 2009). In addition, it has been reported that isometric strength is significantly lower in females, with 1-6% of body weight compared to males (Willson et al., 2006, Beutler et al., 2009). Isokinetic (concentric and eccentric) abductor and external rotator strength has been reported as being greater in males compared to females (Claiborne et al., 2006), while a recent study reported the same results in hip abductors between male and females (Sugimoto, Mattacola, Mullineaux, Palmer, and Hewett, 2014). Sugimoto et al. (2014) reported hip abductor isokinetic tests across 36 (20 females, 16 males) collegiate athletes and found significant differences between genders ($p = 0.03$). Furthermore, the concentric and eccentric torque of hip abductors has been shown to be approximately 39 Nm greater in men (Claiborne et al., 2006), though Jacobs and Mattacola (2005) found that peak eccentric hip abductor torque relative to body weight was not different between recreationally active men and women. This finding may be because Jacobs and Mattacola (2005) used 120° per second as an angular velocity; however, Sugimoto et al. (2014) and Claiborne et al. (2006) used 60° per second. For hip extension and abduction, it has been stated that 60° per second is a good representation of both the concentric and the eccentric capabilities of each hip abductor and extensor (Boling et al., 2009). In recreationally active subjects, no differences between genders in peak hip eccentric abductor strength have been found when normalised to body weight (Jacobs and Mattacola, 2005). However, Claiborne et al. (2006) did not report the sample population's activity level, which makes it difficult to determine if differences in population would affect the results.

With regards to muscle activation, no significant differences have been found between genders in G Med activation during single-leg landing, though females activated the G Max more so than males (Zazulak et al., 2005). Nonetheless, the study did not investigate how this affects lower limb kinematics, especially dynamic knee valgus. The same results have been reported by Russel et al. (2006) for single-leg drop jumps, albeit the subjects' activity levels were not reported in the study, as some individuals may have experienced landing and others not so. Both previous studies used drop landing to analyse activation, and yet this task was not as challenging when compared to single-leg multi-directional landing. Supporting this hypothesis is a study carried out with Division One football players doing forward hop landings 100 cm apart from the force plate (Hart et al., 2007). The study found significant differences in G Med activity between genders. Moreover, significant differences in G Med activation were found during side-step cutting tasks (Hanson et al., 2008). It would be reasonable to assume that the task's increased difficulty would have an effect on muscle activation. However, again, both studies did not include knee kinematics, which makes it unclear in determining the influence of these different impacts on dynamic knee valgus.

In summary, ACL injuries have been seen frequently when hip adduction, internal rotation and knee valgus angles increase in combination with lesser knee flexion while landing and changing direction (Boden et al., 2000a, Olsen et al., 2004). A combination of these motions has been called "dynamic knee valgus," and it is widely believed to be one of the primary causes of the disproportion in injury rates (Hewett et al., 2005, Ford et al., 2003, Zeller et al., 2003). Correcting these risk factors may potentially decrease injury rates. There are different factors that have an impact on ACL injuries, but some of them are not modifiable or easy to modify, for example anatomical or hormonal between genders. Despite limited sources in this regard, it is important to understand their influence on ACL injuries. On the other hand, neuromuscular and biomechanical risk factors can be modified through intervention programmes. Therefore, understanding how these factors influence lower limb biomechanics, especially dynamic knee valgus, is essential to determine the appropriate intervention programme that can help in reducing injury risk. From the previous context, most studies focus on strength or activation in isolation. It would be reasonable, therefore, to assume that any decrease in muscle strength would expose females to ACL injury. However, this might be questionable, as strong athletes still have ACL injuries. Therefore, understanding the role of muscle activation is important as well. Thus, investigating both strength and activation will provide a better picture of how they influence dynamic knee valgus during functional tasks. It might be that a combination thereof will have an influence over ACL injury risk factors.

2.5 Intervention Programmes:

Modification of risk factors has been considered as a way of preventing knee injuries such as ACL damage. Numerous neuromuscular intervention studies which target movement modification are at the forefront of this area, and they have demonstrated some accomplishments in diminishing potential biomechanical hazard variables (Myer et al., 2008, Irmischer et al., 2004, Lephart et al., 2005, Pollard et al., 2006). However, not all programmes have been beneficial in helping decrease ACL injuries. For instance, Pfeiffer et al. (2006) carried out a randomised control study focusing on improving lower limb alignment while landing. The training programme consisted of 20 minutes of plyometric exercise twice a week. The study reported no significant decrease in ACL injury rate in female players during landing and deceleration along with changing direction while running. Another study was carried out on female athletes who had greater knee abduction moment and risk of injuring the ACL during vertical drop jumping tasks (Myer et al., 2008). The study was able to reduce knee abduction moment by using a programme that included core strengthening, balance training and plyometric and resistance exercises. However, they still did not reduce their moments to that group's previously prescribed risk cut off value. However, from both studies, it would be difficult to know which exercises had an effect on the lower limbs. In a study carried out on basketball players, the knee abduction angle decreased significantly after a four-week jump training programme (Herrington, 2010).

In a study carried out with 1,263 high school volleyball, basketball and football players (Hewett, et al., 1999), the trained groups contained 185, 97 and 84 people for volleyball, football and basketball, respectively, and the untrained groups 81, 193 and 189 for each sport, respectively. The study aimed to investigate the effect of neuromuscular training, including plyometric, strengthening and flexibility, over six weeks and conducted three times a week. Each training session lasted 90 minutes. The study found five non-contact ACL injuries in the untrained female group, and six ACL injuries overall. Although there was uneven distribution of players, regardless of the high number of participants, only six players had an ACL injury. There were a high number of volleyball players in the trained group. A later study reported low ACL injury rates in volleyball players (Hootman et al., 2007).

The effect of intervention programmes on lower limb biomechanics has been investigated through several studies. Hewett et al. (2005) stated that the knee abduction angle and moment play an essential role in ACL injuries, as revealed by a number of other studies (Myer et al., 2006; Barendrecht et al., 2011; Chappell and Limpisvasti, 2008; Myer et al., 2005), most of which used a combination of different interventions, which makes it difficult to understand which one had the most effect. Some studies had a training session up to 90 minutes (Hewett et al., 1999; Heidt et al., 2000; Myer et al., 2005), but it has been suggested that a long training session is difficult to conduct (Herrington and Munro, 2010).

The existing intervention programmes differ in terms of session time and intensity, and so the effect of these interventions on reducing ACL injury rates or minimising risk factors is still unclear. Consequently, further research is needed to investigate the effect of each element and to assess how they reach the goals of reducing ACL injuries.

Several research studies have used feedback training to improve knee kinematics and kinetics (Ford et al., 2014, Barrios et al., 2010, Willy et al., 2012, Herman et al., 2009). Motor learning with a hip-strengthening programme, for instance, did improve strength and single-leg squat performance, though it did not improve running performance (Willy and Davis, 2011), albeit it is not clear which intervention had the most effect concerning those changes. In another study carried out by Willy et al. (2012), using verbal and mirror feedback while running on a treadmill, the authors identified decreased hip adduction momentand angle, which improved running performance. Nevertheless, further investigations are needed to find out whether this feedback intervention could improve single tasks such as single-leg squats and landing. Tasks such as single-leg squat distal and proximal variables must be considered, too, because they can influence lower limb loading (Herrington and Munro, 2014). Herrington and Munro (2014) performed a qualitative analysis of single-leg squats and considered knees, feet, pelvis, trunk and arm movement strategies. High scores on QASLS, which indicates poor SLS performance, are linked to the 3D motion that may increase injury risk. The authors concluded that qualitative analysis of a single-leg task seemed to improve lower limb biomechanics (Herrington and Munro, 2014).

In preliminary work, Ford et al. (2014) used two different visual feedback programmes in young (high school) female football players during double-leg squatting on vertical drop jump landing tasks. Knee abduction angle and moment decreased by 33% and 31.5%, respectively, suggesting a carryover of the effects of feedback between tasks. In separate training, participants also received visual kinematic feedback regarding the knee abduction angle, but this technique only helped the athletes hit the target knee abduction momentrange 29.3% of the time. Following training, knee abduction angle and moment were not significantly different from the baseline (Ford et al., 2014).

The combined visual and verbal feedback model was presented by Onate et al. (2005). This type of feedback has been shown to reduce ground reaction force and knee abduction significantly moment (Oñate et al., 2005). The study found that a self-and-expert model was more effective than viewing participants' performance only. It may be that the key feature of video-and-verbal feedback practice, which could improve performance, is expert performance as well as verbal instruction.

2.6 3D Motion Analysis:

There are three planes (sagittal, frontal and transverse) with six degrees of freedom (three rotation and three translation), allowing the knee joint to move in twelve directions (Woo et al., 2006). The tibia can rotate on the femur in the sagittal plane through flexion and extension, abduction and adduction in the frontal plane and internal and external rotation in the transverse plane (Woo et al., 2006). Moreover, the knee can translate anteriorly and posteriorly, medially and laterally and compress and distract motion in the sagittal, frontal and transverse planes, respectively. The knee joint structure can be damaged by excessive knee joint loading, which in turn leads to motion along these three planes (Myer et al., 2008).

To assess performance and the risk of injury in sport rehabilitation, 3D motion analysis techniques are commonly used. According to Hewett et al. (2005), dynamic knee valgus is a combination of hip, knee and ankle motion in the frontal and transverse planes. Several studies have focused on assessing the lower extremities using 3D motion analysis techniques, which allows researchers to quantify all three joint planes during tasks (Ford et al., 2003, Hewett et al., 2005, Milner et al., 2012, Cappozzo et al., 1996). The reliability of instrumented motion analysis is heavily dependent on the repeatability of marker placement between sessions. Mynard et al. (2003) studied sagittal plane motion during walking, reporting low to moderate reliability at the ankle (ICC= 0.45) and hip (ICC= 0.62), with higher values for the knee (ICC= 0.87). Furthermore, a study carried out by Ferber et al. (2003) illustrated good reliability for sagittal plane motion (ICC=0.85-0.93), though they found values were lower (ICC= 0.54 to 0.83) for secondary planes of motion. These results are similar to those found in a walking study by Kadaba et al. (1989), who reported the best results for the sagittal plane, with the lowest reliability for the secondary planes. Biomechanical variables measured within session were found to be more reliable than data from different sessions (Kadaba et al., 1989). This result has also been found during running (Ferber et al., 2003, Queen et al., 2006), vertical drop landing (Ford et al., 2007), pivoting (Webster et al., 2010), and stop-jump landing (Milner et al., 2012).

Moreover, differences between session reliability stated by some researchers are in specific planes. Across measurements, the sagittal plane has the greatest stability value during gait, running, stop-jump and vertical drop landings (Milner et al., 2012, Ferber et al., 2003, Queen et al., 2006, Kadaba et al., 1989). It is believed that the transverse and frontal plane are more sensitive to marker placement errors (Kadaba et al., 1989), which explains the lower reliability value in different sessions. The most common errors found during gait analysis were in the rotations of the hip and knee (McGinley et al., 2009). Motion in the frontal and transverse planes, in particular dynamic knee valgus, is seen as being key to high-risk movements associated with ACL (Hewett et al., 2005, Myer et al., 2010). Consequently, errors in marker placement have the greatest influence on between-session reliability (Ford et al., 2007, Queen et al., 2006). Measurement accuracy is also prone to error, due to skin movement artefacts (Cappozzo et al., 1996), the removal of which should involve using rigid marker arrays (Manal et al., 2000), while it would be better if only one examiner attaches all the markers in all study trials. In addition, the calibration anatomical systems technique (CAST) has been used to determine each segment of movement during a trial (Cappozzo et al., 1996), along with reducing skin movement artefacts by attaching the markers in the centre of the segment rather than close to the joints (Alenezi et al., 2014).

In healthcare research, intra-class correlation coefficients (ICCs) have been used widely to assess reliability. It is important to understand the reliability and measurement errors of the methodological tool that is used in screening. Batterham and George (2003) state that reliability is the ability of a subject to provide a score that can be reproduced in ensuing tests by the same subjects. Reliability can be affected by several factors, such as random errors and systematic bias, and random errors may be due to within subject variations and errors made by the examiner or measurement protocol (Hopkins, 2000). On the other hand, systematic bias could have an influence because of fatigue or the learning effect (Batterham and George, 2003). An ICC includes both systematic bias and random errors in the calculation, and it can be used when more than one test is compared to another (Atkinson and Nevill, 1998). There are several ICC models, every one of which provides different results. ICC values will be interpreted according to Coppieters et al. (2002), i.e. Poor $<.40$, Fair .40 to .70, Good .70 to .90, Excellent $>.90$. However, ICCs are sensitive to sample heterogeneity, with the lack of information relating to the actual differences between measures being a particular disadvantage (Rankin and Strokes, 1998). Therefore, they appear to be easy to interpret, but alone they cannot present a full picture of reliability, and standard errors of measurement should be measured with ICC (Rankin and Strokes, 1998). A low SEM with a high ICC indicates good measurement reliability. The advantage of SEM is that it presents the unit of measurement by providing an estimate of measurement accuracy (Denegar and Ball, 1993), which allows the researcher to compare the results with other research. Denegar and Ball (1993) state the calculation of SEM using the following formula:

$$SD\ (pooled) * (\sqrt{(1-ICC)}).$$

In summary, the reliability of single-leg landing and squats has been investigated previously in the literature, and different sample groups have been used, such as young individuals or top athletes. Some studies have examined kinematic or kinetic in isolation, but no research has investigated biomechanical variables during single-leg multi-directional landing, or how these are associated with measurement errors.

2.7 Functional Tasks:

Several functional tasks have been described in the literature to assess the biomechanical risk factors for ACL injuries, and these have been linked to the effect of the gluteal muscles and their function during several functional tasks (Homan et al., 2013, Hollman et al., 2009, Hollman et al., 2013, Hollman et al., 2014, Zazulak et al., 2005, Souza and Powers, 2009, Claiborne et al., 2006, Stearns et al., 2013). SLS is a controlled, dynamic motion that can be utilised in many functional (Claiborne et al., 2006) tasks, such as single-leg landing and changes of direction. In addition, Di Mattia et al. (2005) state that SLS is a potential task that is used in many daily activities, such as stair climbing, running and landing, the latter of which, especially on a single leg, is a common scenario in numerous sports, such as handball, football, volleyball and basketball. With this part of the game comes the potential for most lower limb injuries, such as ACL injuries, most of which have long-term consequences (Mather et al., 2013). In the literature, different landing tasks have been used to examine lower limb biomechanical variables, but one study compared lower limb biomechanics during drop landing, vertical drop jumps and forward jump landing with a vertical jump (Cruz et al., 2013). The authors found significant differences in knee valgus moment among the tasks, which might indicate the importance of using different tasks in order to examine this particular issue. However, it has also been found that non-contact ACL injury mechanisms require multi-directional manoeuvres (Olsen et al., 2004). Therefore, a single-leg squat and single-leg multi-directional landing will be used herein, to analyse kinematics and kinetics in the main study.

2.7.1 Single-leg Squat (SLS):

In many sporting activities, SLS motion involves repeating positions that require controlling the lower limbs and pelvis in all three planes (Zeller et al., 2003). Controlling the lower limbs helps prevent unlikely motions such as dynamic knee valgus, thus preventing injuries. In a study conducted between uninjured and injured ACL subjects, the authors aimed to differentiate between kinematic variables during SLS tasks. Injured male subjects performed SLS with lesser hip external rotation and more varus angle than uninjured subjects (Yamazaki et al., 2009a). On the other hand, injured female subjects performed SLS with higher knee valgus than uninjured subjects. It has been suggested, using 3D video analysis, that SLS tasks are related to movement such as landing and cutting (Stensrud, Myklebust, Kristianslund, Bahr, and Krosshaug, 2011). For instance, Alenezi et al. (2014) found strong correlations in knee abduction between SLS and running (r = 0.70), and moderate correlations between SLS and cutting (r = 0.54) (Alenezi et al., 2014). As a result, it would be beneficial to analyse lower limb biomechanics and investigate how it relates to the gluteal muscles during SLS. The SLS has emerged frequently to assess lower extremity alignment in general, and to determine the relationship between the gluteal muscles and knee kinematics in particular (Zeller, et al., 2003; Caliborne et al., 2006; Wilson et al., 2008; Herrington, 2013; Nakagawa et al., 2014; Willy and Davis, 2011). Destifano et al. (2009) reported gluteal muscles recruited more than 50% maximum voluntary isometric contraction during squatting (64% and 59% of gluteus maximus and gluteus medius, respectively), while Zeller et al. (2003) carried out a study and concluded that knee abduction increased during SLS, which made it reasonable for it to increase during activities such as landing.

From the literature, there are different protocols for using SLS as a screening task. Caliborne et al. (2006) instructed their subjects to squat 60° in six seconds; however, it is unclear how this was achieved, as it is difficult to be measure visually. To avoid fatigue, the examiner gave 2 minutes between each squat. Another study by Zeller et al. (2003) instructed subjects to stand on their dominant leg, cross their arms over their chest, squat down as far as possible and return to a single-leg standing position, without losing their balance, for five seconds. Nonetheless, the rest of the time was not mentioned in their study, so it is not possible to consider the effect of fatigue. Another study measured knee flexion with a goniometer while practicing and then asked participants to squat between 45° and 60° for 5 seconds (Herrington, 2013). Yamazaki et al. (2010) instructed their participants to cross their arms over their chest and perform a half squat while remaining balanced, with the duration of the squat being ten seconds or less. Subjects performed two single-leg half squats with both injured and uninjured legs, while subjects in the control group performed squats with the dominant leg. DiMattia et al. (2005) were more specific in their SLS method, ensuring that the arms were in a standard position (straight out in front of the subject at 90°), the contralateral leg was positioned at 45° hip flexion and 90° knee flexion off the ground and each SLS, lasting six seconds, limited to 60° knee flexion for the dominant leg. Therefore, there is a range of methodologies for SLS tasks (Claiborne et al., 2006; Hollman et al., 2014; Stickler, Finley, and Gulgin, 2015; Willson et al., 2006). In another study, Wyer et al. (2010) instructed their participants to squat down as far as possible and return to a single-leg stance without losing their balance, as they believed this better represented a clinical setting.

While the SLS is a good representative of other tasks and can be used at different activity levels, it might not be representative enough for the athletic population when looking to identify biomechanical or neuromuscular ACL risk factors, since it is a low-demand task.

2.7.2 Landing Task:

Non-contact ACL collapse has been linked to single-leg landing. Previous research studies have also examined the role of the hip muscles in controlling the lower extremities to prevent knee injuries (Homan et al., 2013, Hollman et al., 2009, Stearns et al., 2013), albeit they used double-leg landing, which, it has been argued, reduces the demands of the task in comparison to single-leg landing. Likewise, it does not represent the common mechanism of ACL injuries (Hewett et al., 2005). Single-leg landing has been correlated with hip muscle strength in several studies (Jacobs and Mattacola, 2005; Lawrence et al., 2008; Suzuki, Omori, Uematsu, Nishino, and Endo, 2015), and it has been found that there is a greater mean knee abduction angle at 14.3° during a single-leg hop landing task (Jacobs and Mattacola, 2005) in comparison to double-leg landing, which was 3.0° (Hollman et al., 2013). Similar results have also been reported in the literature (Myklebust et al., 1998; Pappas et al., 2007). A study by McNair and Prapavesis (1999), for instance, presents their normative data on vertical ground reaction forces during landing from a jump. The study tested 234 subjects performing a jump from a 30-cm height. The subjects were categorised by sex, activity level and sport played. No significant differences in peak vertical ground reaction force were noted in any of the three categories. Zhang et al. (2000) found that knee flexion increased as the landing height increased from 46°, 48° and 53° for 30 cm, 50 cm and 70 cm, respectively, and 52°, 56° and 63° for 32 cm, 62 cm and 103 cm in height, respectively. However, the exact instructions given to the participants for landing were not mentioned. In addition, this knee flexion angle increase could be a common strategy used to attenuate ground reaction forces upon impact.

It has been reported that a soft landing results in increased knee flexion angle, while an erect landing decreases the knee flexion angle (DeVita and Skelly, 1992). Thus, an erect landing results in higher ground reaction force. The relationship between landing and peak ground reaction force has been reported by Hewett et al. (2005). Moreover, Myer et al. (2005) reported that adolescent players with ACL injuries had a 20% greater ground reaction force in comparison to healthy players. These studies conclude that landing with a higher ground reaction force may increase the risk of ACL damage.

Side-to-side (frontal plane) movement has been counted as the most common manoeuvre that leads to ACL injuries in sports (Besier et al., 2001). It seems to be that this movement is performed most often at high speeds and when avoiding an opponent during competition. These movements increase the likelihood of uncontrolled movements and thus increase the risk of injury. Therefore, Suzuki et al. (2015) used side medial landing from a 20-cm box to assess knee kinematics. The study found significant correlations in this regard. Moreover, a control study carried out by Ortiz et al. (2011) aimed to compare the landing mechanism between non-injured women and women with ACL reconstruction. All subjects performed side-to-side hops across two lines marked 30 cm apart on two individual force plates. The study revealed that knee angle was similar between injured and non-injured subjects. It would be using a step as increasing the difficulty of the task could affect the kinematics. Another study was carried out with university football and basketball players in single medial and lateral drop landings from a 13.5 cm step (Ford et al., 2006). The study compared dynamic frontal plane excursion between genders. During medial landing, higher ankle eversion was noted in females than in males, which may cause an increase in the valgus load on the knee. The study also found that females had greater knee abduction angles, knee frontal plane excursion and hip frontal plane excursion during both types of landing.

To sum up, several studies in the literature have focused recently on SLS and landing tasks to analyse the biomechanical variables of lower limbs, in order to investigate the risk factors behind ACL injuries and interventions that reduce injury rates, not only to save time but also to save money. Gluteal muscles' function in dynamic knee valgus has been investigated; however, conflicting results have been reported regardless of the different methodological tools used. Therefore, a systematic review might be important, in order to search through the topic area and understand the quality of studies that have investigated the influence of gluteal muscles on dynamic knee valgus. This could help in the drawing conclusions regarding gaps in the current research, using search and selection methods precisely and including studies.

2.8 Synthesised Systematic Review:

The aim of conducting a systematic review is to understand the quality of the studies in prior literature and to reach some conclusion regarding the next appropriate research questions, plug gaps in current understanding where current evidence allows. Previous literature has used different tasks, populations and methodologies, which make a systematic review in this area an important undertaking, in order to establish a global view. Furthermore, previous reviews aimed to provide a background to and rationale for ACL injuries by reviewing risk factors, interventions and screening tools. Thus, it was difficult to clearly identify well defined gaps in knowledge without systematically reviewing the literature.

2.8.1 Methodology:

This structured literature review was conducted in accordance with PRISMA (preferred reporting items for systematic reviews and meta-analyses) statement guidelines. The PRISMA statement provides a set of systematic strategies for conducting electronic and/or manual searches, screening and excluding potential studies, with reasons, and finally including relevant studies. As elaborated previously by Liberati et al. (2009), the PRISMA statement guideline is suitable for summarising evidence from studies evaluating outcomes in healthcare interventions.

Search Strategy:

Primary studies investigating the EMG activity of either the G Max or the G Med in landing and squatting involving dynamic knee valgus motion, especially hip adduction, internal hip rotation and knee valgus or abduction, were searched in four electronic bibliographic databases: MEDLINE (the Library of Medicine and National Institutes of Health) via PubMed, SPORTDiscus, PEDro (Physiotherapy Evidence Database) and Web of Science. These databases were suitable for the searches, as they play host to articles on sports medicine, exercise science, physical education, biomechanics, physiology, coaching, injury prevention, rehabilitation, nutrition and recreation. The databases were searched from inception to April 2017, using search strings structured using two Boolean operators (AND and OR): (gluteus maximus OR gluteus medius) AND (function OR activity OR activation) AND (dynamic knee valgus OR valgus collapse OR medial knee displacement) AND (electromyography OR EMG) AND (hip abduction strength OR hip extension strength) AND (Land OR squat). Additional searches were performed through a manual bibliographic exploration of relevant articles retrieved from the electronic search strategy.

Study identification:

Electronic database searches returned a total of 142 potentially relevant citations (Fig. 2), which were exported into the EndNote reference manager (version X7, for Mac) to help screen out duplicates, which were removed through the manual screening of citation titles, authors, publication dates, journal volumes and issue numbers or digital object identifiers (DOI). Citations were screened manually based on titles and abstracts, and irrelevant citations were eliminated. A bibliographic hand search of the full-text articles and relevant reviews for additional citations otherwise missed during the electronic database searches yielded six additional potentially relevant citations. A total of 12 full-text articles were retrieved for further eligibility evaluation, based on the inclusion and exclusion criteria outlined below.

The inclusion criteria were: active, healthy participants, hip muscle strength assessment, hip muscle activation assessment and kinematic and kinetic analysis. Studies were selected if they had investigated the relationship between dynamic knee valgus and both gluteal muscle strength and activation during landing or squat tasks. Furthermore, studies in which hip-knee kinematic or kinetic data were captured using 3D cameras or motion analysis were also included. Finally, only published studies were included in the study, though no restrictions were in place regarding publication date or use of the English language. However, the exclusion criteria were: studies that examined participants with pathology or any previous history of injuries. Moreover, studies were assessed the biomechanics before and after a design intervention protocols. After applying the inclusion and exclusion criteria above, eight of the 12 studies were excluded, meaning that four studies were included in this systematic literature review for qualitative synthesis (Fig. 2.4).

Assessment of methodological quality and risk of bias:

A modified version of the Downs and Black checklist was used to evaluate the quality of the methods and risk of bias for all included articles (Hart et al., 2016). This tool is suitable for evaluating both randomised and non-randomised studies and shows good interrater ($r = 0.75$) and test-retest reliability ($r = 0.88$) (Downs and Black, 1998). A maximum of 15 scores were included in the modified version (see Appendices 1). Two independent PhD. students reviewed the included articles (Z.N and A.A). A score of 12 or more suggests high methodological quality, while 10-11 suggests moderate quality and less than a 10 score suggests low quality (Munn et al., 2010).

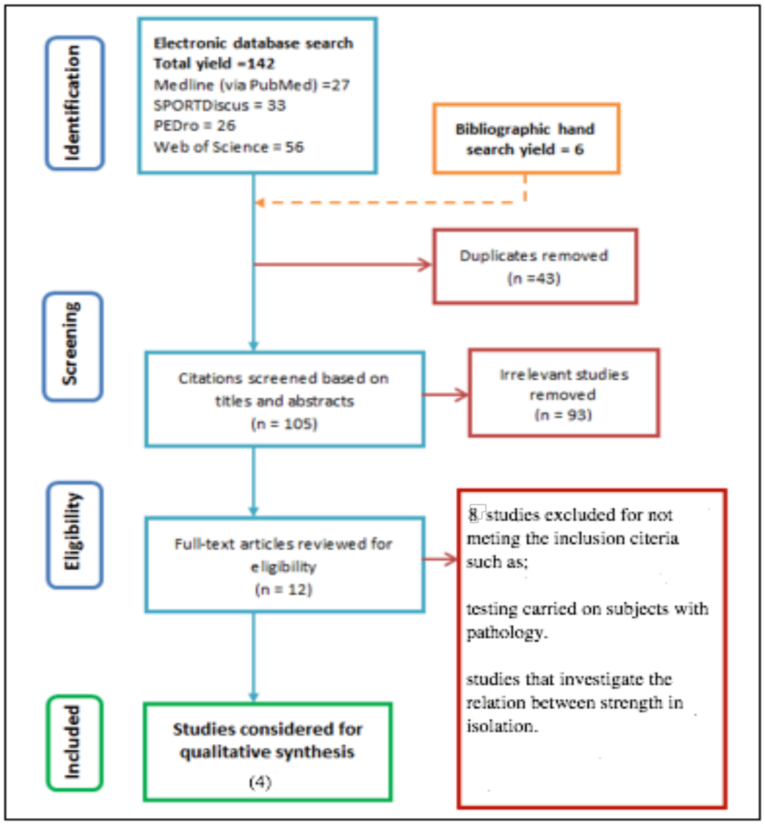


Figure 2.4 Prisma flow chart for study identification, screening, eligibility evaluation and inclusion.

Results:

Four studies (Homan et al. 2013; Hollman et al. 2013; Hollman et al. 2014; Nguyen et al. 2011) with a total of 216 participants (80 men, 136 women) were included in the study. The characteristics of the included studies are summarised in Table 2.1. Hip abduction and extension strengths were measured isometrically, using a hand-held dynamometer in all studies. Two studies administered single-leg squat tests to the participants (Nguyen et al. 2011; Hollman et al. 2014), and the remaining two studies administered double leg jump-landing tasks (Hollman et al. 2013; Homan et al. 2013).

All four studies demonstrated some relationship between EMG activity and the knee valgus angle, though the relationship was variable. Hollman et al. (2013) demonstrated that hip extensor strength and the recruitment of G Max are both associated with frontal knee motions during a dynamic weight-bearing task. Hollman et al. (2014) demonstrated that an increased G Med, hip rotation and abduction enhanced the recruitment of G Max, which then correlated well with an increased knee abduction angle. Homan et al. (2013) also demonstrated no relationship between G Max or G Med activity and dynamic knee valgus motion.

Table 2. 1 Characteristics of included studies:

Author, date	N	Subject (sex and age)	Methodology (hip strength test, EMG activity capture, amplitude presentation)	Tasks included in the study	Correlation reported
Hollman et al. 2014	41	Healthy physically active women (18 to 36 years) Strong group (21): age (23.8±1.8 yrs), height (1.682±0.071 m), mass (61.3±8.2 kg) Weak group (20): age (24.4±2.9 yrs), height (1.611±0.071 m), mass (61.3±9.6 kg)	Hip extension and abduction strength measured using hand-held dynamometer G max and G med recruitment were Mean EMG activity were normalized to % MVIC.	5-repetition single-leg squat tests 3-dimensional hip and knee kinematics during the task were captured using 3-dimensional hip and knee angles measured at the completion of the eccentric phase of each squat	No correlation reported between lower limb biomechanics and hip muscle strength or G max or G med EMG However, partial r was significant between G Max and knee abduction angle (0.35)

Table 2.1 Continued…

Nguyen et al. 2011	60	Men (n=30): (age = 23.9 ± 3.6 years, height = 1.785 ± 0.099 m, mass = 82.0 ± 14.1 kg) Women (n=30): age= 22.2 ± 2.6 years, height = 162.4 ± 6.3 cm, mass = 60.3 ± 8.1 kg)	Dynamometer torque data were recorded as the maximum peak torque obtained from 3 MVIC trials each for hip abduction and hip extension. G Max and G Med EMG amplitudes normalized to % MVIC	5-repetition single-leg squat tests Kinematic data for the pelvis, thigh, shank, and foot measured by electromagnetic sensors	No correlation reported between lower limb biomechanics and hip muscle strength or G max or G med EMG

Table 2.1 Continued…

Author, date	N	Subject (sex and age)	Methodology (hip strength test, EMG activity capture,	Tasks included in the study	Correlation reported
Hollman et al. 2013	40	Healthy female volunteers (18-36 years) (height = 1.65 ± 0.06 m, body mass = 63.1 ± 8.5 kg, mean BMI = 23.2 ± 2.8 kg/m2)	Isometric hip extension strength measured by hand-held dynamometer G Max activity was conducted bilaterally. EMG signals were normalised and expressed as % MVIC	3-repetition maximum vertical double jump-landing task 3D kinematic data were collected using motion analysis system and infrared digital camera at a sampling rate of 100 Hz.	Hip extension strength with knee abduction (r = .21) Gluteus maximus EMG with knee abduction (r = .13)
Homan et al. 2013	75	Healthy physically active volunteers. (height = 1.65 ± 0.06 m, body mass = 63.1 ± 8.5 kg, mean BMI = 23.2 ± 2.8 kg/m2)	Isometric hip abduction and external rotation strength were measured first in a randomised order using hand-held dynamometer EMG signals for G Max and G Med were normalised and expressed as % MVIC	5-repetition double-leg jump landing task Hip-knee kinematics during the double-leg jump landing task were recorded with 3D motion-capture system.	No correlation reported between lower limb biomechanics and hip muscle strength or G max or G med EMG

2.8.2 Discussion:

The role of gluteal muscles during functional tasks:

The results of the systematic review showed that the EMG activation of the gluteal muscle is associated with dynamic knee valgus during landing and squatting tasks. However, the relationship appeared weak and varied, which could be explained most obviously by the different methodologies used. Moreover, differences in activation level exist between weak and strong groups, albeit no significant differences are reported between the two groups.

As the gluteal muscles are the group responsible for working eccentrically to control excessive adduction and internal rotation during functional movement (Neumann, 2010), all four studies measured muscle strength isometrically. Neumann (2010) claims that the G Max is better at producing external rotation force compared to others. Moreover, the G Med has the greatest moment arm for producing abduction compared to the gluteus minimus and tensor fascia latae (Neumann, 2010). As a result, the hip tends to be rotated internally and adducted during landing (Powers, 2010), though the gluteal muscles try to elongate, putting the hip in position, which can improve their force capacity (Neumann, 2010).

Homan et al. (2013) investigated the influence of hip strength on gluteal activation and knee kinematics through 82 healthy participants during double-leg jump-landing tasks. The study noted no differences between the weak and strong groups in knee abduction motion, although the former showed greater muscle activation (Homan et al., 2013). Using the same task, Hollman et al. (2013) examined hip extension strength and G Max activation, which have both been associated with frontal knee motion. Hip motion in the transverse plane may be correlated with knee abduction motion (partial $r = 0.72$). However, double-leg landing tasks are not representative of those activities during which ACL injuries occur.

Hollman et al. (2014) aimed to examine the relationship between hip muscle strength and G Max and G Med activation in 41 females during a single-leg squat task. The study found that G Max activity may modulate with knee frontal motion (partial $r = 0.35$). All previous studies measured muscle strength isometrically using a hand-held dynamometer, which might explain the differences in their results. Another study also measured knee kinematics with female participants during single-leg step downs (Hollman et al., 2009). Twenty healthy women participated, to identify the relationship between hip muscle strength, function and knee abduction angle. The study found G Max activity has more of an effect on the knee abduction angle during a stepping down task ($r = -0.45$), thereby indicating that increasing the recruitment of the G Max limits knee abduction motion. However, Hollman's studies included female participants and did not use tasks representative of those during which ACL injury occurs, which make the data incomparable to studies involving more challenging tasks such as single-leg landing or SLS. Moreover, strength measurement was assessed again using isometric contraction, though hip muscles work eccentrically to control lower limb kinematics during dynamic motion. Nguyen et al. (2011) reported no correlation when investigating lower limb kinematics among 60 participants (30 males and 30 females) during an SLS task. However, the study used electromagnetic sensors to measure kinematics, and isometric strength was used to assess hip muscle strength.

Several studies in the literature have investigated the relationship between gluteal strength or activity in isolation (Claiborne et al., 2006, Itoh et al., 2016, Jacobs and Mattacola, 2005, Malloy et al., 2017, Suzuki, Omori, Uematsu, Nishino, and Endo, 2015b, Willson et al., 2006). Suzuki et al. (2015), for instance, used side medial landing from a 20-cm box to assess knee kinematics in 43 college basketball players (20 males and 23 females). The study did indeed find significant negative correlations between isometric hip muscle strength and knee kinematics, though it would have been better if isokinetic muscle strength had been measured instead of the isometric alternative, to give more understanding on how the muscles work concentrically and eccentrically to control landing.

A study carried out by Claiborne et al. (2006) investigated the relationship between both the concentric and the eccentric strength of hip muscles and the knee abduction angle during single-leg squats. The study found a significant correlation between concentric hip abduction and the knee valgus angle ($r= -0.37$, $R^2= 0.13$). However, this correlation was still weak, which might be explained by other factors contributing to dynamic knee valgus. Lawrence et al. (2008) found hip external rotators had no effect on knee frontal and sagittal plane angles. In previous studies, conflicting results might be related to differences in methodological tools, since Claiborne et al. (2006) used concentric and eccentric force, whereas Willson et al. (2006) and Lawrence et al. (2008) used isometric force, which makes it difficult to correlate isometric strength with dynamic movement. This concept has been supported by several studies (Jacobs et al., 2007, Sigward et al., 2008, Willson and Davis, 2008). Furthermore, another study investigated the effect of hip extensor and abductor strength on the knee valgus angle during double-leg landing (Stearns, Keim, and Powers, 2013). After four weeks of a strengthening intervention programme, peak isometric strength was measured. The study found that pre- and post-programme, the hip extensor and abductor peaks increased, while the knee abduction peak decreased (6.8 – 5.6°). This small but detectable change could be due to the task involving double-leg landing in comparison to single-leg landing, and so using more difficult tasks such as single-leg landing may have led to greater extent of change and statistical significance. Jacobs and Mattacola (2005) reported that during single-leg landing, females have a greater peak knee abduction angle than males (p = 0.07, Effect size = 0.62), although no significant differences have been reported between them in eccentric peak torque terms. Test positioning could be the reason for this lack of difference in peak strength, as the examiners used a standing position to test abduction eccentric force, which might lead to more effort in the contralateral side, in order to stabilise the body (Jacobs and Mattacola, 2005). In addition, a previous study carried out on 47 participants aimed to investigate the relationship between hip abductor and external rotator strength with knee motion during lateral step downs (Norcross et al., 2009). The study found isometric hip abductors correlated negatively with frontal plane knee angles (r= -0.37, p= 0.01), and no correlation was found in eccentric hip abductors and eccentric and isometric hip external rotator strength. However, the study used a task that was not challengeable enough and did not include 3D motion or EMG to analyse biomechanical parameters.

A cross-sectional study, carried out by Souza and Powers (2009), aimed to determine whether PFP females differ in hip kinematics, strength and activation, compared to a control group. It has been reported that dynamic knee valgus is a biomechanical risk factor for PFP syndrome (Hewett et al., 2005, Ireland, 1999). The study showed that PFP females had more internal hip rotation compared to the control group. Furthermore, the PFP group had 14% less hip abductor strength and 17% less hip extensor strength; however, there was a significant increase in gluteal maximus recruitment in the PFP group (Souza and Powers, 2009). Nonetheless, this cross-sectional study did not reflect the exact cause and effect of the relationship, and so abnormality may exist because of knee pathology, as supported by Mascal et al. (2003). In addition, the study cannot be generalised, as it was carried out with young females and it used isometric strength, which does not reflect the nature of dynamic tasks.

2.8.3 Conclusion:

As the aim of the review was to reach a conclusion regarding the research question that this PhD should pursue. The systematic review demonstrated that there are no clear results in relation to the influence of gluteal muscles on dynamic knee valgus. This might be because of the limited research carried out in this area, or because of differences in the methods used previously in the literature. However, the systematic review helped define the precise gaps related to the topic. It would be beneficial to understand the relationship between gluteal muscle strength and activation, and how they relate to dynamic knee valgus. Consequently, better interventions could be implemented to prevent ACL injuries. However, all previous studies have tested a single task and none have considered all of the movement directions that are appropriate. Furthermore, no investigation to date has investigated the possible link between isokinetic gluteal muscle strength (eccentric and concentric force generation) and muscle activation with dynamic knee valgus through single-leg squats and single-leg multi-directional landings in active, healthy participants.

The suggested position of injury includes movement that mostly occurs in the frontal plane, such as hip adduction and knee abduction (Markolf et al. 1995, McLean et al. 2004, Hewett et al. 2005). Furthermore, the majority of the studies examined a bilateral landing task, which does not reflect sport-specific movement adequately, as noted by Myer et al. (2008) and Edwards et al. (2010). Moreover, a bilateral test may not identify limitation of unilateral function and may miss the important unilateral events commonly necessary during sport (Myer et al. 2011, Augustsson et al. 2006).

Furthermore, knee injuries mostly occur when the body weight shifts on a single leg (Olsen et al., 2004). Investigation of single leg performance is more challenging for the movement control strategies, matches the reality of sport, and the muscles must produce more load than in bilateral tasks (Myer et al. 2004, Olsen et al., 2004).

In previous literature different tasks, participant groups, dependent variables and methodologies were used, which makes a systematic review in this area of importance for making a decision about evidence-based practice (Gopalakrishnan and Ganeshkumar, 2013). . It should also keep clinicians updated and help them judge the advantages and disadvantages of any intervention. Moreover, it may help guide the direction of future research and act as evidence to compare and contrast recent findings (Moher, Liberati, Tetzlaff, and Altman, 2009).

2.9 Gap in the Literature:

First of all, the literature has described the use of hip muscle strengthening exercises for patients with knee pathology; however, there is a conflict regarding the direct impact of hip strength on dynamic knee valgus, regardless of differences in the methodology. Some equivocal results are stated in the literature regarding gluteal muscle activation, with some reporting a relationship between this and dynamic knee valgus, while others state there is no association between them. Others have investigated the relationship between strength or activity and lower limb biomechanics during single-leg squats or single-leg multi-directional landing, but no one has looked at both (at the same time) except for a small selection of authors (Hollman et al., 2009, 2014, 2013, Homan et al., 2013, Nguyen et al., 2011). In addition, no one has assessed concentric and eccentric gluteal muscle strength to examine the relationship.

Most of the studies have measured gluteal muscles' strength isometrically, using an isokinetic dynamometer or a hand-held dynamometer, except for a small cohort (Claiborne et al., 2006, Jacob and Mattacola, 2005), who measured hip strength isokinetically, with Claiborne et al. investigating the impact of hip and knee strength on knee abduction during single-leg squats only, though neither included muscle activation in their study.

All previous studies have used the test as a single task and nobody has looked at all directions of single-leg landing. Furthermore, the majority of the studies examined bilateral landing tasks that do not adequately reflect sport-specific movements, and no investigations to date have linked the relationship between gluteal muscle function and dynamic knee valgus through single-leg squats, forward single-leg landing and single-leg side landing (medial and lateral).

Finally, there no investigations have studied lower extremity kinetics and kinematics while landing on a single leg from different directions and over increasing vertical landing heights and horizontal landing distances. Using SLS and multi-directional single-leg landing tasks in the current study might add to the literature, thereby providing a better understanding with regard to how hip muscle strength and/or gluteal EMG activity influence lower limb biomechanics, especially dynamic knee valgus and associated factors. In addition, the study could help in the current injury prevention measures, namely through G Max and G Med strengthening, or neuromuscular training could contribute to and mimic dynamic knee valgus. Furthermore, it is important to examine the reliability of the methodology used to collect data, as a reliable method would provide consistent measurements in which the clinician or researcher can be confident when seeking to detect differences.

Chapter 3

Methodology:

In this chapter, the biomechanical methods and strength measurements used in the abovementioned studies will be discussed. Moreover, reliability studies are presented in this chapter. Before investigating the project's main goal, it is important to conduct the study using appropriate measurement procedures that give consistent and reproducible values with small measurement errors. The outcome of this reliability study will provide a clearer understanding of the methods used.

3.1. The Reliability of Isometric and Isokinetic Strength Testing Hip Abductor and Extensor Muscles, using the Biodex System

Study Aims:

The study aims to investigate the within-day and between-days reliability of hip abductors and extensors (G max and G medius) during isokinetic (concentric and eccentric) and isometric action. Furthermore, it will investigate the correlation between the isometric and isokinetic results. The study's hypothesis is that no correlation exists between isometric and isokinetic results; therefore, isokinetic muscle testing will be included in the main study, as the muscles work concentrically and eccentrically during dynamic tasks such as SLS and single-leg multi-directional landing. The findings of this study might be important for evaluation and rehabilitation.

3.1.2 Methods:

Fifteen recreationally active, healthy students (eight males and seven females) from Applied Sports Science and Sport Rehabilitation courses were recruited to take part in the study. The male age was 22.50 ± 3.34 years, height 178.12 ± 7.6 cm and mass 81.70 ± 8.76 kg, and for females age 22.20 ± 3.93, height 169.85 ± 7.08 cm and mass 66.68 ± 7.489 kg. Subjects were physically active and had performed at least 30 minutes of physical activity three times a week on a regular basis over the previous six months (Munro and Herrington, 2011). Healthy participants over 18 years of age and able to extend, abduct and externally rotate his/her hip joint were included in this study. Informed consent must be submitted before testing, which was approved by the College of Health and Social Care Research Ethics Panel at University of Salford. However, subjects with any pathology or minor pain in a lower limb that may affect testing, or a history of major lower-limb injury such as a broken bone, torn ligaments or dislocation over the previous six months, or being unable to give informed consent, were excluded from the study.

Ethical approval was gained for the reliability studies from the University of Salford's Research, Innovation and Academic Engagement Ethical Approval Panel (HSCR 15/19). All participants gave informed consent before participating in the study (Appendices 2).

3.1.3. Study Procedure:

For each participant, data on isometric and isokinetic muscle strength for both concentric and eccentric contractions were taken for both legs. Two different tests were carried out using the Biodex system, namely hip abduction and hip extension. Subjects were asked to wear training clothes, and testing was carried out at two different times on the same day they attended, and then subsequently after one week (Maffiuletti et al., 2007). A maximum of 45 minutes was needed for testing. To avoid any possible injuries, participants were asked to warm up for five minutes on a stationary bike, before starting the test (Woods et al., 2007). Moreover, to become familiar with the tests, participants had the chance to practise every test with sub-maximum efforts (Requiao et al., 2005). Also, to avoid muscles overloading, two minutes' rest were given to participants between each test (Reid et al., 2007). Participants were asked to perform three repetitions of three strength sets. For the isokinetic set, 60°/sec was the testing speed (Boling, Padua, and Creighton, 2009, Julia et al., 2010, Myer, Sugimoto, Thomas, and Hewett, 2013, Widler et al., 2009). According to Perrin (1993) more concentric power can be produced at slower isokinetic speed, and as the eccentric speed increases, the force will remain the same or might increase slightly. Testing orders were randomised. After isokinetic testing, participants were asked to perform three maximal voluntary isometric contractions for three seconds, with 30 seconds' rest period between them. Up to five minutes were given between different muscle group tests. All measurements were carried out by the one examiner and peak torque was corrected automatically for gravity by Biodex software, by taking a static torque at approximately 45° of the hip extension test, and 30° for the hip abduction test prior to testing.

Hip abduction test (Figure 3.1):

Subjects were placed in a side-lying position by reclining the backrest of the testing chair to allow a fully flat position, with the non-testing leg stabilised using straps around the thigh and above the ankle. The dynamometer's axis of rotation of movement was aligned from the medial to the greater trochanter. The lever arm provided resistance against the lateral aspect of the mid-thigh. For the isokinetic test, the average range of motion when testing hip abduction ranged from 0° to 30°. For the isometric test, the hip was in a neutral position.

Hip extension test (Figure 3.2):

For isokinetics, subjects were placed in a supine position on the testing chair by reclining the backrest with straps around their waist to stabilise the body. The dynamometer's axis of rotation of movement was aligned to the level of the greater trochanter. The lever provided resistance against the posterior mid-thigh. The average range of motion was from approximately 45° hip flexion to 0° for the isokinetic test. For the isometric test, subjects were in a prone position on a testing bed, with the lever arm providing resistance against the posterior mid-thigh as well (Figure 3.3).

3.1.4. Statistical Analysis:

All statistical analyses were performed using SPSS for Mac (version 20). Peak torque was selected as the outcome parameter. The means of three trials from the first and second sessions were used for within-day reliability, and the means of the first and third sessions were applied for between-days reliability. Intra-class correlation (ICC) was measured and ICC values interpreted according to Coppieters et al. (2002): Poor <.40, Fair .40 to .70, Good .70 to .90, Excellent >.90. However, the ICC appeared to be easy to interrupt and in isolation could not provide a full picture of reliability. Therefore, confidence intervals (CIs) and standard errors of measurement should be measured with ICC. A low SEM with a high ICC indicates good measurement reliability. The advantage of SEM is that it presents the unit of measurement by providing an estimate of measurement accuracy (Denegard and Ball, 1993), which allows the researcher to compare results with other research studies. Denegard and Ball (1993) performed the calculation for SEM using the following formula:

$$SD\ (pooled) * (\sqrt{(1-ICC)})$$

Finally, a Pearson's correlation coefficient (r) was used to determine the relationship between isometric, concentric and eccentric. The strength of the correlation coefficient was illustrated by the interpretation used in the study by Hopkins et al. (2009): small (0.1-0.3), moderate (0.3-0.5), large (0.5-0.7), very large (0.7-0.9) and extremely large (0.9-1.0).

Figure 3.1 Hip Abduction Test

Figure 3.2 Isokinetic Hip Extension Test

Figure 3.3 Isometric Hip Extension Test

3.1.5 Results:

Table 3.1 contains ICC values for 95% CI. The ICC values for both the hip extension and abduction were higher for the within-day (0.62 – 0.98) than the between-days (0.59 – 0.93) results. Therefore, all tests showed good to excellent ICC apart from right eccentric hip abduction for both within-day and between days, and left concentric hip abduction was found to be fair by ICC. Table 3.2 contains the mean SEM values for isokinetic hip extension for both the concentric and the eccentric elements, which ranged from 10.82 Nm to 13.99 Nm. The isokinetic hip abduction for both concentric and eccentric aspects ranged from 4.91 Nm to 9.92 Nm, isometric hip extension was 10.56 Nm to 11.97 Nm and isometric hip abduction was 7.18 Nm to 8.65 Nm)

Table 3.3 contains a Pearson's correlation coefficient showing a strong correlation between concentric and eccentric forces, ranging between r = 0.67 – 0.70 for hip abduction and r = 0.87 -0.95 for hip extension. However, no significant correlation was noted between isometric and concentric or eccentric forces.

Table 3. 1 Intra-class correlation (ICC) and 95% confidence intervals (CIs):

Test	Within-day		Between days	
	Right	Left	Right	Left
Isometric hip extension	0.98 (0.96 - 0.99)	0.96 (0.88 - 0.98)	0.93 (0.80 - 0.97)	0.89 (0.68 - 0.96)
Isometric hip abduction	0.95 (0.86 - 0.98)	0.96 (0.89 - 0.98)	0.91 (0.74 - 0.97)	0.91 (0.74 - 0.97)
Concentric hip extension	0.91 (0.73 - 0.97)	0.78 (0.36 - 0.92)	0.88 (0.64 - 0.96)	0.78 (0.37 - 0.92)
Concentric hip abduction	0.94 (0.84 - 0.98)	0.92 (0.77 - 0.97)	0.76 (0.39 - 0.92)	0.62 (0.21 - 0.87)
Eccentric hip extension	0.93 (0.83 - 0.97)	0.85 (0.56 - 0.95)	0.88 (0.50 - 0.94)	0.83 (0.52 - 0.94)
Eccentric hip abduction	0.62 (0.40 - 0.84)	0.78 (0.34 - 0.92)	0.59 (0.49 - 0.83)	0.90 (0.70 - 0.96)

Table 3. 2 Mean strength score and standard error of measurements (SEM/SEMs%):

Test		Within-day		Between days	
		Right	Left	Right	Left
Isometric hip extension	(N-m)	162.59 (11.14 / 6.8%)	168.08 (11.05 / 6.5%)	161.87 (10.56 / 6.5%)	169.93 (11.97 / 7.0%)
Isometric hip abduction	(N-m)	127.05 (8.44 / 6.6%)	124.59 (8.65 / 6.9%)	130.02 (7.18 / 5.5%)	129.34 (7.97 / 6.1%)
Concentric hip extension	(N-m)	233.02 (13.65 / 5.8%)	218.44 (10.82 / 4.9%)	226.48 (12.97 5.7%)	223.00 (11.99 / 5.3%)
Concentric hip abduction	(N-m)	106.41 (9.92 / 9.3%)	110.64 (8.48 / 7.6%)	99.48 (9.07 / 9.11%)	101.46 (6.55 / 6.4%)
Eccentric hip extension	(N-m)	251.40 (13.99 / 5.5%)	233.98 (11.56 / 4.9%)	239.90 (12.71 / 5.2%)	232.01 (12.25 / 5.2%)
Eccentric hip abduction	(N-m)	138.21 (5.32 / 3.8%)	130.72 (5.30 / 4.0%)	133.75 (5.81 / 4.3%)	129.70 (4.91 / 3.7%)

Table 3. 3 Pearson's correlation coefficient between isometric, concentric and eccentric forces:

Test		Right	Left
Hip Extension			
		***r* value *P* value**	***r* value *P* value**
Isometric vs. Concentric	Peak Torque	0.54 (0.37)	0.20 (0.46)
Isometric vs. Eccentric	Peak Torque	0.52 (0.48)	0.29 (0.29)
Concentric vs. Eccentric	Peak Torque	0.95 (0.005)	0.87 (0.005)
Hip Abduction			
Isometric vs. Concentric	Peak Torque	0.31 (0.26)	0.12 (0.65)
Isometric vs. Eccentric	Peak Torque	0.45 (0.08)	0.21 (0.43)
Concentric vs. Eccentric	Peak Torque	0.70 (0.003)	0.67 (0.006)

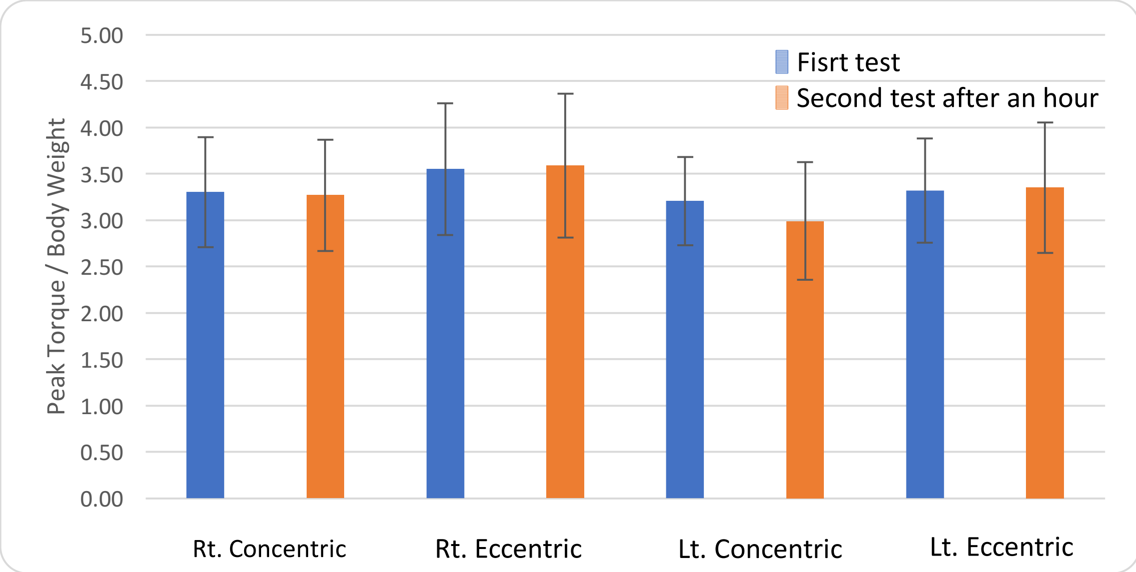


Figure 3.4 Within-day tests for concentric and eccentric hip extensions

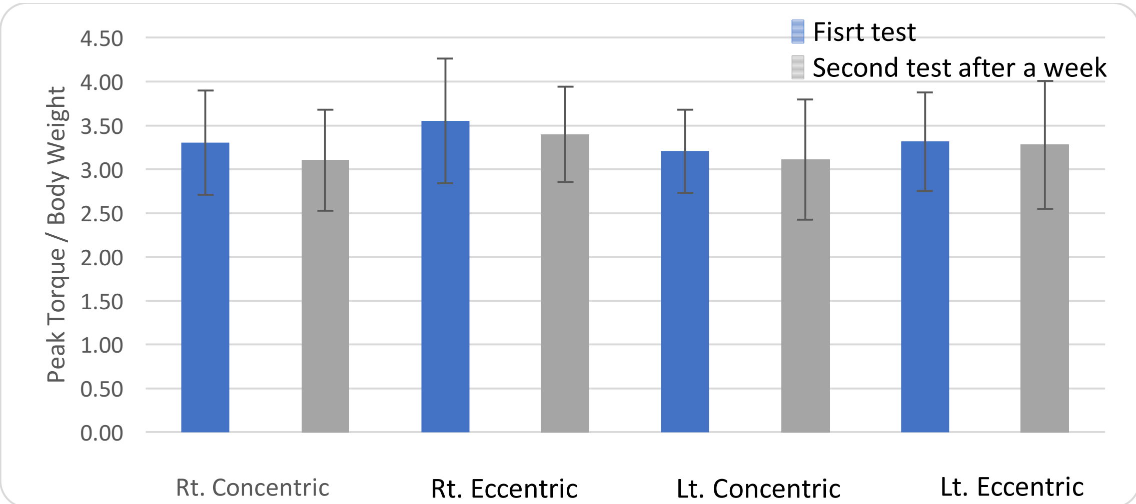


Figure 3.5 Between-days tests for concentric and eccentric hip extensions

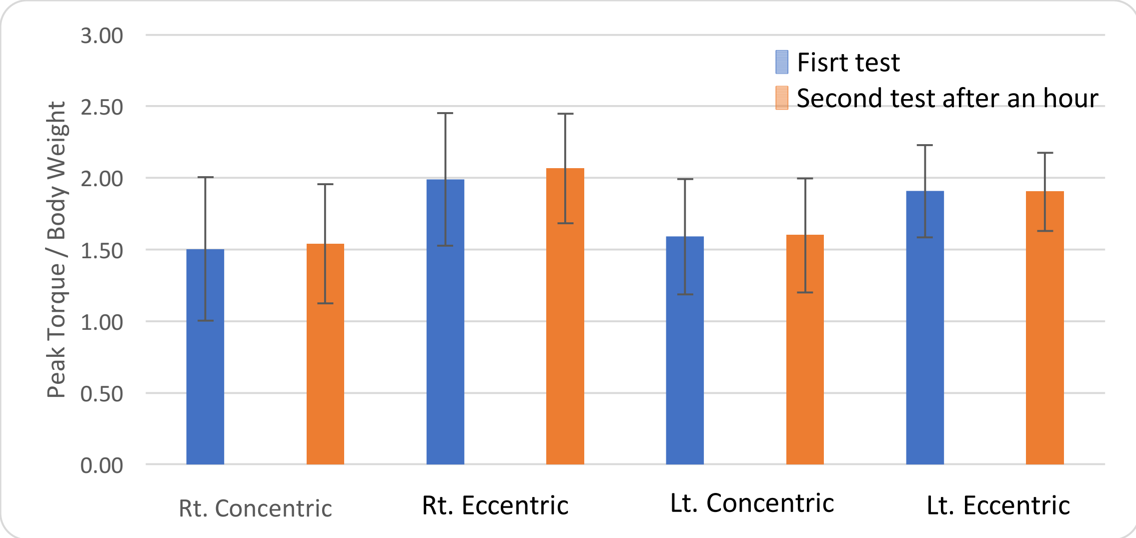


Figure 3.6 Within-day tests for concentric and eccentric hip abductions

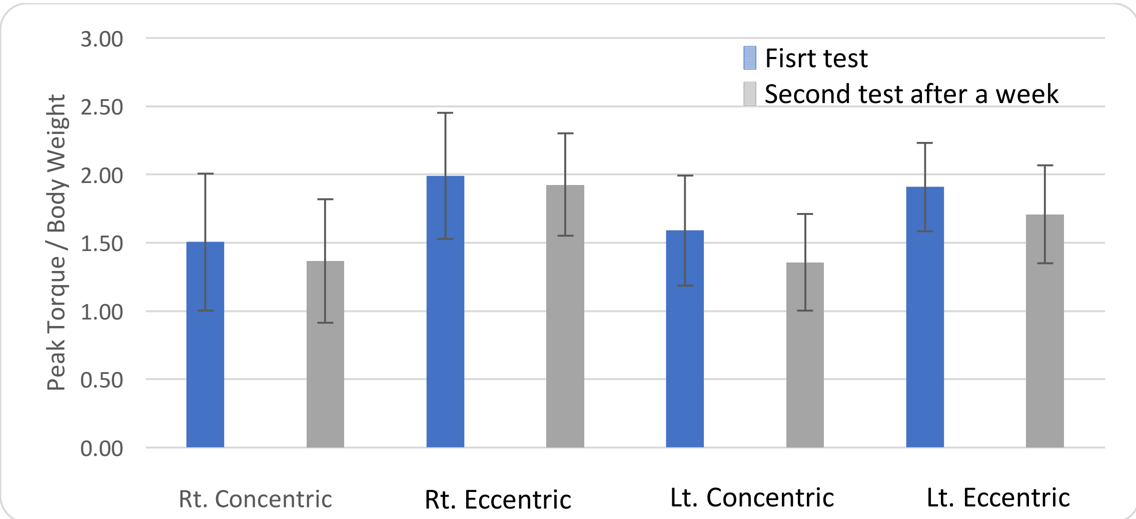


Figure 3.7 Between-days tests for concentric and eccentric hip abductions

3.1.6 Discussion:

The main objectives of the study were to examine the within-day and between-days reliability of isometric and isokinetic concentric and eccentric hip abductor and extensor muscle strength, using the Biodex system in recreationally active students. In addition, the study examined the correlation between isometric and isokinetic strength, in order to choose one type of strength test to include in the main study. In the present investigation, the ICC value for both hip extension and abduction were higher for the within-day (0.62 – 0.98) than the between-days (0.59 – 0.93) results. In general, isokinetic testing showed fair to high ICC values ($0.59 \leq$ ICC ≤ 0.94), and for isometric tests, ICCs were found to be high ($0.89 \leq$ ICC $\leq$ 0.98). Similar results were found in the literature (Arokoski et al., 2002, Claiborne et al., 2009, Julia et al., 2010, Meyer et al., 2013). Meyer et al.'s (2013) study aimed to standardise a method to assess hip joint strength, using the Biodex system with 18 participants asked to perform isometric and isokinetic (eccentric and concentric) contractions of the hip muscles. The study reported ($0.68 \leq$ ICC ≤ 0.97); however, the examination was on the non-dominant leg only, whereas the current study concerned both legs. Another study carried out by Claiborne et al. (2009) aimed to find out the test-retest reliability of isokinetic hip torque, using Biodex as well. Thirteen healthy adult subjects participated in two experimental tests over a week. Isokinetic hip torque speed was 60˚/sec. High torque reliability was found in concentric hip flexion (right and left), extension (right), as well as eccentric hip flexion (right) and extension (right and left) ICC range (0.81–0.91). Moderate torque reliability ICC (0.49–0.79) was found for concentric hip extension (left) and eccentric hip flexion (left) (Claiborne et al., 2009). Moreover, a study by Julia et al. (2010) reported ICC values of 0.62 to 0.94 for concentric hip extension at 60°/s, tested in a supine position, though ICC values of only 0.9 were found by Arokoski et al. (2002b). These two studies also showed different findings for hip extensions, which could be explained by the varied range of motions tested. On the other hand, although Arokoski et al. (2002b) and Claiborne et al. (2009) used different methodologies for testing isometric hip abduction, for instance standing versus supine, they found comparable results. However, a study carried out by Widler et al. (2009) stated that a side lying position is the most reliable and valid method for measuring isometric hip strength and comparing with different positions using a hand-held dynamometer. Furthermore, Cahalan et al. (1989) and Laroche et al.'s (2008) studies reported ICC values of 0.96 and 0.95, respectively, with both of them carrying out the test twice within 48 hours. The best ICC values reported on active, healthy adults, but two studies reported high ICCs in hip osteoarthritis and hemiparesis subjects (Arokoski et al., 2002, Eng, Kim, and MacIntyre, 2002).

Previous studies have been carried out with different populations and have demonstrated the reliability of hip strength tests. Eng et al. (2002), for instance, reported the high reliability of isokinetic hip extensions for hemiparetic old participants (ICC = 0.9). In addition, high reliability has been reported for older participants with hip osteoarthritis (Arokoshi et al., 2002), with ICC values ranging from 0.84 to 0.98 for hip abduction/adduction and flexion/extension.

The current study shows fair to high ICC values within and between days, along with low SEM. For all measurements, SEM ranged from 4.1 Nm to 13.9 N). These values have been reported by Meyer et al. (2013) and Claiborne et al. (2009), though Claiborne used a standing position to measure isokinetic strength. In addition, the current study shows no relationship between isometric and concentric or eccentric forces, apart from right isometric hip extension and concentric (r = 0.54), eccentric (r = 0.52). This result supports the difficulty in correlating isometric force with dynamic movement – a concept n supported by several studies (Sigward et al., 2008, Willson and Davis, 2008, Jacobs et al., 2007). Therefore, isokinetic strength testing will be included in the main study.

The study was not without its limitations. First, testing orders were randomised, which may have affected the results. However, in order to reduce this impact on the results, it was decided to examine isokinetic strength and then isometric. In addition, rest time was given to all participants and they were always asked if they were ready to be examined or not, to prevent muscle fatigue. Another factor that might affect the results was motivational status; however, the examiner tried to provide all the encouragement needed during all testing trials. Moreover, it was difficult to make the lever arm run parallel to the participant's leg, although all participants showed good ability through the range of motion, and the results show medium to strong reliability in most testing positions. This study used 60° per second as angular velocity with a range of motion from 0 to 30° and 45° for hip abduction and extension, respectively. However, as stated earlier in the literature review, for hip extension and abduction, 60° per second is a good representation of both the concentric and eccentric capabilities of each muscle group (Boling et al., 2009).

3.1.7 Conclusion:

The current study has demonstrated that certain variables show good to high reliability, along with low standard errors of measurement. These results are relevant to those undertaking hip strength measurements. In addition, isokinetic muscle testing did not correlate with isometric testing, apart from in the right hip extension test. Therefore, isokinetic testing will be included in the main study for the strength measurement, as hypothesised.

3.2 The Repeatability of Lower Limb Biomechanical Variables and the Electromyography Activity of Gluteus Maximus and Medius during Single-leg Squats and Single-leg Multi-directional Landing

As stated earlier, this section will test the reliability of the methods used in this thesis. This will also include the repeatability and measurement errors for the 3D assessment of motion and EMG activity produced by G Max and G Med. For 3D motion, the marker placement error has the most influence in reliability studies (Ferber et al., 2006). The use of a surface EMG provides information on muscle activation, pattern and degree of activation, though this information is highly variable regardless of the study design and methods used in terms of normalisation, filtration and electrode positioning.

Unlike the isokinetic strength reliability study, this study aimed to investigate the consistency of marker and electrode placement, as the former accounted for the greatest errors in 3D motion analysis (Ford et al., 2007). Nevertheless, strength testing was carried out within and between-days, because in the main study the subjects will be tested in two different labs and on two different occasions within a week. It will therefore be important to ensure that strength does not vary over 7 days, as strength testing always comes after testing the biomechanical variables.

3.2.1 Study Aims:

- To investigate the consistency of the biomechanical variables (kinematics and kinetics) and EMG activity of G Max and G Med during single-leg squats and multi-directional single-leg landings within 24 - 48 hours.
- To establish SEMs during these tasks for active, healthy participants.

3.2.2 Methods:

Ten recreationally active and healthy students from Applied Sports Science and Sport Rehabilitation courses were recruited to take part in the study (five males and five females). The male age range was 28.2 ± 1.1 years, height 169.12 ± 5.2 cm, and mass 76.70 ± 9.58 kg, and for females their age was 27.2. ± 4.4, height 163.36 ± 5.17 cm, and mass 61.46 ± 5.46 kg. Subjects were physically active and had performed at least 30 minutes of physical activity three times a week on a regular basis over the previous six months (Munro and Herrington, 2011). Healthy participants over 18 years of age who are able to hop, land and squat on a single leg were included in this study. Informed consent was obtained before testing and approved by the College of Health and Social Care Research Ethics Panel at the University of Salford. Subjects with any pathology or minor pain in a lower limb that may affect testing, a history of a major lower limb injury such as a broken bone, torn ligaments or dislocation over the last previous months or those unable to give informed consent were excluded.

3.2.3 Study Procedure:

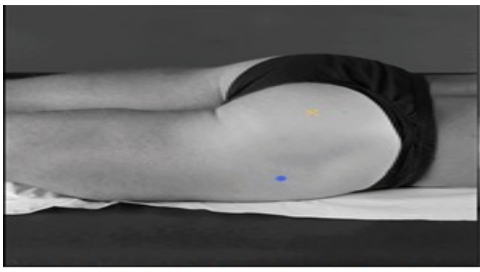
3D protocol:

Fifteen cameras (Qualisys, Gothenburg, Sweden), sampling at 240 Hz in a motion analysis system, and one force platform (AMTI BP400600, USA), sampling at 1200 Hz and embedded into the floor, were used to collect kinematic and kinetic lower limb variables during different tasks. At the beginning of the procedure, 40 reflective markers were attached to both lower limbs' anatomical landmarks. These markers were used to define the anatomical reference frame and the joint rotation centres. Reflective markers were placed as follows: anterior superior iliac spines, posterior superior iliac spines, iliac crest, greater trochanters, medial and lateral femoral condyles, medial and lateral malleoli, posterior calcanei and the heads of the first, second and the fifth metatarsals in both limbs were placed on a standard training shoe. Finally, four rigid plates, each one consisting of four reflective markers, were attached to the antero-lateral aspect of the thigh and shank. The calibration anatomical systems technique (CAST) was used to determine each segment's movement during the trial (Cappozzo et al., 1996). The static trial position was calibrated as a subject's neutral alignment from standing over the force plate with weight distributed equally over both lower limbs. At this is a trial, it was checked and reflective markers viewed by the cameras and Qualysis software was used for the identification of tracking and anatomical markers prior to extraction to the post-processing software. Following the satisfactory capture of all static markers, the anatomical markers were detached, keeping only 28 as tracking markers (16 markers over four cluster plates, eight markers attached to standard shoes and four markers on ASISs and PSISs). These clusters were fastened securely to the antero-lateral aspect of the thigh and shank of both legs. Manal et al. (2000) found that using rigid clusters is the optimal configuration, compared to individual skin markers (Manal et al., 2000). Both static and tracking markers are illustrated in Figure 3.8. The participants wore standard lab shoes (New Balance, UK), to control the shoe-surface interface.

Electromyography Data Capture:

Gluteus maximus (G Max) and gluteus medius (G Med) activity was recorded using a Noraxon Desktop DTS system (Noraxon USA Inc., model 586 Tele Myo DTS Desk Receiver), synchronised with the 3D capture and sampled at 1500 Hz. A disposable, self-adhesive Noraxon surface electrode was fixed over the muscle (parallel to the muscle fibre). A surface electrode was prepared and placed, following the recommendations of the SENIAM project (SENIAM, 2011). Before electrode placement, the skin was shaved and cleaned using isopropyl alcohol. For the G Max, a surface electrode was placed at 50% between the sacral vertebrae and the greater trochanter, parallel to the muscle fibres (Figure 3.9). For the G Med, a surface electrode was placed at 50% from the line of the iliac crest to the greater trochanter (Figure 3.10). The electrodes were carefully placed, with consideration given to the orientation of the direction of muscle fibres, and then fixed using tape. The EMG signal was tested after placement of the electrodes during straight leg raising in extension and abduction. Before the testing session, participants maximum voluntary isometric contractions (MVIC) for each muscle were obtained so that data could be normalised. An MVIC for both G Max and G Med was performed according to the standard clinical testing methods defined by Norcross et al. (2010). For the gluteal maximus, participants were prone with their hips extended 10°. For the gluteal medius, participants were on their side, with hips and knees in neutral and the hip at 10° abduction to establish MVC. Five minutes of low-intensity warm up and stretching exercises were performed by the subjects before testing.

Figure 3.8 Static (left) and tracking (right) marker placement



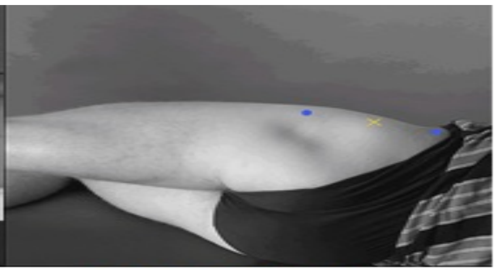


Figure 3.9 Gluteus Maximus Application **Figure 3.10 Gluteus Medius Application**

Functional tasks:

Single-leg Squat (SLS): In this study, subjects were instructed to stand in the middle of the force plate. The subject was then asked to squat down as far as possible to at least 45° and no greater than 60° while keeping the trunk as upright as possible (Zeller et al., 2003). Each trial was conducted over a period of five seconds, using an electronic counter. The first count was to initiate the squat, the third count indicated the lowest point of the squat and the fifth count indicated the end of the trial (Herrington, 2014) (Figure 3.11a).

Forward Land (FL): Subjects were instructed to stand on a step (30cm height) and then stand on one leg and jump forward off it, landing on the force platform. The distance between the step and the platform was 30cm. The subject was asked to practise the task three times, to become familiar with it (Figure 3.11b).

Side land with the force platform from inside of knee (SML): Subjects were instructed to stand on a step (30cm height) and then, starting from a single-leg position, to perform a medial jump onto the force platform. The distance between the step and the platform was 30cm. The subject was asked to practise the task before testing, to become familiar with it (Figure 3.11c).

Side land with the force platform from outside of knee (SLL): Subjects were instructed to stand on a step (30cm height). Starting from a single-leg position, they were asked to perform a lateral jump onto the force platform. The distance between the step and the platform was 30cm. The subject was asked to practise the task before testing, to become familiar with it (see Figure 3.11d).

Figure 3.11 a SLS task

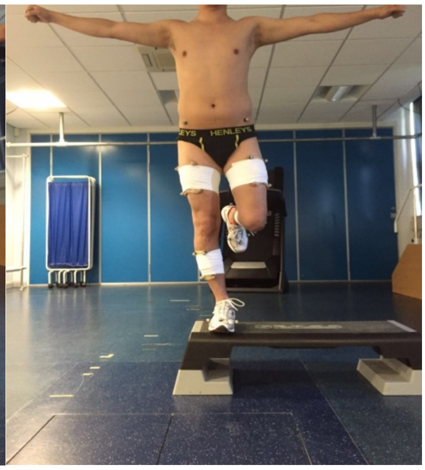


Figure 3.11 b FL task

Figure 3.11c SML task

Figure 3.11d SLL task

Data Processing:

Visual3D motion capture software (Version 4.21, C-Motion Inc., Rockville, MD, USA) was used to analyse and calculate the kinetic and kinematics data. A Butterworth 4th order bi-directional low-pass filter was used to filter the motion and force plate, with cut-off frequencies of 12 Hz and 25 Hz for kinematics and kinetics, respectively, and based on residual analysis (Yu et al., 1999). Joint kinematics were calculated using an X-Y-Z Euler rotation sequence (X = flexion-extension, Y = abduction-adduction or varus-valgus and Z = internal- external rotation). Joint kinetic data were calculated using three-dimensional inverse dynamics, and the joint data were normalised to body mass and presented as an external moment. Six degrees of freedom were determined by using CAST during all dynamic tasks (Cappozzo et al., 1996). Before dynamic trials, a static capture was obtained by standing on the force plates, where the cameras could view all attached markers, and the Qualisys software prior to extraction for post-processing software. The positions of these anatomical markers offered reference points for identifying bone movement through only the tracking markers set during the movement trials. As can be seen in Figure (3.12), the model had seven rigid segments attached to the joint. Each segment is considered to have six variables that describe its position (three variables describe the position of the origin, and three variables describe the rotation) in 3D space. Specifically, three variables describe the segment translation along three perpendicular axes (vertical, medial-lateral and anterior-posterior), and three variables describe the rotation about each axis of the segment (sagittal, frontal and transverse). Each segment of the pelvis, thigh, shank and foot was modelled to determine the proximal and distal joint/radius. The hip joint centre is automatically calculated by using ASIS and PSIS markers according to the regression equation from Bell, Brand and Pedersen (1989).

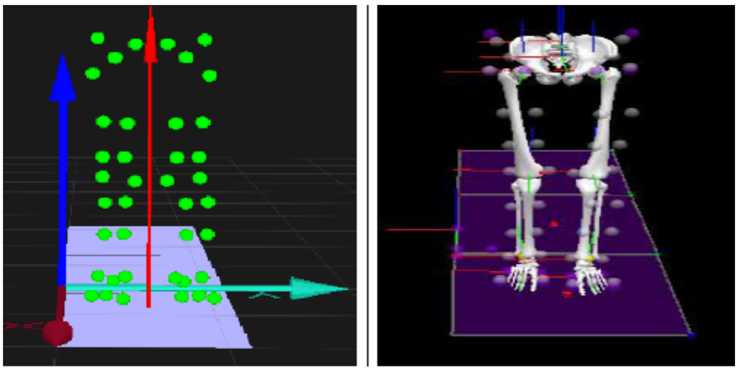


Figure 3.12 Qualysis static model (left), visual 3D (right)

Muscle activity profiles were used to determine any changes in the EMG activity of the muscles 100 milliseconds before landing and two seconds after initial contact, or until the participant was fully balanced; however, during squatting, EMG activity was recorded during ascent and descent until the subject fully extended the knee. EMG activity for each landing and squatting trial was synchronised with the task data. EMG activity from the muscles (G Max, G Med) during these tasks was analysed as raw signals in Visual3D. The data were bandpass filtered (25 – 450) and a 60-Hz notch filter was applied. A moving root mean squared (RMS) algorithm was used with a 100-millisecond window to produce a linear envelope. Corresponding muscle activity during the MVIC was also analysed in the same manner, and each set of data for each muscle, and each activity, was exported as a text file to Microsoft Excel 2010 (Microsoft, Washington, USA). The mean average of each maximum muscle activity from the three trials was taken, and this maximum was normalised to the corresponding MVIC.

Main outcome measures:

- Maximum vertical ground reaction force.
- Maximum hip and knee joint moment.
- Maximum joint angle (hip and knee in frontal, sagittal and transverse planes).
- EMG activity for gluteus maximus and gluteus medius normalised to maximum voluntary isometric contraction (MVIC).

3.2.4 Statistical Analysis:

All statistical analyses were done with SPSS for Mac (version 20). The means of the three trials from the first and second sessions were used for within-day reliability, and the means of the first and third sessions employed for between-day reliability. Intra-class correlation (ICC) was measured, and ICC values were interpreted according to Coppieters et al. (2002): Poor <.40, Fair .40 to .70, Good .70 to .90, Excellent >.90. However, the ICC appears to be easy to interrupt and cannot present a full picture of reliability in isolation. Therefore, confidence intervals (CIs) and standard errors of measurement should be measured with the ICC. A low SEM with a high ICC indicates good measurement reliability. The advantage of the SEM is that it presents a unit of measurement by providing an estimate of measurement accuracy (Denegard and Ball, 1993), which then allows the researcher to compare the results with other research studies. Denegard and Ball (1993) state the calculation of SEM using the following formula:

SD (pooled) * (√ (1-ICC)).

3.2.5 Results:

Kinematic and kinetic variables:

Table 3.4 shows the results for the SLS task. For the right limb, ICC values for kinematics and kinetics ranged between 0.77 and 0.98. The SEM values ranged between 0.14 and 3.84° for angles, and 0.02 to 0.17 Nm-Kg for moments. The lowest ICC was found in knee abduction moment, and the highest ICC was found in the hip adduction angle. For the left limb, ICC values ranged between 0.61 and 0.99. The SEM value ranged between 0.38 and

2.31° for angles and 0.01 and 0.08 Nm-Kg for moments. The lowest ICC was found in knee abduction moment and the highest in knee flexion moment.

Table 3.5 shows the results of the FL task. For the right limb, ICC values for kinematics and kinetics ranged between 0.52 to 0.99. The SEM values ranged between 0.73 to 3.81° for angles, and 0.06 to 0.32 Nm-Kg for moments. The lowest ICC was found in internal hip rotation moment, and the highest ICC was found in the internal hip rotation angle. For the left limb, ICC values ranged between 0.55 and 0.98. The SEM value ranged between 0.06 and 2.59° for angles and 0.07 and 0.48 Nm-Kg for moments. The lowest ICC was found in hip flexion moment and the highest in the knee abduction angle.

Table 3.6 shows the results of the SML task. For the right limb, ICC values ranged between 0.47 and 0.95. The SEM values ranged between 0.85 and 4.24° for angle, and 0.10 and 0.24 Nm-Kg for moments. The result for hip flexion moment was unreliable. For the left limb, ICC values ranged from 0.43 to 0.97. The SEM values ranged between 0.93 and 6.28° for angles and 0.20 and 0.40 Nm-Kg for moments. The lowest ICC was found in the knee abduction moments.

Finally, Table 3.7 shows the results for the SLL task. Hip flexion moments for both limbs showed the lowest results. For the right limb, ICC values ranged between 0.64 and 0.97. The SEM values ranged between 1.20 and 4.16° for angles and 0.05 and 0.18 Nm-Kg for moments. For the left limb, ICC values ranged from 0.50 and 0.98. The SEM values ranged between 1.18 and 3.18° for angles and 0.08 and 0.13 Nm-Kg for moments. The lowest ICC was found for knee abduction moments.

Table 3.4 Single-leg Squat (Intra-class Correlations (ICCs), Confidence Intervals (CIs), Mean and SEM):

Variables	ICC (95% CI)	Mean	SEM	CV%
Right				
Joint angles(°)				
Hip Int. Rotation	0.97 (0.89 – 0.99)	12.00	1.20	10
Hip Adduction	0.98 (0.917 - .095)	17.69	1.12	6.33
Knee abduction	0.97 (0.875 – 0.99)	-0.94	0.14	14.83
Moments (Nm/Kg)				
Hip Int. Rotation	0.81 (0.415 – 0.95)	-0.43	0.06	13.95
Hip Adduction	0.81 (0.406 – 0.94)	-1.08	0.11	10.18
Knee Abduction	0.77 (0.401 – 0.93)	-0.10	0.04	40
Left				
Joint angles(°)				
Hip Int. Rotation	0.99 (0.964 – 0.99)	13.18	1.67	5.08
Hip Adduction	0.93 (0.743 – 0.98)	14.98	1.86	12.41
Knee Abduction	0.98 (0.906 – 0.99)	-1.77	0.38	21.46
Moments (Nm/Kg)				
Hip Int. Rotation	0.65 (0.106 – 0.82)	-0.53	0.06	11.32
Hip Adduction	0.63 (0.366 – 0.89)	-1.11	0.08	7.20
Knee Abduction	0.61 (0.331 – 0.88)	-0.15	0.06	40

- For hip flexion, knee flexion biomechanics and vertical ground reaction force results, see appendices (6).

Table 3.5 Forward Landing (Intra-class Correlations (ICCs), Confidence Intervals (CIs), Mean and SEM):

Variables	ICC (95% CI)	Mean	SEM	CV%
Right				
Joint Angles(°)				
Hip Int. Rotation	0.99 (0.95 – 0.99)	11.51	2.89	7.73
Hip Adduction	0.79 (0.45 – 0.84)	11.58	2.83	24.4
Knee Abduction	0.91 (0.69 – 0.97)	-1.40	0.73	52.14
Moments (Nm/Kg)				
Hip Int. Rotation	0.52 (0.21 – 0.75)	-0.44	0.08	3.52
Hip Adduction	0.71 (0.48 - .081)	-1.63	0.13	7.97
Knee Abduction	0.81 (0.50 – 0.94)	0.09	0.06	41.23
Left				
Joint Angles(°)				
Hip Int. Rotation	0.96 (0.85 – 0.99)	14.19	2.38	9.72
Hip Adduction	0.91 (0.69 – 0.97)	8.76	1.22	13.92
Knee Abduction	0.98 (0.93 – 0.99)	-0.79	0.06	7.59
Moments (Nm/Kg)				
Hip Int. Rotation	0.41 (0.23 – 0.67)	-1.18	0.17	14.40
Hip Adduction	0.49 (0.28 – 0.60)	-1.81	0.26	14.36
Knee Abduction	0.61 (0.34 – 0.88)	0.18	0.07	38.33

- For hip flexion, knee flexion biomechanics and vertical ground reaction force results, see appendices (6).

Table 3.6 Side Medial Landing (Intra-class Correlations (ICCs), Confidence Intervals (CIs), Mean and SEM):

Variables	ICC (95% CI)	Mean	SEM	CV%
Right				
Joint Angle (°)				
Hip Int. Rotation	0.93 (0.74 – 0.98)	11.95	2.13	17.82
Hip Adduction	0.91 (0.68 – 0.97)	11.85	1.62	13.67
Knee Abduction	0.95 (0.80 – 0.98)	-1.57	0.61	38.21
Moments (Nm/Kg)				
Hip Int. Rotation	0.47 (0.18 – 0.73)	-0.79	0.10	12.65
Hip Adduction	0.65 (0.37 – 0.89)	-1.52	0.18	11.84
Knee Abduction	0.64 (0.36 – 0.89)	0.34	0.15	44.11
Left				
Joint Angle (°)				
Hip Int. Rotation	0.93 (0.74 – 0.98)	16.08	3.79	11.13
Hip Adduction	0.97 (0.87 – 0.99)	10.36	0.93	8.97
Knee Abduction	0.62 (0.23 – 0.89)	-1.34	0.58	43.28
Moments (Nm/Kg)				
Hip Int. Rotation	0.62 (0.43 – 0.89)	-1.17	0.23	19.6
Hip Adduction	0.48 (0.25 – 0.64)	-1.94	0.30	15.46
Knee Abduction	0.43 (0.17 – 0.75)	0.70	0.18	25.71

- For hip flexion, knee flexion biomechanics and vertical ground reaction force results, see appendices (6).

Table 3.7 Side Lateral Landing (Intra-class Correlations (ICCs), Confidence Intervals (CIs), Mean and SEM):

Variables	ICC (95% CI)	Mean	SEM	CV%
Right				
Joint Angle (°)				
Hip Int. Rotation	0.94 (0.77 – 0.98)	11.91	3.93	16.20
Hip Adduction	0.83 (0.46 – 0.95)	9.28	2.57	27.69
Knee Abduction	0.88 (0.59 – 0.97)	-2.38	0.87	36.55
Moments (Nm/Kg)				
Hip Int. Rotation	0.94 (0.76 – 0.98)	-0.85	0.05	5.88
Hip Adduction	0.89 (0.61 – 0.97)	-1.40	0.09	6.42
Knee Abduction	0.87 (0.57 – 0.96)	0.36	0.06	16.66
Left				
Joint Angle (°)				
Hip Int. Rotation	0.98 (0.90 – 0.99)	14.63	3.18	8.06
Hip Adduction	0.91 (0.67 – 0.97)	6.29	1.59	25.27
Knee Abduction	0.91 (0.69 – 0.97)	-1.45	0.73	50.34
Moments (Nm/Kg)				
Hip Int. Rotation	0.82 (0.43 – 0.95)	-1.12	0.08	7.14
Hip Adduction	0.87 (0.56 – 0.96)	-1.96	0.13	6.63
Knee Abduction	0.50 (0.14 – 0.84)	0.09	0.06	66.66

- For hip flexion, knee flexion biomechanics and vertical ground reaction force results, see appendices (6).

Electromyography activity data:

Table 3.8 shows the results of the SLS and all landing tasks. For SLS, ICC values ranged between 0.60 and 0.84. The SEM values ranged between 4.35 and 6.69%. For FL, ICC values ranged between 0.92 and 0.96, and the SEM values ranged between 2.01 and 2.79% in both the right and the left limbs. For SML, ICC values ranged between 0.92 and 0.97. The SEM values ranged between 1.49 and 2.86%. Finally, for SLL, ICC values ranged between 0.95 and 0.97, and the SEM of EMG activity ranged from 1.09 to 1.89% in both limbs.

Table 3.8 Electromyography activity data for gluteus maximus and gluteus medius normalised to MVIC (intra-class correlations ICCs), confidence intervals (CIs), Mean and SEM):

Single-leg Squat	**ICC (95% CI)**	**Mean**	**SEM/SEM%**
Right			
EMG (RMS) gluteus max.	0.60 (0.356 – 0.89)	55.35	5.41 / 9.77%
EMG (RMS) gluteus med.	0.84 (0.781 – 0.98)	61.74	4.35 / 7.04%
Left			
EMG (RMS) gluteus max.	0.66 (0.205 – 0.85)	49.80	6.69 / 13.43%
EMG (RMS) gluteus med.	0.64 (0.216 – 0.74)	52.03	5.65 10.85%
Forward Land			
Right			
EMG (RMS) gluteus max.	0.92 (0.73 – 0.98)	28.50	2.79 / 9.78%
EMG (RMS) gluteus med.	0.95 (0.83 – 0.98)	34.15	2.66 / 7.78%
Left			
EMG (RMS) gluteus max.	0.92 (0.71 – 0.97)	22.24	2.09 / 9.39%
EMG (RMS) gluteus med.	0.96 (0.86 – 0.99)	32.48	2.01 / 6.18%
Side Medial Land			
Right			
EMG (RMS) gluteus max.	0.96 (0.85 – 0.99)	28.71	1.49 / 5.18%
EMG (RMS) gluteus med.	0.97 (0.89 – 0.99)	35.45	1.68 / 4.73%
Left			
EMG (RMS) gluteus max.	0.95 (0.81 – 0.98)	25.79	1.60 / 6.20%
EMG (RMS) gluteus med.	0.92 (0.71 – 0.98)	33.53	2.86 / 8.52%
Side Lateral Land			
Right			
EMG (RMS) gluteus max.	0.96 (0.94 – 0.99)	30.65	1.09 / 3.55%
EMG (RMS) gluteus med.	0.97 (0.91 – 0.99)	34.17	1.89 / 5.53%
Left			
EMG (RMS) gluteus max.	0.95 (0.89 – 0.99)	27.03	1.42 / 5.25%
EMG (RMS) gluteus med.	0.97 (0.91 – 0.99)	34.35	1.45 / 4.22%

3.2.6 Discussion:

The purpose of this study was to examine repeatability, using 3D motion analysis to measure lower limbs' biomechanical variables and to examine the within-day reliability of the EMG activity of G Max and G Med muscles during SLS and multi-directional single landing tasks. The second aim was to establish SEM for all variables during these tasks in a healthy active group.

Depending on the literature, this study hypothesised that all biomechanical variables would show good ICC values. However, the results showed that angles were the most reliable data, not the vertical ground reaction force. Similar results found in the literature will be discussed later in this chapter. According to the author's knowledge, this is the first study to measure these variables during single-leg multi-directional landings.

<u>Kinematics and kinetics:</u>

For kinematic data, the results show that repeatability was good to excellent, ranging from .73 - .99 except for hip flexion and knee abduction angles during left side medial landings, which showed fair reliability at 0.53 and 0.62, respectively. The highest ICC value was reported for the left internal hip rotation angle during single-leg squats. The lowest was reported for the right knee flexion angle during side medial landings. For kinetics data, ICC values were lower than kinematics data, and reported fair to excellent reliability, ranging from 0.41 – 0.99, except for right hip flexion during side medial landing and both legs' hip flexion during side lateral landing, which showed poor reliability. The task's difficulty may play a role in the variability of the results concerning side landing. Moreover, the participants were active and healthy, which produces different performances and different ways of landing, and this may have had an effect on the results as well.

Previous studies have examined the reliability of lower limb biomechanics during different screening tasks, such as single-leg drop landing (Alenezi et al., 2014, DiCesare et al., 2015, Ortiz et al., 2007), single-leg squats (Alenezi et al., 2014, Nakagawa et al., 2014) and double-leg landing (Ford et al., 2007). Alenezi et al. (2014) investigated within-day reliability during single-leg landing and SLS tasks. The study found average ICC values for landing and squatting of 0.9 and 0.87, respectively. However, the study used single-leg drop landing with arms crossed, to minimise the effect of the arms, whereas the current study did not give any instructions to the participants regarding arms. Crossing arms may not reflect the true picture during sport activities. Similarly, it might be that single-leg landing tasks are more dynamic than drop landing, which requires subjects to push off from the step. Similar results were found by Nakagawa et al. (2014) during SLS and Ford et al. (2007) during double-leg landing. However, Nakagawa used an electromagnetic tracking system (ETS), which is a six degrees of freedom measuring device that can track the 3D position and orientation using multiple sensors attached to different body segments to measure only kinematic data. In addition, Ford et al. (2007) used a double-leg drop, but the use of a single-leg task is related more to the ACL injury scenario (Hewett et al., 2005). Ortiz et al. (2007) found good reliability during single-leg drop jumps, shown for three trials' average for all kinematics (ICC $\geq$ 0.75) and kinetics (ICC $\geq$ 0.86), except the knee flexion angle (0.29). However, the study used a drop jump task from a 40-cm step, with only healthy females who engaged in fitness, jogging and weightlifting exercises. Moreover, the study did not measure the joint's moments and included only VGRF and contact time as kinetic data. Another study found good reliability (ICC ranged from .68 - .95) in knee kinematics and kinetics (Milner, Fairbrother, Srivatsan, and Zhang, 2012). Similar findings were reported during a vertical drop jump (ICC ranged from .59 to .92) (Ford et al., 2007). Nonetheless, both studies used landing jump or stop-jump tasks and analysed data from landing after a vertical jump. The current study used a horizontal hop from a 30-cm step, which might be more difficult.

Several factors might influence a reliability test, for example marker movement, marker re-application and the task's level of difficulty (Alenezi et al., 2014, Ford et al., 2007, Kadaba et al., 1989). In order to reduce variability within the study, the CAST model protocol has been used to improve anatomical relevancy and to reduce skin movement artefacts (Cappozzo et al., 1996). The advantage of this model is to improve anatomical relevance by attaching the markers to the centre of the segments rather than to the joints, as in Helen Hayes' model (Collins et al., 2009).

During all tasks, SEM values have been provided as reference values that may be useful for intervention outcomes. SEM ranged between 0.05° and 6.28° for joint angles and 0.05 and 0.48 Nm/kg for moments. According to Portney and Watkins (1993), SEM allows the clinician to be 68% confident that the true value lies within ± 1 SEM of a given value. In the current study, the higher SEM found in hip flexion angles might be explained by the greater range of motion in the sagittal plane compared to the frontal or transverse. Similar results were found by Alenezi et al. (2014). In contrast, Nakagawa et al. (2014) reported a lower SEM value for hip flexion angles during SLS, which might be due to the age of the participants in their study (21 ± 1.1 vs 27.7 ± 3).

EMG Activity:

Surface electromyography reliability was tested to ensure electrode fixation and to determine the consistency of the EMG activity of G Max and G Med during the tasks. All ICC values during landing tasks were excellent (ICC ≥ 0.92), and SEM values ranged from 1.09% – 6.69%. However, squatting showed less reliability. In the SLS task, high ICC was in the right G Med (0.84), and the lowest was in the right G Max (0.60). This could be explained by dynamic instability because of the associated movement while ascending and descending. Similar results were stated in the literature when testing G Med EMG activity during weight-bearing and non-weight-bearing exercises (Bolgla and Uhl, 2005). During weight-bearing exercises, ICC values ranged from 0.95- 0.96; however, in this study, the subject stood on one leg while testing the counter leg. Thus, the study tested the activity of the G Med during the exercises. During hopping tasks (forward, sideways and transverse), less reliability was found, ranging between 0.37 and 0.56 and higher SEM values of 30% - 41% (Distefano et al., 2009). The subjects completed eight repetitions of 12 therapeutic exercises. In this case, fatigue may have affected the results. Barton et al. (2014) reported moderate to excellent reliability, with ICC ranging from 0.64 – 0.92 for G Max and G Med, though reliability was measured from the MVIC only, not after normalising activity data from the functional tasks.

The literature includes single-leg squats and landing tasks in screening programmes (Zeller et al., 2003, Homan et al., 2013, Hollman et al., 2013, Hollman et al., 2014, Suzuki et al., 2015, Claiborne et al., 2006). Therefore, it is important to know the relationship between the variability of this outcome and the subject's performance and the methodology used. In the current study, the majority for ICC values for joints, moments and vertical GRF were higher in SLS and FL than in side single-leg landings, and the majority of ICC kinetic values were lower than kinematics, especially for both sides landing. Nonetheless, the side landing tasks showed poor results in hip flexion events, it has been stated that non-contact ACL injury mechanisms require multidirectional manoeuvres (Olsen et al., 2004). Apart from the poor results, this finding supports using these tasks as screening tools for the main study.

The current study has several limitations. First is the difficulty in controlling squat depth while trying not to lose balance. This could influence the trunk position and significantly affect the demand placed on the hip muscles. The variability of the results may be due to the dynamic nature of the tasks used in the study. Thus, differences in landing strategy with respect to the trunk position and centre of mass may increase variability, especially in kinetic data. Second, the testing shoes were standardised, which did not reflect the same shoes being worn in practice or suitability for the surface being played on. Furthermore, cross- talk may occur and affect EMG information. For instance, when measuring the G Med, the activation of tensor fascia latae and/or G minimus contributed to the action. This was solved by carefully following SENIAM guidelines for applying surface EMG electrodes and ensuring they were in the correct position. Finally, testing was carried out on active, healthy participants. Further research is needed on the timing of gluteal muscle activation, as it has been stated that G Med activation is delayed in PFPS subjects during running and stepping down (Barton et al., 2012). Moreover, more research is needed to investigate the relationship between gluteal muscle activity and lower limb biomechanics during squatting and landing on a single leg from different directions on active, healthy participants.

3.2.7 Conclusion:

This current study has determined that certain variables reveal good to excellent consistency with respect to low standard error measurement values, which might be relevant to others undertaking interventions in their studies. The current results will be used for the main study to explore the relationship between gluteal muscle EMG activity and lower limb kinetics and kinematics, and this will include muscle strength during concentric and eccentric phases to reflect dynamic tasks in real practice.

This chapter has determined that strength, 3D motion analysis and EMG activity data measurement techniques can be used later in this thesis to answer the study question. The techniques have shown generally good reliability and low measurement errors. Additionally, this chapter has described the method used for strength, 3D and EMG activity capture in detail, and this same method will be used in the main study of this thesis.

Chapter 4

Kinetics and Kinematics Variables of Single-leg Squat and Multi-directional Single-leg Landing

4.1 Introduction:

Having established the reliability of strength, 3D and EMG activity capture in Chapter 3, the aim of this chapter is to explore biomechanical variables during SLS and single-leg multi-directional landing, and to investigate the differences between legs, tasks and genders. Providing this information might help in understanding factors that could affect the relationship between dynamic knee valgus and hip neuromuscular control. The outcome of this chapter will demonstrate the differences, if indeed they exist, between limbs, tasks and genders.

Study Hypothesis:

- Hip (adduction and internal rotation) and knee (abduction) angles and moments are different between limbs and across all tasks.
- Hip and knee joint angles and moments during SML and SLL are greater than SLS and FL.
- Females in all tasks demonstrate higher knee abduction, hip adduction and internal hip rotation (angles and moments) compared to males.

4.2 Methods:

Participants:

Thirty-four (17 females and 17 males) active, healthy participants were recruited to participate in this study. Post hoc analysis will be calculated to achieve a power of 0.80 with a=0.05, and the effect size will determined dependent on the coefficient of determination, which will be found between variables in Chapter 7. Demographic information is listed in Table 4.1, for which the same inclusion and exclusion criteria were used as stated earlier in the reliability studies in Chapter 3, including participants who were healthy, active and without any lower limb injuries. A consent form was read and signed by all participants before taking part in the study. Ethical approval was gained for the reliability studies from the University of Salford's Research, Innovation and Academic Engagement Ethical Approval

Panel (HSCR 15/47). All participants gave informed consent before participating in the study (appendices 2).

Instrumentation:

Fifteen cameras (Qualisys, Gothenburg, Sweden) sampling at 240 Hz in a motion analysis system, and one force platform (AMTI BP400600, USA) sampling at 1200 Hz and embedded in the floor, were used to collect kinematic and kinetic lower limb variables during the different tasks. Gluteus maximus and gluteus medius EMG activity was recorded simultaneously using the 3D capture using Noraxon Desktop DTS system (www.noraxon.com) at 1500 Hz via a disposable self-adhesive Noraxon surface electrode fitted over the muscle (parallel to the muscle fibre). The same instruments, including filtration, calibration, marker list, training shoes, functional tasks and biomechanical model, were used as described in Chapter 3.

Table 4.1 Demographic information for all participants

Demographic		Number	Mean	SD
Age (years)	Female	17	25.71	4.48
	Male	17	26.93	3.82
	All	34	26.26	4.17
Height (cm)	Female	17	168.18	4.78
	Male	17	171.07	5.66
	All	34	169.48	4.87
Weight (Kg)	Female	17	64.18	7.28
	Male	17	69.79	6.63
	All	34	66.70	7.84

Statistical Analysis:

First, to check whether or not the data were normally distributed (parametric or non-parametric), a Shapiro-Walk test was used and diagrams evaluated (Malloy et al., 2017). Additionally, a descriptive analysis (mean and standard deviation) was done for each dependent variable in each functional task. For parametric variables, a paired t-test was used to examine differences between legs and biomechanical variables, and for non-parametric variables, a Wilcoxon Rank test was used (Edwards, Steele, Cook, Purdam, and McGhee, 2012). The level of significance was set at 0.05, data were not normally distributed if values were less than or equal to 0.05 and the mean value of three trials for each test were calculated to find differences in performance between legs. Gender differences were examined by an independent t-test for parametric variables and a Mann-Whitney U test for non-parametric variables.

A repeated measures one way analysis of variance (ANOVA) was used for parametric data, to explore differences in the kinematics and kinetics between tasks used in the study. However, for non-parametric values, a Friedman test was used to identify any differences. If significant differences were found, post hoc comparisons were performed using a pairwise t-test for parametric and a Wilcoxon-rank test for non-parametric variables with a Bonferroni adjustment. Partial eta squared was obtained to determine the effect size, using the guideline proposed by Cohen (1988) (0.01 = small, 0.06= moderate and 0.14 large).

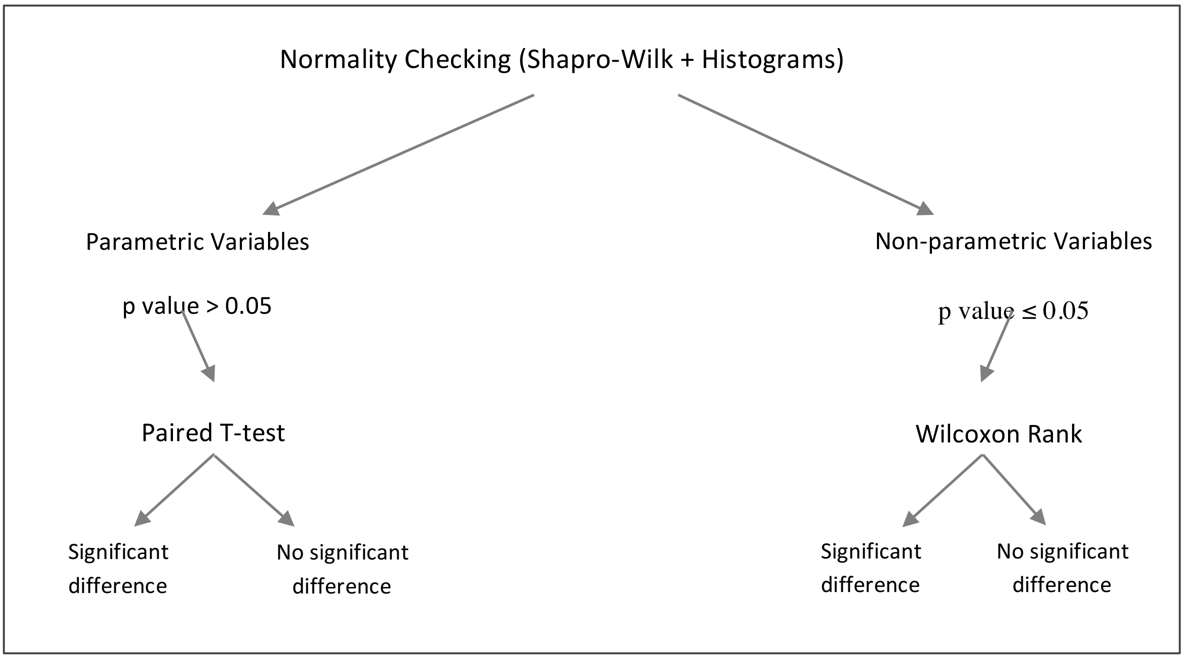


Figure 4.1 Statistical outlines for the differences study

4.3 Results:

The normality test (Shapiro-Wilk) for kinetic and kinematic variables revealed that all variables were normally distributed for both legs, apart from left knee abduction moment in SLS task, right knee abduction moment, left knee abduction moment, left internal hip rotation moment in FL task, right hip flexion moment, right knee abduction moment, left internal hip rotation moment and left knee abduction moment in SML. Finally, SLL data were revealed as normally distributed except for the right hip flexion moment, right knee abduction moment, left internal hip rotation moment and left knee abduction angle. See Appendices (3) for the normality test for all variables.

A paired samples t-test was used to determine whether there was a statistically significant mean difference between the right and left legs when participants performed SLS, FL, SML and SLL, while for non-parametric variables a Wilcoxon Rank test was used.

Table 4.2 reveals that during SLS, only the left internal hip rotation angle was significantly higher than the right internal hip rotation angle (MD = 2.54, SD= 6.85 and $p = 0.03$). During FL, the left internal hip rotation angle was significantly higher than the right internal hip rotation (MD = 2.2, $p = 0.04$), while right knee flexion was significantly higher than left knee flexion (MD = 3.79, $p = 0.005$). However, the hip flexion, hip adduction and knee abduction angles were not significantly different during FL. In SML, right hip and knee flexions were significantly higher (MD= 2.73, $p = 0.01$ and MD= 3.08, $p = 0.01$). Finally, during SLL, the right knee flexion angle was significantly higher than the right (MD= 3.74, $p = 0.005$). On the other hand, the moments were significantly different in all variables apart from hip flexion (MD = 0.04, $p = 0.23$) in SLS. Differences between legs in kinetics during FL were shown in internal hip rotation moment and hip adduction moments ($p = 0.004$ and 0.007, respectively). Also, moments were significantly different during SML, except for hip flexion moment ($p = 0.14$). Finally, during SLL, no significant differences showed in hip flexion moment and knee flexion moment.

Table 4. 2 Kinetics and kinematics differences between legs during SLS, FL, SML and SLL

Variables	Single-leg squat				Forward land				Side medial land				Side lateral land			
	Rt.	**Lt.**	MD	**P value**	**Rt.**	**Lt.**	MD	**P value**	**Rt.**	**Lt.**	**MD**	**P value**	**Rt.**	**Lt.**	**MD**	**P value**
Joint Angle (°)																
Hip Int. Rotation	7.43	9.97	2.54	.03*	7.59	9.86	2.27	.05*	9.10	11.49	2.39	.08	8.50	9.45	.94	.43
Hip Adduction	13.59	13.43	.15	.89	7.70	7.53	.16	.88	9.62	9.78	.16	.89	6.20	5.65	.55	.64
Knee Abduction	-1.14	-.56	.57	.31	-1.60	-1.05	.54	.36	-1.48	-1.72	.23	.73	-2.49	-2.07	.42	.72
Moments (Nm/Kg)																
Hip Int. Rotation	-.40	-.50	.10	.005*	-.78	-1.02	.23	.005*	-.71	-1.02	.30	.005*	-.73	-.97	.23	.005*
Hip Adduction	-.98	-1.07	.08	.01*	-1.62	-1.85	.23	.005*	-1.50	-1.93	.43	.005*	-1.53	-1.93	.40	.005*
Knee Abduction	-.05	-.11	.06	.02*	.15	.13	.02	.74	.36	.14	.21	.005*	.31	.12	.19	.005*

- * The mean difference is significant at the .05 level.
- For hip flexion, knee flexion biomechanics and vertical ground reaction force results, see Appendices (6).

To compare the kinematic and kinetic variables between SLS, FL, SML and SLL tasks, a repeated measure ANOVA was conducted for parametric variables and a Friedman test for non-parametric variables. There was a significant effect of tasks across all kinematic and kinetic variables for both the right and the left legs. For the right leg, moments varied highly among all tasks, with the Wilks' lambda ranging between 0.09 and 0.36, $p < 0.0005$ and a multivariate partial eta squared ranging from 0.63 for hip flexion moment and 0.90 for knee flexion moment. Moreover, knee abduction moment was significantly different between FL and the other side landing tasks. However, no significant differences were observed between SML and SLL. The angles were also different for all tasks, with a Wilks' lambda ranging from 0.25 to 0.70, $p < 0.01$ and a multivariate partial eta squared ranging from 0.30 for the internal hip rotation angle and 0.77 for the hip addiction angle. Table 4.3 shows kinetics and kinematics differences among all tasks for the right leg. Similar results were obtained in the left leg. The moments vary widely among all tasks, with the Wilks' lambda ranging between 0.12 and 0.31, $p < 0.0005$ and a multivariate partial eta squared ranging from 0.60 to 0.88 for knee abduction moment and hip adduction, respectively. The angles also varied across all tasks, with a Wilks' lambda ranging from 0.14 – 0.72, $p < 0.01$ and a multivariate partial eta squared ranging from 0.27 for the knee abduction angle and 0.72 for the hip addiction angle. Table 4.4 reveals kinetics and kinematics differences for all left leg tasks.

Table 4. 3 Kinetics and kinematics differences for the right leg between SLS, FL, SML and SLL tasks

Variables	**SLS vs. FL**	**SLS vs. SML**	**SLS vs. SLL**	**FL vs. SML**	**FL vs. SLL**	**SML vs. SLL**
Joint Angles (°)	**MD / P value**	**MD / P value**	**MD / P value**	**MD / P value**	**MD / P value**	**MD / P value**
Hip Int. Rotation	0.16 / 0.63	1.67 / 0.01*	1.07 / 0.44	1.51 / 0.01*	0.91 / 0.25	0.59 / 0.80
Hip Adduction	5.89 / 0.005*	3.97 / 0.005*	7.38 / 0.005*	1.91 / 0.01*	1.49/ 0.02*	3.41 / 0.005*
Knee Abduction	0.46 / 0.14	0.34 / 0.60	1.35 / 0.13	0.11 / 0.48	0.88 / 0.02*	1.00 / 0.01*
Moments (Nm/kg)						
Hip Int. Rotation	0.38 / 0.005*	0.31 / 0.005*	0.33 / 0.005*	0.06 / 0.23	0.04 / 0.77	0.01 / 1.00
Hip Adduction	0.63 / 0.005*	0.51 / 0.005*	0.54 / 0.005*	0.11 / 0.25	0.09 / 1.00	0.02 / 1.00
Knee Abduction	0.21 / 0.005*	0.41 / 0.005*	0.36 / 0.005*	0.20 / 0.005*	0.15 / 0.02*	0.04 / 0.93

- * The mean difference (MD) is significant at the .05 level for parametric and .012 for non-parametric
- Significant adjustment for multiple comparisons: Bonferroni.
- For hip flexion, knee flexion biomechanics and vertical ground reaction force results, see Appendices (6).

Table 4. 4 Kinetics and kinematics differences for the left leg between SLS, FL, SML and SLL tasks

Variables	SLS vs. FL	SLS vs. SML	SLS vs. SLL	FL vs. SML	FL vs. SLL	SML vs. SLL
Joint Angles (°)	**MD / P value**	**MD / P value**	**MD / P value**	**MD / P value**	**MD / P value**	**MD / P value**
Hip Int. Rotation	0.11 / 1.00	1.51 / 0.43	0.52 / 1.00	1.63 / 0.01*	0.41 / 1.00	2.04 / 0.02*
Hip Adduction	5.89 / 0.005*	3.65 / 0.005*	7.78 / 0.005*	2.24 / 0.001*	1.88 / 0.03*	4.12 / 0.005*
Knee Abduction	0.49 / 1.00	1.15 / 0.19	1.50 / 0.01*	0.66 / 0.66	1.01 / 0.14	0.34 / 1.00
Moments (Nm/kg)						
Hip Int. Rotation	0.51 / 0.005*	0.51 / 0.005*	0.46 / 0.005*	0.05 / 0.87	0.01 / 1.00	0.05 / 0.30
Hip Adduction	0.77 / 0.005*	0.86 / 0.005*	0.86 / 0.005*	0.08 / 0.93	0.08 / 0.96	0.002 / 1.00
Knee Abduction	0.24 / 0.005*	0.26 / 0.005*	0.23 / 0.005*	0.01 / 1.00	0.01 / 1.00	0.02 / 0.93

- * The mean difference is significant at the .05 level for parametric and .012 for non-parametric
- Significant adjustment for multiple comparisons: Bonferroni.
- For hip flexion, knee flexion biomechanics and vertical ground reaction force results, see Appendices (6).

Differences between Genders:

Right Leg:
For the right leg, the knee abduction angle was significantly greater in females than in males during SLS ($p < .005$), FL ($p < .005$), SML ($p < .005$) and SLL ($p = 0.01$). In addition, the hip adduction angle was significantly higher in females than in males during SLS ($p = .01$), FL ($p < .005$), SML ($p = 0.001$) and SLL ($p = 0.001$). Moreover, there was a significant difference in knee abduction, moment with females higher than males ($p < .005$). Similarly, in FL, females had significantly greater right knee abduction moment compared to their male participants ($p = .002$)

Left Leg:
For the left leg, females also performed with a significantly greater knee abduction angle and hip adduction angle during SLS, FL, SML and SLS compared to males with $p < 0.03$. However, males (M = -.17, SD = .10) significantly performed SLS with greater knee abduction moment than females during SLS ($p = .02$). However, during FL, females had significantly greater left knee abduction moment ($p = .02$). See Tables 4.5, 4.6, 4.7 and 4.8.

Table 4.5 Gender differences during the SLS task

	Right					Left				
Variables	Females		Males		P value	Females		Males		P value
	Mean	SD	Mean	SD		Mean	SD	Mean	SD	
Joint Angles (°)										
Hip Int. Rotation	6.61	7.36	8.25	6.48	.49	9.26	7.51	10.69	8.85	.61
Hip Adduction	16.86	8.34	10.31	5.04	.01*	15.44	6.53	11.42	4.05	.03*
Knee Abduction	-3.39	4.54	1.11	4.14	.005*	-2.23	4.27	1.10	2.34	.01*
Moments (Nm/kg)										
Hip Int. Rotation	-.43	.16	-.38	.13	.34	-.51	.14	-.49	.15	.71
Hip Adduction	-1.03	.25	-.93	,19	.21	-1.07	.23	-1.06	,13	.90
Knee Abduction	.03	.18	-.14	.16	.005*	-.05	.18	-.17	.10	.02*

- The mean difference is significant at the .05 level.
- For hip flexion, knee flexion biomechanics and vertical ground reaction force results, see Appendices (6).

Table 4.6 Gender differences during the FL task

Variables	Right					Left				
	Females		Males		P value	Females		Males		P value
	Mean	SD	Mean	SD		Mean	SD	Mean	SD	
Joint Angles (°)										
Hip Int. Rotation	7.22	7.75	7.96	6.76	.77	9.16	7.88	10.56	8.25	.61
Hip Adduction	11.00	6.71	4.40	5.11	.005*	9.37	3.99	5.70	3.35	.005*
Knee Abduction	-4.03	4.92	.82	4.12	.005*	-2.93	3.77	.81	3.97	.005*
Moments (Nm/kg)										
Hip Flexion	-1.83	.54	-1.72	.66	.59	-1.59	.56	-1.93	.80	.15
Hip Int. Rotation	-.79	.27	-.77	.19	.84	-1.00	.26	-1.04	.32	.64
Hip Adduction	-1.68	.25	-1.55	,32	.20	-1.93	.25	-1.77	,40	.18
Knee Abduction	.28	.25	.03	.17	.005*	.20	.22	.05	.13	.02*

- The mean difference is significant at the .05 level.
- For hip flexion, knee flexion biomechanics and vertical ground reaction force results, see Appendices (6).

Table 4.7 Gender differences during the SML task

Variables	Right					Left				
	Females		Males		P value	Females		Males		P value
	Mean	SD	Mean	SD		Mean	SD	Mean	SD	
Joint Angles (°)										
Hip Int. Rotation	8.35	8.01	9.85	7.08	.56	10.01	7.36	12.89	9.76	.35
Hip Adduction	12.97	5.49	6.26	4.75	.005*	12.21	4.04	4.04	7.34	.005*
Knee Abduction	-3.68	4.64	.70	4.53	.005*	-3.34	3.96	.93	5.40	.04*
Moments (Nm/kg)										
Hip Int. Rotation	-.73	.22	-.70	.18	.64	-1.01	.31	-1.04	.28	.77
Hip Adduction	-1.55	.32	-1.44	.25	.28	-1.99	.38	-1.87	.30	.32
Knee Abduction	.46	.35	.25	.28	.06	.16	.29	.12	.24	.68

- The mean difference is significant at the .05 level.
- For hip flexion, knee flexion biomechanics and vertical ground reaction force results, see Appendices (6).

Table 4.8 Gender differences during the SLL task

Variables	Right					Left				
	Females		Males		P value	Females		Males		P value
	Mean	SD	Mean	SD		Mean	SD	Mean	SD	
Joint Angles (°)										
Hip Int. Rotation	7.66	7.34	9.34	7.56	.51	9.12	7.73	9.78	10.70	.84
Hip Adduction	9.65	5.97	2.76	4.28	.005*	7.75	4.08	3.55	4.94	.01*
Knee Abduction	-4.55	4.21	-.43	5.16	.01*	-3.99	4.84	-.41	4.57	.02*
Moments (Nm/kg)										
Hip Int. Rotation	-.67	.19	-.79	.21	.09	-.97	.28	-.96	.20	.91
Hip Adduction	-1.52	.33	-1.53	.38	.99	-2.01	.39	-1.85	.29	.18
Knee Abduction	.37	.31	.25	.23	.22	.16	.20	.07	.12	.11

- The mean difference is significant at the .05 level.
- For hip flexion, knee flexion biomechanics and vertical ground reaction force results, see Appendices (6).

Female group:

A paired samples t-test was used for parametric variables and a Wilcoxon Rank test for non-parametric variables, to determine whether there was a statistically significant mean difference between the right and left legs when female participants performed SLS, FL, SML and SLL. During SLS, the hip flexion angle was significantly different between legs in the female group (MD = 2.24, SD = 4.17, $p < 0.04$ (two-tailed)). Kinetic variables were significantly different in internal hip rotation moment (MD = 0.08, SD = 0.08, $p < 0.001$(two-tailed)) and knee abduction moment ($z = -2.05$, $p = 0.03$). During landing tasks, kinematic variables were not significantly different between legs in the female group, except for the knee flexion angle during FL (M Differences = 3.79, SD = 6.82, $p < 0.03$ (two-tailed)). Significant differences were found between legs in kinetics during landing in internal hip rotation and hip adduction moments during FL ($p < 0.001$ and 0.007, respectively). In addition, kinetic variables were significantly different during SML, except for hip flexion moment. During SLL, internal hip rotation moment (M D = 0.29, SD = 0.20, $p < 0.0005$ (two-tailed)) and hip adduction moment (M Difference = 0.48, SD = 0.48, $t (16) = 4.16$, $p < 0.001$ (two-tailed)) were significantly different between legs among the females. Please see Table 4.9.

Table 4.9 Kinetics and kinematics differences between legs during SLS, FL, SML and SLL in the female group

Variables	Single-leg squat				Forward land				Side medial land				Side lateral land			
	Rt.	**Lt.**	MD	**P value**	**Rt.**	**Lt.**	MD	**P value**	**Rt.**	**Lt.**	**MD**	**P value**	**Rt.**	**Lt.**	**MD**	**P value**
Joint Angle (°)																
Hip Int. Rotation	6.61	9.26	2.64	.16	7.22	9.16	1.94	.30	8.35	10.10	1.74	.39	7.66	9.12	1.46	.46
Hip Adduction	16.86	15.44	1.42	.48	11.00	9.37	1.62	.34	12.97	12.21	.75	.68	9.65	7.75	1.89	.31
Knee Abduction	-3.39	-2.23	1.15	.10	-4.03	-2.93	1.10	.26	-3.68	-3.34	.33	.69	-4.55	-3.99	.56	.57
Moments (Nm/Kg)																
Hip Flex	-1.13	1.06	.06	.19	-1.83	-1.59	.24	.12	-1.72	-2.06	.33	.84	-1.76	-1.79	.02	.85
Hip Int. Rotation	-.43	-.51	.08	.005*	-.79	-1.00	.20	.005*	-.73	-1.01	.27	.005*	-.67	-.97	.29	.005*
Hip Adduction	-1.03	-1.07	.04	.42	-1.68	-1.93	.24	.005*	-1.55	-1.99	.43	.005*	-1.52	-2.01	.48	.005*
Knee Abduction	-.03	-.05	.09	.03*	.28	.20	.07	.36	.46	.16	.30	.005*	.37	.16	.20	.05*

- * The mean difference is significant at the .05 level.
- For hip flexion, knee flexion biomechanics and vertical ground reaction force results, see appendices (6).

A repeated measures ANOVA showed a significant effect of tasks across all kinematic and kinetic variables for both the right and left legs, except for the right and left internal hip rotation angle and the knee abduction angle. For the right leg, the moments were highly different between all tasks, with a Wilks' lambda ranging from 0.10 – 0.44, $p < 0.009$ and a multivariate partial eta squared ranged from 0.55 for hip flexion moment and 0.89 for knee flexion moment. Differences in kinetics and kinematics between all tasks for the right leg are presented in Table 4.10. The angles were also different between all tasks, with a Wilks' lambda ranging between 0.13 and 0.18, $p < 0.005$ and a multivariate partial eta squared ranged from 0.84 for the hip flexion angle and 0.86 for the knee flexion angle. Almost the same results were found in the left leg. Moments differed highly between all tasks, with a Wilks' lambda ranging between 0.08 and 0.44, $p < 0.008$ and a multivariate partial eta squared ranging from 0.55 – 0.91 for knee abduction moment and knee flexion moment, respectively. The angles were also different between all tasks, apart from the internal hip rotation angle and the knee abduction angle, with a Wilks' lambda ranging between 0.12 and 0.25, $p < 0.005$ and a multivariate partial eta squared ranging from 0.74 for the hip adduction angle and 0.87 for the knee flexion angle. Table 4.11 shows kinetics and kinematics differences between all tasks for the left leg in the female group.

Table 4.10 Kinetics and kinematics differences of right leg between SLS, FL, SML and SLL tasks in the female group

Variables	SLS vs. FL	SLS vs. SML	SLS vs. SLL	FL vs. SML	FL vs. SLL	SML vs. SLL
Joint Angles (°)	**MD / P value**	**MD / P value**	**MD / P value**	**MD / P value**	**MD / P value**	**MD / P value**
Hip Int. Rotation	0.60 / 1.00	1.73 / 0.05	1.04 / 0.41	1.12 / 0.38	0.43 / 1.00	0.68 / 0.90
Hip Adduction	5.86 / 0.005*	3.89 / 0.11	7.21 / 0.005*	1.97 / 0.20	1.34 / 0.43	3.32 / 0.001*
Knee Abduction	0.63 / 1.00	0.28 / 1.00	1.15 / 0.61	0.35 / 1.00	0.52 / 0.81	0.87 / 0.27
Moments (Nm/kg)						
Hip Int. Rotation	0.36 / 0.005*	0.30 / 0.005*	0.24 / 0.005*	0.05 / 0.88	0.11 / 0.04*	0.05 / 0.34
Hip Adduction	0.65 / 0.005*	0.52 / 0.005*	0.49 / 0.005*	0.12 / 0.19	0.15 / 0.19	0.02 / 1.00
Knee Abduction	0.25 / 0.005*	0.43 / 0.005*	0.33 / 0.005*	0.18 / 0.12	0.08 / 1.00	0.09 / 1.00

- * The mean difference is significant at the .05 level for parametric and .012 for non-parametric
- Significant adjustment for multiple comparisons: Bonferroni.
- For hip flexion, knee flexion biomechanics and vertical ground reaction force results, see Appendices (6).

Table 4. 11 Kinetics and kinematics differences of left leg between SLS, FL, SML and SLL tasks in female

Variables	SLS vs. FL	SLS vs. SML	SLS vs. SLL	FL vs. SML	FL vs. SLL	SML vs. SLL
Joint Angles (°)	**MD / P value**	**MD / P value**	**MD / P value**	**MD / P value**	**MD / P value**	**MD / P value**
Hip Int. Rotation	0.09 / 1.00	0.83 / 0.76	0.13 / 1.00	0.93 / .26	0.04 / 1.00	0.97 / 0.69
Hip Adduction	6.06 / 0.005*	3.22 / 0.07	7.68 / 0.005*	2.84 / 0.006*	1.61 / 0.52	4.46 / 0.001*
Knee Abduction	0.69 / 1.00	1.11 / 0.33	1.75 / 0.12	0.41 / 1.00	1.06 / 1.00	0.64 / 1.00
Moments (Nm/kg)						
Hip Int. Rotation	0.48 / 0.005*	0.49 / 0.005*	0.45 / 0.005*	0.01 / 1.00	0.02 / 1.00	0.03 / 1.00
Hip Adduction	0.85 / 0.005*	0.91 / 0.005*	0.93 / .005*	0.06 / 1.00	0.08 / 1.00	0.01 / 1.00
Knee Abduction	0.26 / 0.007*	0.22 / .01*	0.22 / .008*	0.04 / 1.00	0.03 / 1.00	0.02 / 0.93

- * The mean difference is significant at the .05 level for parametric and .012 for non-parametric
- Significant adjustment for multiple comparisons: Bonferroni.
- For hip flexion, knee flexion biomechanics and vertical ground reaction force results, see Appendices (6).

Male group:

During SLS, there were no significant differences between the right and the left knee in kinematic variables. However, kinetic variables were significantly different in all variables ($M = 0.11$, $SD = 0.08$, $p < 0.0005$ (two-tailed)) for internal hip rotation moment, ($M = 0.13$, $SD = 0.19$, $p < 0.01$(two-tailed)) for hip adduction moment, and ($M = 0.16$, $SD = 0.24$, $p < 0.014$ (two-tailed)) for hip flexion moment. Knee abduction moment was not significantly different between legs.

During landing tasks, kinematic variables were not significantly different between legs in the male group except for the hip flexion angle during SML ($M = 3.49$, $SD = 5,61$, $p < 0.02$ (two-tailed)), the knee flexion angle during SML ($M = 5.19$, $SD = 6.46$, $p < 0.004$ (two-tailed)) and the knee flexion angle during SLL ($M = 6.00$, $SD = 8.84$, $t(16) = 2.79$, $p < 0.01$ (two-tailed)). On the other hand, differences between legs in kinetics during landing were found in internal hip rotation during FL ($M = 0.27$, $SD = 0.30$, $t(16) = 3.63$, $p < 0.002$ (two-tailed)). In addition, kinetic variables were significantly different during SML except for hip flexion moment and knee abduction moment. Finally, during SLL, all moments were significantly different between legs among the males. Details are presented in Table 4.12.

Table 4. 12 Kinetics and kinematics differences between legs during SLS, FL, SML and SLL in the male group

Variables	Single-leg squat				Forward land				Side medial land				Side lateral land			
	Rt.	**Lt.**	MD	**P value**	**Rt.**	**Lt.**	MD	**P value**	**Rt.**	**Lt.**	**MD**	**P value**	**Rt.**	**Lt.**	**MD**	**P value**
Joint Angle (°)																
Hip Int. Rotation	8.25	10.69	2.64	.12	7.96	10.56	2.04	.06	9.85	12.89	2.39	.10	9.34	9.78	.94	.77
Hip Adduction	10.31	11.42	1.09	.43	4.40	5.70	1.62	.35	6.26	7.34	.16	.45	2.76	3.55	.55	.61
Knee Abduction	1.11	1.10	1.15	.99	.82	.81	1.10	.99	.70	-.93	.23	.46	-.43	-.14	.42	.74
Moments (Nm/Kg)																
Hip Int. Rotation	-.38	-.49	.11	.005*	-.77	-1.04	.27	.005*	-.70	-1.04	.34	.005*	-.79	-.96	.17	.005*
Hip Adduction	-.93	-1.06	.13	.01*	-1.55	-1.77	.23	.06	-1.44	-1.87	.43	.005*	-1.53	-1.85	.32	.02*
Knee Abduction	-.14	-.17	.03	.22	.15	.13	.02	.63	.25	.12	.21	.16	.25	.17	.08	.005*

- * The mean difference is significant at the .05 level.
- For hip flexion, knee flexion biomechanics and vertical ground reaction force results, see Appendices (6).

Significant effects of tasks across all kinetic variables for both the right and the left legs. For the right leg, the moments varied widely between all tasks, with a Wilks' lambda ranging between 0.08 and 0.23, $p < 0.005$ and a multivariate partial eta squared ranging from 0.76 for hip flexion moment and 0.91 for knee flexion moment. Moreover, knee abduction moment was significantly different between FL and the other side landing tasks. For the left leg, the Wilks' lambda ranged between 0.09 and 0.29, $p < 0.005$ and a multivariate partial eta squared ranged from 0.70 for hip flexion moment and 0.90 for knee flexion moment. However, kinematic variables were significantly different in the hip adduction angles in both the right (Wilks' lambda (0.23), $p < 0.005$ and a multivariate partial eta squared 0.76) and the left (Wilks' lambda (0.28), $p < 0.005$ and a multivariate partial eta squared 0.71) legs. Table 4.13 shows kinetics and kinematics differences between all tasks for the right leg. The table reveals that no significant differences were observed between SML and SLL ($p = 1.00$). Similar results are found in the left leg, as illustrated in Table 4.14.

Table 4. 13 Kinetics and kinematics differences for the right leg between SLS, FL, SML and SLL tasks in the male group

Variables	SLS vs. FL	SLS vs. SML	SLS vs. SLL	FL vs. SML	FL vs. SLL	SML vs. SLL
Joint Angles (°)	**MD / P value**	**MD / P value**	**MD / P value**	**MD / P value**	**MD / P value**	**MD / P value**
Hip Int. Rotation	0.28 / 1.00	1.60 / 0.53	1.09 / 1.00	1.89 / 0.09	1.38 / 0.28	0.51 / 1.00
Hip Adduction	5.91 / 0.005*	4.05 / 0.005*	7.55 / 0.005*	1.86 / 0.27	1.64 / 0.20	3.50 / 0.005*
Knee abduction	0.29 / 1.00	0.40 / 1.00	1.54 / 0.67	0.11 / 1.00	1.25 / 0.08	1.14 / 0.11
Moments (Nm/kg)						
Hip Int. Rotation	0.39 / 0.005*	0.32 / 0.005*	0.41 / 0.005*	0.07 / 1.00	0.01 / 1.00	0.09 / 1.00
Hip Adduction	0.62 / 0.005*	0.51 / 0.005*	0.59 / 0.005*	0.10 / 0.25	0.02 / 1.00	0.03 / 1.00
Knee abduction	0.17 / 0.005*	0.39 / 0.005*	0.39 / 0.005*	0.22 / 0.01*	0.22 / 0.008*	0.004 / 1.00

- * The mean difference is significant at the .05 level for parametric and .012 for non-parametric
- Significant adjustment for multiple comparisons: Bonferroni.
- For hip flexion, knee flexion biomechanics and vertical ground reaction force results, see Appendices (6).

Table 4. 14 Kinetics and kinematics differences for the left leg between SLS, FL, SML and SLL tasks in the male group

Variables	SLS vs. FL	SLS vs. SML	SLS vs. SLL	FL vs. SML	FL vs. SLL	SML vs. SLL
Joint Angles (°)	**P value**	**P value**	**P value**	**P value**	**P value**	**P value**
Hip Int. Rotation	0.13 / 1.00	2.19 / 1.00	0.91 / 1.00	2.33 / 0.08	0.78 / 1.00	3.11 / 0.10
Hip Adduction	5.72 / 0.005*	4.08 / 0.005*	7.87 / 0.005*	1.64 / 0.18	2.15 / 0.19	3.79 / 0.004*
Knee abduction	0.29 / 1.00	1.19 / 1.00	1.25 / 0.24	0.90 / 1.00	0.96 / 0.18	0.05 / 0.11
Moments (Nm/kg)						
Hip Int. Rotation	0.55 / 0.005*	0.54 / 0.005*	0.46 / 0.005*	0.009 / 1.00	0.08 / 1.00	0.07 / 1.00
Hip Adduction	0.70 / 0.005*	0.80 / 0.005*	0.78 / 0.005*	0.10 / 1.00	0.07 / 1.00	0.02 / 1.00
Knee abduction	0.23 / 0.005*	0.30 / 0.005*	0.24 / 0.005*	0.07 / 1.00	0.01 / 1.00	0.05 / 1.00

- * The mean difference is significant at the .05 level for parametric and .012 for non-parametric
- Significant adjustment for multiple comparisons: Bonferroni.
- For hip flexion, knee flexion biomechanics and vertical ground reaction force results, see Appendices (6).

4.4 Discussion:

The project's main goal was to explore how gluteus maximus and gluteus medius relate to biomechanical variables in active, healthy subjects during single-leg squats and multi-directional single-leg landing. To achieve this aim, it was important to analyse and understand how these groups preform the tasks. Therefore, the goals of this chapter were to investigate differences in kinetics and kinematics between legs, tasks and genders during single-leg multi-directional landing and single-leg squat tasks.

During SLS in the study, the average knee abduction angle value was -1.14° for the right leg and -.56° for the left leg, with internal hip rotation angles of 7.43° and 9.97° for the right and left, respectively. The average hip adduction value was 13.59 for the right leg and 13.43 for the left leg. Similar results were reported in the literature during SLS tasks, except that the knee abduction angles were higher in previous studies (Alenezi et al., 2014, Baldon et al., 2011, DiMattia, Livengood, Uhl, Mattacola, and Malone, 2005, Graci, Van Dillen, and Salsich, 2012, Nakagawa, 2012, Nguyen et al., 2011, Scattone Silva and Serrão, 2014, Weeks, Carty, and Horan, 2012, Willy and Davis, 2011, Zeller et al., 2003). The possible answer for this is the different methodological tools used in the previous studies, such as marker list models. It has been stated that the results of studies that use different marker list models cannot be directly compared (Collins et al., 2009). In the current study, the CAST model was used, as it improved anatomical relevance when compared to Helen Heyes (Kadaba et al., 1989). The advantage of using CAST is being able to use an attachment in the centre of the segments instead of single marker close to the joint, in order to help reduce skin movement artefacts (Alenezi et al., 2014). Furthermore, limb asymmetry was found in the internal hip rotation angle in this study. A significant higher internal rotation angle value was noted in the left leg (9.97 vs 7.43), but when gender data were analysed separately, there was no significant difference between the right and the left limbs when performing SLS in either gender; in fact, differences were found only in the hip flexion angle ($p = .04$) in the female group.

Kinetic variables were significantly different between legs, except in hip flexion moment and ground reaction force. Furthermore, differences in moments were found as well when data obtained from males and females were analysed separately. This asymmetry between legs, especially in knee abduction moment which has been described previously as an ACL injury risk factor (Hewett et al., 2005). This significant difference might be explained by the theory of non-dominant leg strength asymmetry (Lanshammar and Ribom, 2011), as three of the 34 participants were left-leg dominant. The literature also reports differences between left and right or dominant and non-dominant limbs in neuromuscular control (Ford et al., 2003, Herrington, 2011). However, the dominance of a limb does not predict ACL injury (Hewett et al., 2005). Neuromuscular asymmetry between legs will be discussed later.

Differences between females and males in the SLS task were noted in this study. Knee abduction angles were significantly higher in females (M = -3.39, SD = 8.34) than in males (M = 1.11, SD = 4.14), and knee adduction angles were also significantly higher in females (M = 16.86, SD = 4.54) than in males (M = 10.31, SD = 5.04). These differences were higher than the SEM value presented in Table 3.4 in the reliability study in Chapter 3. This finding is supported by the literature for the same task (Dwyer, Boudreau, Mattacola, Uhl, and Lattermann, 2010, Yamazaki, Muneta, Ju, and Sekiya, 2009b, Zeller et al., 2003). The current study did not found significant differences in the internal hip rotation angle, though Zeller et al. (2003) did report significant differences in internal hip rotation motion, which might be explained by the differences in sample sizes, as our study used 34 participants whereas Zeller et al. (2003) used 18.

Knee abduction and hip adduction angles in association with the internal hip rotation angle form part of the definition for the dynamic knee valgus manoeuvre, which has been reported as an ACL injury risk factor (Hewett et al., 2005). This might be explained by the strength differences between genders, as females demonstrate lower peak isometric and isokinetic strength compared to males in lower limb muscles (Claiborne et al., 2006, Dwyer et al., 2010). In the current study, the strength of hip abductors and extensors was measured isokinetically for all subjects and will be discussed later in Chapter 6, while the relationship to movement will be discussed in Chapter 7.

With regards to FL, the average knee abduction angle value was -1.60° for the right leg and -1.05° for the left leg, with internal hip rotation angles of 7.59° and 9.86°, respectively, for the right and left legs. The average hip adduction angle was 7.70° for the right leg and 7.53° for the left leg. Similar results were found by Pappas et al. (2007) when comparing unilateral landing to bilateral landing across 32 athletes. During unilateral landing from a 40-cm box, the study reported a -0.96° knee abduction angle (Pappas et al., 2007). Nonetheless, some studies reported higher results (Kiriyama, Sato, and Takahira, 2009, Orishimo, Kremenic, Pappas, Hagins, and Liederbach, 2009). Kiriyama et al. (2009), for instance, reported knee abduction angles across 169 healthy young subjects when a single-leg drop landing was performed from a 20-cm box. The knee abduction angle was -3° for females and -2° for males. This slight difference could be explained by differences in the ages of the target population, as Kirimaya et al. (2009) involved younger people (mean age 17 ± 1 years). However, a study reported higher knee abduction angles when subjects performed forward drop landings from a 30-cm box (-11.5° and -8.4° for females and males, respectively) (Orishimo et al., 2009). In addition, hip adduction angles were higher (15.4° and 15.3° for females and males, respectively). This difference could be because Orishimo et al. (2009) targeted 33 professional ballet dancers, while participants in the current study carried were healthy and recreationally active only.

During the SML task, this study found an average knee abduction angle of -1.48° for the right leg and -1.72° for the left leg. This study also found average internal hip rotation angles of 9.10° and 11.49° and average hip adduction angles of 9.62° and 9.78° for the right and the left leg, respectively. Greater knee abduction angles have been reported during the SML task from a 20-cm box (Suzuki et al., 2015b), albeit the study used a different marker lists model, so it cannot be compared directly to the current study. In addition, college basketball players were included. Although participants performed SML with higher hip adduction and internal rotation angles, this may indicate more effort from the gluteal muscles to prevent dynamic knee valgus and ACL injury, as they play an important role in reducing the knee abduction angle during weight-bearing activities in the frontal plane (Claiborne et al., 2006). A strong correlation was found between hip muscles and medial knee position on landing (Suzuki et al., 2015b). However, the study used an isometric strength test instead of concentric and eccentric strength used in the current study.

Differences between legs have been found significantly in the knee flexion angle, internal hip rotation moment and hip adduction moment. Participants in this current study performed FL with lower knee flexion in the left leg than in the right (67.97° vs 64.17°, $p = .001$). Internal hip rotation angles between legs were also different, but not significantly (7.59° and 9.86° for the right and the left, respectively, $p = 0.05$). As stated earlier, this difference might be explained by differences in neuromuscular control between legs, which will be discussed later.

During SLL, the average knee abduction angle value was -2.49° for the right leg and -2.07° for the left leg. Internal hip rotation angles were 9.10° and 11.49° for the right and left, respectively. The average hip adduction angle was lower at 9.62° for the right leg and 9.78° for the left leg. To date, no study has used a step or a box to perform lateral single-leg landings to examine active, healthy lower limb biomechanics. One study did examine the relationship between hip function during medial and lateral side hop landing (Itoh et al., 2016), which was performed by standing on two different force plates (30 cm apart) with participants performing ten repetitions on each leg. When compared to the lateral and medial landing, lateral landing was reported with a higher knee abduction angle than medial landing. This supports our results, but a larger knee abduction angle was reported than the current study. This can be explained by the different marker lists model, but the angle during the touchdown phase, a smaller sample size and fatigue due to 10 repetitions by the participants might also have affected the results.

Differences between females and males during all landing tasks were noted in the knee abduction angle and the hip adduction angle for both the right and the left legs in this current study. The same finding has also been found during single-leg hop landing with a greater knee abduction angle in women (Jacobs et al., 2007), a 40-cm drop bilateral and unilateral landing (Pappas et al., 2007) and a 50-cm drop landing (Lawrence et al., 2008). However, no significant difference has been found during side medial drop landing (Suzuki et al., 2015b), though a significantly higher internal hip rotation angle in females than in males has been cited. This may explain, therefore, why females have higher injury rates than males, as these increases in knee abduction, hip adduction and internal hip rotation in females may lead to higher dynamic knee valgus and more load on the ACL. These differences might be due to variances in the neuromuscular control of the gluteus maximus and medius muscles between both groups. Another factor that might cause these anomalies is anatomical differences, as females have a higher quadriceps angle (Q angle) than their male counterparts (Beutler et al., 2009). Females landed with greater ground reaction force in both legs than males during SML and SLL, which might also explain why females are more prone to injury than males, since higher ground reaction force has been linked to ACL damage (Hewett et al., 2005).

With regards to making a comparison between tasks, a significant effect of tasks across all kinematic and kinetic variables was noted for both the right and the left legs. Knee abduction angle was significantly affected by tasks $p = .003$ with a large effect size (partial eta squared = 0.31). Similar results were found for internal hip rotation and the hip adduction angles with large effect sizes (.30 and .75, respectively). Differences were mainly found between SLS and all landing tasks. A significant higher knee abduction angle was identified during SLL (-2.49°) than in FL (- 1.60°) and SML (- 1.49) when examining the right leg. However, a significantly higher hip adduction angle during SML (9.62°) than in FL (7.70°) and SLL (6.20°) was established. This is not surprising, as SML and SLL are more difficult to perform, thus indicating that direction influenced the results and was more predictive of ACL injury. The knee abduction angle on the left leg also showed significant differences between SLS and SLL. However, the hip adduction angle was also higher during SML (9.78°) than in FL (7.53°) and SLL (6.20°). Similarly, kinetic variables were significantly different between SLS and landing tasks in both the right and the left leg, with a partial eta squared ranging from 0.63 for hip flexion moment and 0.90 for knee flexion moment. Knee abduction moment was significantly different between FL and the side landing tasks. However, no significant differences were observed between SML and SLL. Similar results were found in the left leg, with a partial eta squared ranging from 0.60–0.88 for knee abduction moment and hip adduction moment, respectively. In a study carried out on 19 female volleyball players, Sinsurin et al. (2017) found that jump landing direction has a significant effect on the ground reaction force. Subjects performed forward, diagonal and lateral hop landings and reported lower knee abduction angles ranging from - 1.2 to -2.5° for the forward and lateral landing. However, tasks performed by volleyball players and the current study used active, healthy subjects in different types of activities.

Not surprisingly, performance was different across all tasks. This could potentially suggest that not just one functional task is required for screening, as each task covered in the current study produced different results. In other words, the use of a single task does not sufficiently identify the risk factors involved in injuries. It is plausible to assume these task differences may place subjects at risk of injury to varying degrees, and so understanding the reasons for these differences in performance is important, as it might help in developing interventions used in training and rehabilitation.

The limb symmetry index is commonly used in order to assess the value of one limb in relation to another, in order to return to play from injury such as ACL, and results more than 85% indicates limb symmetry (Munro and Herrington, 2011). However, the differences reported when landing by active, healthy subjects suggest that it would be better to investigate each limb in isolation during landing tasks, without depending on another. Differences between genders exist, as females squat and land with greater knee abduction and hip adduction angles. Not surprisingly, these differences have been reported in the literature as risk factors, which might answer the question as to why females exhibit more ACL injuries than males. Knee abduction in the current study moment was significantly different during single-leg squats and forward single-leg landing.

There were limitations in the current study. First, it was conducted on active, healthy participants with different activity levels, so it would be difficult to generalise the results to a population with lower limbs pathology. Second, some of the participants might have more experience in squatting or landing. Third, it was difficult to standardise the squat depth. Furthermore, the lab was a high-safety and well-controlled environment compared to sports and training fields, so further studies are needed, in order to transfer our findings to real-life sporting environments.

Conclusion:

In conclusion, the study established typical kinematic and kinetic variables for an active, healthy population, which will provide greater understanding when exploring the relationship between these variables and gluteal function during single-leg squats and single-leg multi-directional landing tasks. Across all tasks, significant differences were found between the right and the left leg, especially for the moment variables hip adduction, internal rotation and abduction). This implies that each limb must be screened and rehabilitated separately when trying to measure kinetics and kinematics. Furthermore, hip adduction and knee abduction angles are significantly higher in females; however, moments were similar apart from knee abduction moment during FL. Finally, there was a significant task effect on lower limb biomechanical variables, especially between SLS and other landing tasks. The findings reported herein will be used in the relationship study, in order to explore whether EMG activity or strength relates to lower limb biomechanics. Thus, a better understanding of how these differences affect the relationship has been established. The following two chapters will investigate whether there are differences between limbs and between genders in strength and EMG activity data when performing single-leg squats and single-leg multi-directional landing.

Chapter 5

Electromyography activity of Gluteus Maximus and Gluteus Medius during Single-leg Squats and Multi-directional Single-leg Landing:

5.1 Introduction:

The previous chapter explored the differences in kinematic and kinetic variables between legs, genders and tasks. This chapter aims to report if there are any differences in the EMG activity of the G Max and G Med muscles, in order to provide more information about how this relates to other biomechanical variables. For example, when females perform single landing with higher hip addiction and knee abduction angles, it would be expected for this to be associated with lower gluteus medius EMG activity. Therefore, this chapter aims to investigate the differences in G Max and G Med EMG activity between legs, tasks and genders during single-leg multi-directional landing and single-leg squat tasks.

Study Hypothesis:

- Differences in G Max and G Med EMG activity exist between limbs during single-leg multi-directional landing and single-leg squat tasks.
- Differences between genders exist especially in G Med EMG activity during single-leg multi-directional landing and single-leg squat tasks.
- Differences exist especially in G Med EMG activity across all tasks.

5.2 Methods:

A total of 34 active, healthy participants, comprising an equal number of males and females, participated in this study. Demographic information is as listed in Table 4.1. The same inclusion and exclusion criteria were used as stated earlier in the reliability studies in Chapter 3. A consent form was read and signed by all participants before taking part in the study. G Max and G Med EMG activity data were recorded simultaneously using the 3D capture Noraxon Desktop DTS system (www.noraxon.com) at 1500 Hz, by placing a disposable self-adhesive Noraxon surface electrode over the muscle (parallel to the muscle fibre). The same instruments, filtration, calibration, marker list, training shoes, functional tasks and biomechanical model described earlier in the reliability studies in Chapter 3 were used in this study.

Statistical Analysis:

The same statistical approaches were as utilised and described earlier in Chapter 4.

5.3 Results:

A normality test (Shapiro-Wilk) for EMG activity revealed that all variables during all tasks were normally distributed, apart from right G Med and left G Max EMG activity during SLS. See Appendix (3) for the normality test for all variables. A paired samples t-test was used for parametric variables and a Wilcoxon-Rank test for non-parametric variables, to determine whether there was a statistically significant mean difference between right and left legs when participants performed SLS, FL, SML and SLL. Table 5.1 shows that there was a significant decrease in left G Max EMG activity than in the right leg ($z = -4.42$, $p = .004$). However, there was no significant difference in EMG activity between legs during all tasks.

To compare the EMG activity of the gluteus maximus and the gluteus medius between the SLS, FL, SML and SLL tasks, a one-way repeated measures ANOVA was conducted for parametric variables and a Friedman test for non-parametric variables. There was a significant effect of tasks across all EMG activity for both G Max and G Med. For right and left G Max, EMG activity showed a significantly different Wilks' lambda (0.25 and 0.38, respectively), $p < 0.005$ and a multivariate partial eta squared of 0.74 for right the G Max and 0.61 for the left G Max. For the right and left G Med, EMG activity was also significantly different, with a Wilks' lambda of 0.50 and 0.46, respectively, $p < 0.005$ and a multivariate partial eta squared of 0.61 for the right G Med and 0.53 for the left G Med, as illustrated in Table 5.2.

Table 5.1 EMG activity data differences between legs during SLS, FL, SML and SLL

Variables	Single-leg squat			Forward landing			Side medial landing			Side lateral landing		
	Rt.	**Lt.**	**P value**	**Rt.**	**Lt.**	**P value**	**Rt.**	**Lt.**	**P value**	**Rt.**	**Lt.**	**P value**
EMG RMS (%)												
Gluteus Max.	.54	.48	.004*	.28	.23	.09	.27	.24	.33	.29	.25	.18
Gluteus Med.	.51	.53	.36	.33	.33	.90	.33	.34	.54	.32	.34	.52

- * The mean difference is significant at the .05 level.

Table 5.2 EMG activity data differences in the right leg between SLS, FL, SML and SLL tasks

Variables	SLS vs. FL	SLS vs. SML	SLS vs. SLL	FL vs. SML	FL vs. SLL	SML vs. SLL
EMG RMS (%)	**MD / P value**	**MD / P value**	**MD / P value**	**MD / P value**	**MD / P value**	**MD / P value**
Rt. gluteus Max.	0.26 / 0.005*	0.25 / 0.005*	0.27 / 0.005*	0.01 / 0.68	0.01 / 0.82	0.02 / 1.00
Rt. gluteus Med.	0.18 / 0.005*	0.18 / 0.005*	0.17 / 0.005*	0.003 / 0.97	0.01 / 0.47	0.01 / 1.00
Lt. gluteus Max.	0.25 / 0.005*	0.26 / 0.005*	0.27 / 0.005*	0.01 / 1.00	0.02 / 0.84	0.01 / 1.00
Lt. gluteus Med.	0.20 / 0.005*	0.21 / 0.005*	0.21 / 0.005*	0.01 / .56	0.01 / 1.00	0.004 / 1.00

- * The mean difference (MD) is significant at the .05 level.
- Significant adjustment for multiple comparisons: Bonferroni.

Gender Differences:

An independent t-test was used for parametric variables and a Mann-Whitney U test for non-parametric variables, to determine whether there was a statistically significant difference between female and male participants performing SLS, FL, SML and SLL. There was no significant difference in EMG activity during the SLS task for G Max and G Med in either leg. However, during landing, there was a significant difference in G Med EMG activity when FL was performed. The magnitude of the differences in the means was M = .14, p = .001 for the right leg and M= .11, p = .007 for the left leg. During SML, there was a significant difference in right G Med EMG activity for males (M = .29. SD = .10) and females (M = .38, SD = .12). Furthermore, there was a significant difference in right G Med EMG activity during SLL for males (M = .27. SD = .08) and females (M = .39, SD = .13), and left G Max EMG activity for males (M = .21. SD = .09) and females (M = .36, SD = .14, t (32) = 2.46, p = .01, two-tailed), as shown in Table 5.3.

Table 5.3 Gender differences in EMG activity data for Gluteus Maximus and Medius during SLS and landing tasks

Variables	Females		Males		P value
	Mean	SD	Mean	SD	
Single-leg Squat					
Rt. gluteus Max.	.30	.18	.26	.12	.48
Rt. gluteus Med.	.40	.12	.26	.09	.005*
Lt. gluteus Max.	.25	.11	.21	.11	.30
Lt. gluteus Med.	.38	.11	.27	.10	.005*
Forward Landing					
Rt. gluteus Max.	.30	.18	.26	.12	.48
Rt. gluteus Med.	.40	.12	.26	.09	.005*
Lt. gluteus Max.	.25	.11	.21	.11	.30
Lt. gluteus Med.	.38	.11	.27	.10	.005*
Side Medial Landing					
Rt. gluteus Max.	.29	.18	.25	.12	.44
Rt. gluteus Med.	.38	.12	,29	,10	.005*
Lt. gluteus Max.	.27	.10	.23	.09	.24
Lt. gluteus Med.	,36	.13	.32	.08	.26
Side Lateral Landing					
Rt. gluteus Max.	.32	.14	.24	.11	.30
Rt. gluteus Med.	.39	.13	.27	.08	.005*
Lt. gluteus Max.	.30	.11	.21	.08	.01*
Lt. gluteus Med.	.36	.14	.31	.09	.16

- * The mean difference is significant at the .05 level.

Female group:

A paired samples t-test was used for parametric variables and a Wilcoxon-Rank test for non-parametric variables, to determine whether there was a statistically significant mean difference between the right and left legs when female participants performed SLS, FL, SML and SLL. There was no significant difference in EMG activity for gluteus maximus and gluteus medius across all tasks, as shown in Table 5.4. Therefore, only the right leg was used to investigate the effect of tasks on EMG activity.

There was a significant effect of tasks across all EMG activity for both the G Max and G Med. For the G Max, EMG activity was significantly different, with a Wilks' lambda of 0.30, p = 0.001 and a multivariate partial eta squared of 0.69. For the G Med, EMG activity was also significantly different, with a Wilks' lambda 0.32, p = 0.001 and a multivariate partial eta squared of 0.67. Table 5.5 revealed the differences between each task, and EMG activity was significantly different between SLS and all landing tasks.

Table 5.4 EMG activity data differences between legs during SLS, FL, SML and SLL in the female group

Variables	Single-leg squat			Forward landing			Side medial landing			Side lateral landing		
	Rt.	Lt.	P value	Rt.	Lt.	P value	Rt.	Lt.	P value	Rt.	Lt.	P value
EMG RMS/MVIC												
Gluteus Max.	.56	.50	.08	.30	.25	.33	.29	.27	.62	.31	.30	.76
Gluteus Med.	.55	.54	.75	.40	.38	.50	.40	.37	.30	.39	.36	.32

- * The mean difference is significant at the .05 level.

Table 5.5 EMG activity data differences between SLS, FL, SML and SLL tasks in the female group

Variables	SLS vs. FL	SLS vs. SML	SLS vs. SLL	FL vs. SML	FL vs. SLL	SML vs. SLL
EMG RMS/MVIC	P value	P value	P value	P value	P value	P value
Rt. gluteus Max.	0.26 / 0.005*	0.25 / 0.005*	0.27 / 0.005*	0.01 / 0.65	0.01 / 0.82	0.02 / 1.00
Rt. gluteus Med.	0.15 / 0.005*	0.15 / 0.005*	0.14 / 0.005*	0.02 / .1.00	0.02 / 0.73	0.01 / 1.00
Lt. gluteus Max.	0.25 / 0.005*	0.27 / 0.005*	0.30 / 0.005*	0.02 / 1.00	0.05 / .82	0.03 / 1.00
Lt. gluteus Med.	0.16 / 0.005*	0.15 / 0.005*	0.14 / 0.005*	0.01 / 0.89	0.02 / 1.00	0.01 / 1.00

- * The mean difference is significant at the .05 level.
- Significant adjustment for multiple comparisons: Bonferroni.

Males group:

There was no significant difference in EMG activity for G Max and G Med across all tasks between legs, except the EMG activity of G Med during SLL for the right (M = .26, SD = .06) and the left (M = .31, SD = .08, t (16) = 2.29, p = .03, two tailed) leg (Table 5.6).

There was a significant effect of tasks across all EMG activity for both G Max and G Med. For G Max, EMG activity was significantly different, with a Wilks' lambda of .20, p < .005 and a multivariate partial eta squared of 0.80. For the right G Med, EMG activity was also significantly different, with a Wilks' lambda of .51, p = .02 and a multivariate partial eta squared of .49. Left G Med EMG activity was also significantly different, with a Wilks' lambda of .49, p = .01 and a multivariate partial eta squared of .50 (see Table 5.7), which revealed the differences between across all tasks. EMG activity was significantly different between SLS and all landing tasks.

Table 5.6 EMG activity data differences between legs during SLS, FL, SML and SLL in the male group

Variables	Single-leg squat			Forward landing			Side medial landing			Side lateral landing		
	Rt.	Lt.	P value	Rt.	Lt.	P value	Rt.	Lt.	P value	Rt.	Lt.	P value
EMG RMS/MVIC												
Gluteus Max.	.51	.45	.15	.26	.21	.13	.22	.22	.16	.26	.21	.06
Gluteus Med.	.52	.58	.40	.26	.27	.65	.28	.32	.10	.27	.31	.03*

- * The mean difference is significant at the .05 level.

Table 5.7 EMG activity data differences between SLS, FL, SML and SLL tasks in the male group

Variables	SLS vs. FL	SLS vs. SML	SLS vs. SLL	FL vs. SML	FL vs. SLL	SML vs. SLL
EMG RMS (%)	P value	P value	P value	P value	P value	P value
Rt. gluteus Max.	0.25 / 0.005*	0.21 / 0.005*	0.25 / 0.005*	0.04 / 1.00	0.001 / 0.82	0.04 / 1.00
Rt. gluteus Med.	0.26 / 0.005*	0.28 / 0.005*	0.27 / 0.005*	0.02 / 0.97	0.01 / 1.00	0.01 / 1.00
Lt. gluteus Max.	0.24 / 0.005*	0.25 / 0.005*	0.24 / 0.005*	0.01 / 1.00	0.003 / 0.84	0.01 / 1.00
Lt. gluteus Med.	0.31 / 0.005*	0.26 / 0.01*	0.25 / 0.005*	0.05 / 0.10	0.04 / 0.10	0.01 / 1.00

- * The mean difference is significant at the .05 level.
- Significant adjustment for multiple comparisons: Bonferroni.

5.4 Discussion:

The project's main goals were to identify if any relationships exist between G Max and G Med EMG activity and the biomechanical variables of active, healthy subjects during single-leg squats and multi-directional single-leg landing. The previous chapter explored the differences in kinematic and kinetic variables between legs, genders and tasks. Thus, it is important to report the differences in G Max and G Med EMG activity, to provide more information about how this relates to other biomechanical variables. Therefore, the goals of this chapter are to investigate the differences in G Max and G Med EMG activity between legs, tasks and genders during single-leg multi-directional landing and single-leg squat tasks.

The primary findings of this study include the following: no significant differences in G Max and G Med EMG activity between the right and left legs during all tasks apart from G Max during the SLS task ($p = .004$). EMG activity for the right G Max (51%) was higher than the left (45%), while EMG activity for the left G Med was almost identical when compared to the right. Furthermore, EMG activity differed significantly between SLS and all other landing tasks, but no differences were found between all landing tasks. Across all tasks, G Med EMG activity was higher than for G Max and ranged from 32% to 53%, except during the SLS task, which found that the right G Max EMG activity was higher than for G Med. Similar results were found in the literature during three different functional exercises (step down, side-step lunge and forward lunge) and EMG activity (14%, 13% and 12%, respectively) (Bouillon et al., 2012). Furthermore, Zazulak et al. (2005) reported G Med EMG activity in male and female groups ranging from 20% to 26%, while other studies reported different EMG activity percentages when normalised to MVIC. For example, for five lower limb tasks, the EMG activity of gluteal muscles ranged from 11% to 14% (Bouillon et al., 2012), and during landing EMG activity ranged from 12% to 26% (Zazulak et al., 2005). Lower results were found during the functional tasks (lunge, side-step and step down) (Boudreau et al., 2009). The study reported G Max and Med activity ranging from 11 to 14%.

Many factors might affect this variance in EMG activity, such as electrode placement and time frame taken to process the EMG amplitude. The current study used 100ms before landing and at least 2 seconds after initial contact for all landing tasks to collect raw EMG, with a 100ms window to smooth the raw signals. Others used different window lengths to smooth the signals, such as 25ms (Homan et al., 2013), 40ms (Zazulak et al., 2005), 75ms (Souza and Powers, 2009) and 125ms (Hollman et al., 2013). It has been suggested that 100ms to 200ms is a typical window to use (Criswell, 2011), as the smaller the time window, the less smooth the data will be. However, more research is needed to determine which window is better to use in smoothing EMG amplitude, especially in lower limb muscles during dynamic tasks.

There were differences in EMG activity between genders during the SLS task for the gluteus medius in both legs, and during landing there was a significant difference in G Med EMG activity when FL was performed, the right G Med during SML and the right G Med during SLL. These differences in EMG activity, especially in the G Med between females and males, might explain the excessive hip adduction angles in females when performing landing tasks – in line with the theory that the G Med works eccentrically to prevent this excessive adduction motion during landing (Powers, 2010). Previous literature compared EMG activity in individuals, but these studies are limited and it would be difficult to compare them because of the absence of method standardisation used for EMG collection. Two studies found no significant differences in G Med activation between genders (Zazulak, et al. 2005, Russell, et al. 2006). Zazulak et al. (2005) used a smaller sample size (13 females, nine males) and low statistical power, while Russell et al. (2006) recorded EMG activity on initial contact and maximum knee flexion angle during a single-leg drop jump task. However, as stated earlier, in the current study, EMG activity was recorded 100 milliseconds before initial contact until the subjects were fully balanced on a single leg. Furthermore, during SLS, no significant differences were found in G Max and G Med EMG activity (Zeller et al., 2003). Hart et al. (2007) and Hanson et al. (2008) found significant differences in G Med activity between males and females. The lack of comparative previous studies indicates a need for further research, since most of the comparative research in the literature found between strong and weak subjects, or between injured and non-injured groups. For example, Homan et al. (2013) investigated the influence of gluteal activation and knee kinematics with 82 healthy participants during double-leg jump landing tasks. The study stated that there were no differences between weak and strong muscles in knee abduction angle, but the weaker group showed greater muscle activation (Homan et al., 2013). Nonetheless, the studies were conducted during controlled double-leg landings, step downs and single-leg squats, and so it is difficult to predict whether these tasks are representative of those during which ACL injury occurs and if they can be compared to the data from more challenging tasks, such as single-leg landings from different directions.

Regarding the comparison of G Max and Med EMG activity between tasks, a significant effect was observed, especially between SLS tasks and other landing tasks. During SLS, EMG amplitude was significantly higher in G Max (0.54% and 0.48% for right and left, respectively) and G Med (0.51% and 0.53% for right and left) than in other landing tasks. This can be explained by the findings in the previous chapter, which revealed that during SLS the hip tends to demonstrate greater flexion and adduction angles, and thus more activity is likely to be needed to control this excessive motion. However, the relationship between them will be analysed and discussed subsequently in Chapter 7.

This study has some limitations regarding EMG activity. It would be difficult to control movement artefacts, which may affect the EMG activity. However, this has been solved by restrictedly and carefully following SENIAM guidelines for applying the surface EMG electrodes and ensuring they are in the proper position and data normalised to MVIC. Moreover, as stated earlier, the subjects in the current study were active and healthy, and so it is thus unclear if these findings could be generalised to a population with lower limb pathologies. Nonetheless, it seems that the findings are clinically relevant to a population at a high risk of ACL injuries.

5.5 Conclusion:

In conclusion, the study established EMG activity for an active, healthy population, which will provide more understanding when establishing the relationship between these findings and lower limb alignment while performing single-leg squats and single-leg multi-directional landing tasks. Across all tasks, no significant differences were found between the right and left legs, apart from EMG activity for the G Max during SLS. G Max EMG activity for the right leg was higher than the left, while there was almost identical EMG activity in the left G Med when compared to the right. Furthermore, EMG activity differed significantly between SLS and all other landing tasks, but no differences were found between all landing tasks. Across all tasks, G Med EMG activity was higher than G Max and ranged from 32% to 53%, except during the SLS task, where it was found that G Max EMG activity was higher than for the G Med. Furthermore, across all tasks, EMG activity for the G Med was significantly higher in females than in males, which might explain the greater hip adduction angles in females when performing landing tasks or might be because of differences in concentric and eccentric strength between genders. Therefore, the following chapter will investigate the differences in hip abductor and extensor concentric and eccentric strength.

.

Chapter 6

Concentric and Eccentric Strength Differences between Legs and Genders

6.1 Introduction:

This chapter aims to investigate differences in hip extensor and abductor strength between legs and between genders during isokinetic muscle testing (concentric and eccentric).

Study Hypothesis:

- There are no differences in hip extensor and abductor strength between limbs during the concentric and the eccentric phase.
- Differences in hip extensor and abductor strength between genders do exist during the concentric and the eccentric phase

6.2 Methods:

Thirty-four active, healthy participants, comprising an equal number of males and females, participated in this study. Socio-demographic characteristics of the participants are presented in Table 4.1. The same inclusion and exclusion criteria were used as discussed earlier in Chapter 3. Informed consent was obtained from the participants, who were asked to perform three repetitions of three strength sets. The testing speed for isokinetic tasks was 60°/sec (Claiborne et al., 2009, Julia et al., 2010, Myer et al., 2013). According to Perrin (1993), more concentric power can be produced at slower isokinetic speed, and as eccentric speed increases, the force will remain the same or might increase slightly. The testing orders were randomised. After isokinetic testing, participants were asked to perform three maximal voluntary isometric contractions for three seconds, with 30-second rest periods between them. The time between different muscle group tests was at least 5 minutes. All measurements were carried out by one examiner, and peak torque was corrected automatically for gravity using Biodex software, by taking a static torque at approximately 45° of the hip extension test, and 30° for the hip abduction test, prior to testing. Positioning and study procedures have been described previously in Chapter 3, as shown in Figures 3.1 and 3.2.

Statistical Analysis:

First, a Shapiro-Wilk test was used to assess whether the data were normally distributed (parametric or non-parametric). In addition, descriptive statistics (mean and standard deviation) were computed for each dependent variable in each task. A paired sample t-test and a Wilcoxon Rank test were used to explore the differences between legs for both parametric and non-parametric variables. The level of significance was set at p less than or equal to 0.05. The data were not normally distributed if values were equal to or less than 0.05. The mean values for the three trials of each test were calculated and compared to find the differences in performance between legs.

6.3 Results:

The normality test (Shapiro-Wilk) for isokinetic (concentric and eccentric) muscle testing variables revealed that all variables were normally distributed for both legs, except in the right and left hip extension concentric tests, as shown in Appendix (3). The paired sample t-test used for parametric variables and the Wilcoxon-Rand test for non-parametric variables found no significant differences in peak torque between right and left limbs across all concentric and eccentric results. The results are presented in Table 6.1 below.

Table 6.1 Isokinetic hip extension and hip abduction strength data differences between legs (females and males):

Test	**Rt.**	**Lt.**	**P value**
Peak Torque (N/M)	**M (SD)**	**M (SD)**	
Concentric Extension	164.09 (40.06)	167.08 (46.84)	.31
Eccentric Extension	176.14 (44.05)	178.57 (47.46)	,59
Concentric Abduction	92.69 (25.73)	96.98 (30.01)	.43
Eccentric Abduction	105.41 (28.43)	103.56 (24.29)	.55
Peak Torque/Body Weight (%)	**M (SD)**	**M (SD)**	
Concentric Extension	245.58 (43.82)	249.07 (49.60)	.51
Eccentric Extension	263.94 53.62)	267.04 54.52)	.65
Concentric Abduction	139.62 (36.68)	146.34 (43.80)	.42
Eccentric Abduction	159.41 (40.48)	156.35 (34.77)	.50

- The mean difference is significant at the .05 level.

Gender Differences:

There were significant differences in peak torque between females and males across all concentric and eccentric strength tests. For concentric extension, males' peak torque (M = 177.89, SD = 37.82) was significantly greater than females' peak torque (M = 150.29, SD = 38.40, t (32) = 2.11, p = .04, two-tailed). Also, for eccentric extension, males' peak torque (M = 191.10, SD = 42.67) was significantly greater than females' peak torque (M = 161.17, SD = 41.31, t (32) = 2.07, p = .04, two-tailed). Similarly, abduction concentric and eccentric strength tests were also significantly greater in males (p = .015 and p = .001, respectively) compares to females, as shown in Table 6.2.

Table 6. 2 Gender differences in isokinetic hip extension and hip abduction strength tests

Variables	Females		Males		P value PT – PT/BW
	Mean PT – (PT/BW)	SD PT – (PT/BW)	Mean PT – (PT/BW)	SD PT – (PT/BW)	
PT (N/M) – PT/BW (%)					
Rt. Concentric extension	150.29 – (232.57)	38.40 – (42.94)	177.89 – (258.60)	37.82 – (41.91)	.04* - .08
Rt. Eccentric extension	161.17 – (250.13)	41.31 – 55.10)	191.10 – (277.76)	42.67 – (49.90)	.04* - .13
Rt. Concentric abduction	82.18 – (128.11)	23.80 – (36.25)	103.20 – (161.14)	23.70 – (34.34)	.01* - .04*
Rt. Eccentric abduction	89.19 – (140.89)	14.53 – (29.21)	121.64 – (177.93)	29.93 – (42.41)	.005* - .006*
Lt. Concentric extension	149.47 – (230.60)	43.68 – (47.24)	184.69 – (267.55)	44.26 – (46.01)	.02* - .02*
Lt. Eccentric extension	160.59 – (248.71)	41.35 – (248.71)	196.55 – (285.36)	47.42 – 54.30)	.02* - .04*
Lt. Concentric abduction	83.57 – (130.48)	19.17 – (26.20)	110.40 – (162.20)	33.28 – 52.31)	.005* - .03*
Lt. Eccentric abduction	90.45 – (141.75)	13.49 – (21.14)	116.67 – (170.96)	25.88 – (39.93)	.005* - .01*

- The mean difference is significant at the .05 level.

Female group:

No significant differences were found in peak torque between the right and left limbs across all concentric and eccentric results in females, as shown in Table 6.3.

Male group:

No significant differences were found in peak torque between the right and left limbs across all concentric and eccentric results, as shown in Table 6.4.

Table 6.3 Isokinetic hip extension and hip abduction strength data differences between legs in the female group

Test	**Rt.**	**Lt.**	**P value**
Peak Torque (N/M)	**M (SD)**	**M (SD)**	**M (SD)**
Concentric Extension	150.29 (38.40)	149.47 (43.68)	.87
Eccentric Extension	161.17 (41.31)	160.59 (41.35)	,92

- * The mean difference is significant at the .05 level.

Concentric Abduction	82.18 (23.80)	83.57 (19.17)	.80
Eccentric Abduction	89.19 (14.53)	90.45 (13.49)	.78

- The mean difference is significant at the .05 level.

Table 6. 4 Isokinetic hip extension and hip abduction strength data differences between legs in the male group

Test	**Rt.**	**Lt.**	**P value**
Peak Torque (N/M)	**M (SD)**	**M (SD)**	**M (SD)**
Concentric Extension	177.89 (37.84)	184.69 (44.26)	.31
Eccentric Extension	191.10 (42.67)	196.55 (47.42)	,59
Concentric Abduction	103.20 (23.77)	110.40 (33.28)	.43
Eccentric Abduction	121.64 (29.93)	116.67 (25.88)	.55

- The mean difference is significant at the .05 level.

6.4 Discussion:

The objectives of the study reported in this chapter were to investigate the differences in hip abductors and hip extensors between legs and between genders during isokinetic muscle tests. The results demonstrated no significant differences in peak torque and peak torque normalised to body weight between right and left limbs across all concentric and eccentric test results. There were significant differences in peak torque between females and males across all concentric and eccentric strength tests, with male peak torque ($M = 177.89$, $SD = 37.82$) being significantly higher than female peak torque ($M = 150.29$, $SD = 38.40$) for concentric extension. Male peak torque ($M = 191.10$, $SD = 42.67$) was significantly greater than for the females ($M = 161.17$, $SD = 41.31$), applying also to eccentric extension. Similarly, abduction concentric and eccentric strength tests were also significantly greater in males ($p = .015$ and $p = .001$, respectively) compared to females. The torque was shown to range approximately between 20 and 30 N/m greater in male subjects. Previous studies reported similar relationships in hip abductor strength between males and females (Claiborne et al., 2006, Sugimoto et al., 2014). Sugimoto et al. (2014) reported hip abductor isokinetic tests across 36 (20 females, 16 males) collegiate athletes and found significant differences between male and female strength levels. Furthermore, concentric and eccentric torque of the hip abductors (38.5-39 N/m) was greater in men (Claiborne et al., 2006). However, Jacobs and Mattacola (2005) found that peak eccentric hip abductor torque relative to body weight was not different between recreationally active men and women. This might be because Jacobs and Mattacola (2005) used 120° per second as an angular velocity, though the current study, Sugimoto et al. (2014) and Claiborne et al. (2006) used 60° per second, testing below which is not recommended, because of the lack of functional significance and excessive compression and shear force on the knee (Wyatt and Edwards 1981), and fatigue may affect the results. For hip extension and abduction, it has been stated that 60° per second is a good representation of both the concentric and eccentric capabilities of each hip abductor and extensor (Boling et al., 2009). According to Perrin (1993), as velocity increases during eccentric contraction, the ability to produce force will remain the same or might slightly increase, and the muscle produces greater concentric force with slower velocity. With testing at speeds of more than 60°/sec, the chances of missing some resistance and forces might occur as a result of the high speed of the dynamometer.

With regards to hip extensors, the current study found significant differences in peak torque in both concentric and eccentric phases of isokinetic testing. However, when peak torque was normalised to body weight, no significant differences were found in right hip extensors. In contrast, Claiborne et al. (2006) found no significant differences between genders in peak torque. However, significant differences were found when the torque was normalised to body weight, possibly because strength testing was carried out from a standing position. It would be difficult to stabilise the pelvis and prevent postural deviation from a standing position, which might affect the results. The fact that females are lower in terms of strength has been considered as a risk factor for ACL injuries, because females demonstrate greater hip adduction and internal rotation. Although this has been supported by different studies in the literature (Hewett et al., 2010), other studies show different findings, which do not support this hypothesis (Beutler et al., 2009, Jacobs et al., 2007, Willson ans Davis, 2009). This could be due to measuring the hip extensors and abductors isometrically, which does not reflect muscle action during dynamic activities. Therefore, it would be important to measure strength in the concentric and eccentric phases, as this is more representative of the type of force required to control dynamic tasks.

This study was not without limitations. First, testing orders were randomised, which may have affected the results as a result of fatigue. However, a rest time of at least 5 minutes was given to all participants, and they were always asked if they were ready to be examined or not, to prevent muscle fatigue. Another factor that might have affected the results is motivational status, but the examiner tried to provide all the encouragement needed during all testing trials. Moreover, it was difficult to move the lever arm parallel to the participants' legs. All participants showed good ability through the range of motion, and the results of the reliability study showed medium to strong reliability in all testing positions, as presented in Chapter 3.

6.5 Conclusion:

In conclusion, the study found no significant differences in hip abductor and extensor peak torque between right and left limbs. However, significant differences in peak torque were found between genders, and similar results were found when peak torque was normalised to body weight, apart from right extension concentric and eccentric strength.

The findings of Chapters 4, 5 and 6 are useful when investigating the relationship between dynamic knee valgus and gluteal muscles, to determine which factor is related, i.e. EMG activity data or strength.

Chapter 7

The Relationship between Gluteus Muscle EMG activity and Strength with Lower Limb Biomechanical Variables

7.1 Introduction:

In the previous chapters, differences in kinematic, kinetic, EMG activity and strength measurements were explored, which helped in understanding how participants perform the tasks. This help in better understanding when determines which factor (EMG activity data or strength) will affect dynamic knee valgus. In order to answer the thesis' title, this chapter aims to:

1) Investigate the relationship between hip abductor and extensor muscle strength and lower limb biomechanical variables during single-leg squats and multi-directional single-leg landing.

2) Investigate the relationship between G Max and G Med EMG activity and lower limb' biomechanical variables during single-leg squats and multi-directional single-leg landing.

7.2 Methods

A total of 34 active, healthy participants, comprising 17 males and 17 females, participated in this study. This number was conducted from a pilot work using G* Power 3 software to provide a statistical power of 80% and an effect size of 0.44, as shown in Appendix (4). The sociodemographic details of the participants are presented in Table 4.1 in Chapter 4. The same inclusion and exclusion criteria were used as stated earlier in the reliability studies in Chapter 3. A consent form was read and signed by all participants before they took part in the study. Fifteen cameras (Qualisys, Gothenburg, Sweden) sampling at 240 Hz in a motion analysis system, and one force platform (AMTI BP400600, USA) sampling at 1200 Hz, embedded into the floor, were used to collect the kinematic and kinetic lower limb variables during the different tasks. G Max and G Med activity were recorded simultaneously using the 3D capture Noraxon Desktop DTS system (www.noraxon.com) at 1500 Hz, with a disposable self-adhesive Noraxon surface electrode fixed over the muscle (parallel to the muscle fibre).

The same instruments, filtration, calibration, markers list, training shoes, functional tasks and biomechanical model described earlier in the reliability studies in Chapter 3 were used.

Statistical Analysis:

The data were analysed using Statistical Package for Social Sciences (SPSS) version 21. First, a Shapiro-Wilk test was used to check whether the data were normally distributed or not (parametric or non-parametric). In addition, mean and standard deviations were calculated for each variable in each functional task. To explore the relationship between biomechanical variables and EMG activity for G Max, G Med, hip abduction isokinetic strength and hip extension isokinetic strength, Pearson's correlation coefficient (r) was used for parametric data, and a Spearman's rank correlation (ϱ) for non-parametric data. Furthermore, the coefficient of determination (R^2) was used in parametric data to represent the amount of variability in one screening test, which is explained by a second screening test (Swearingen et al., 2011). Table 5.1 illustrates the interpretation of the strength of correlation coefficients used in this study (Hopkins et al., 2009).

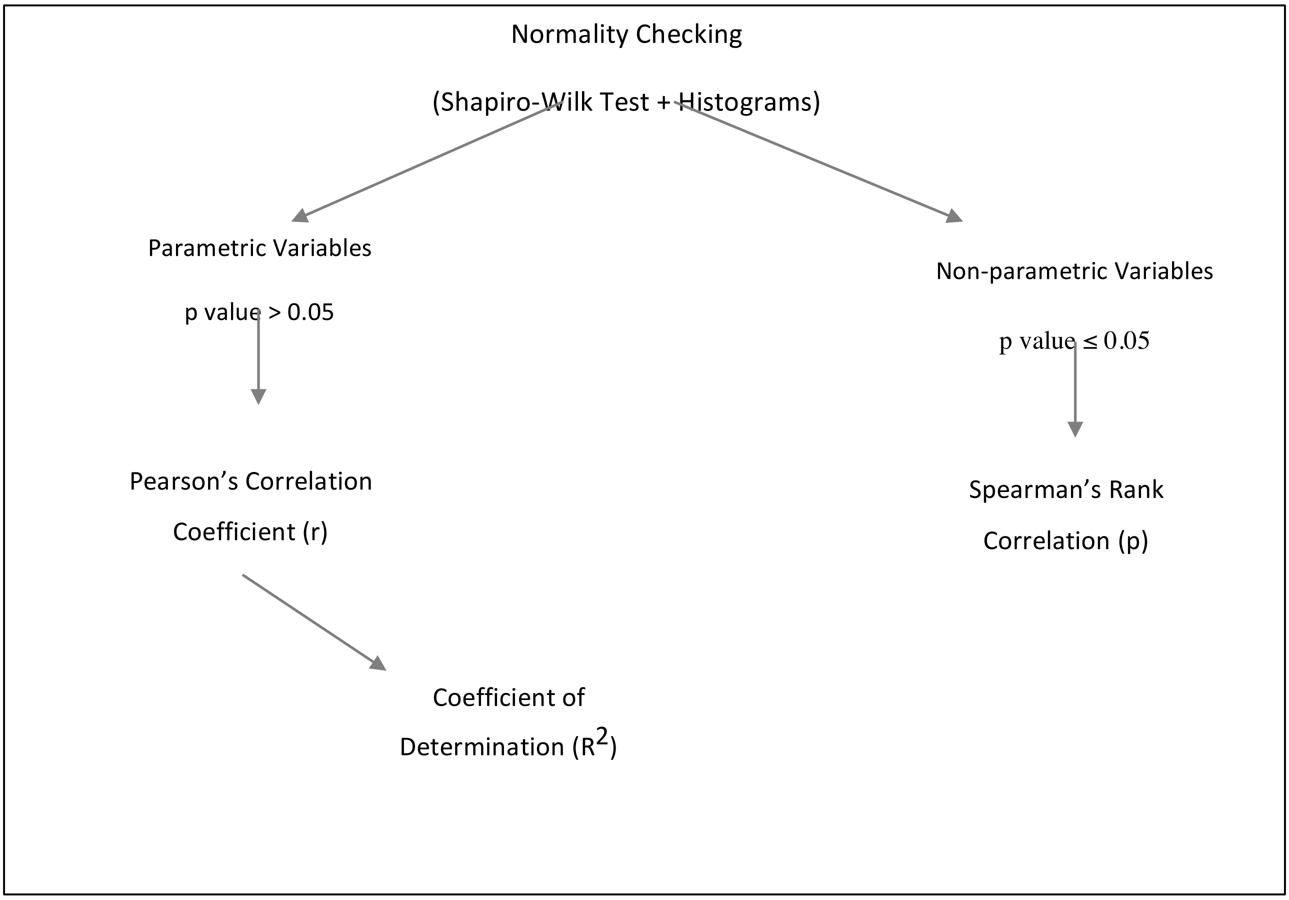


Figure 7.1 Statistical analysis outline for the correlation study

Table 7.1 **Correlation coefficient scores and level of association (Hopkins et al., 2009)**

Score	Level of Correlation
(.1 - .3)	Weak
(.3 - .5)	Moderate
(.5 - .7)	Strong
(.7 - .9)	Very strong
(.9 – 1)	Extremely strong

7.3 Results:

The normality test (Shapiro-Wilk) for kinetic and kinematic variables revealed that all variables were normally distributed for both legs, apart from left knee abduction moment in the SLS task, right knee abduction moment, left knee abduction moment and left internal hip rotation moment in the FL task, right hip flexion moment, right knee abduction moment, left internal hip rotation moment and left knee abduction moment in SML, and finally SLL data were normally distributed except for the right hip flexion moment, right knee abduction moment, left internal hip rotation moment, and the left knee abduction angle. See Appendices (3) for the normality tests for all variables.

Table 7.2 Descriptive analysis for the SLS task

Descriptive Statistics				
	Right		Left	
Variables	Mean	Std. Deviation	Mean	Std. Deviation
Hip Flex Angle	69.74	10.07	69.56	9.74
Hip Flex Moment	-1.06	0.38	-1.10	0.37
Hip Int. Rot Angle	7.43	6.88	9.98	8.12
Hip Int. Rot Moment	-0.41	0.15	-0.51	0.15
Hip Add. Angle	13.59	7.56	13.43	5.73
Hip Add. Moment	-0.98	0.23	-1.07	0.19
Knee Abduction Angle	-1.14	4.85	-0.57	4.11
Knee Abduction Moment	-0.05	0.19	-0.12	0.16
Knee Flex. Angle	83.33	7.02	82.39	7.80
Knee Flex. Moment	1.66	0.25	1.56	0.27
GRFV	1.12	0.09	1.12	0.09
GMax EMG	0.54	0.10	0.48	0.16
GMed EMG	0.54	0.18	0.51	0.11

Table 7.3 Descriptive analysis for the FL task

Descriptive Statistics				
	Right		Left	
Variables	Mean	Std. Deviation	Mean	Std. Deviation
Hip Flex Angle	57.34	12.65	54.85	13.27
Hip Flex Moment	-1.78	0.60	-1.77	0.71
Hip Int. Rot Angle	7.60	7.17	9.87	7.98
Hip Int. Rot Moment	-0.79	0.23	-1.03	0.29
Hip Add. Angle	7.70	6.77	7.54	4.08
Hip Add. Moment	-1.62	0.30	-1.85	0.34
Knee Abduction Angle	-1.61	5.13	-1.06	4.26
Knee Abduction Moment	0.16	0.25	0.13	0.20
Knee Flex. Angle	67.98	12.46	64.18	14.01
Knee Flex. Moment	2.78	0.47	2.70	0.42
GRFV	3.22	0.43	3.25	0.46
GMax EMG	0.29	0.16	0.23	0.12
GMed EMG	0.34	0.13	0.33	0.12

Table 7.4 Descriptive analysis for the SML task

Descriptive Statistics				
	Right		Left	
Variables	Mean	Std. Deviation	Mean	Std. Deviation
Hip Flex Angle	56.81	14.71	54.07	13.21
Hip Flex Moment	-1.75	0.63	-1.96	0.60
Hip Int. Rot Angle	9.11	7.49	11.50	8.59
Hip Int. Rot Moment	-0.72	0.20	-1.03	0.29
Hip Add. Angle	9.62	6.10	9.78	4.85
Hip Add. Moment	-1.50	0.29	-1.94	0.35
Knee Abduction Angle	-1.49	5.04	-1.72	4.95
Knee Abduction Moment	0.36	0.34	0.15	0.27
Knee Flex. Angle	66.83	11.63	63.74	10.76
Knee Flex. Moment	2.77	0.54	2.45	0.55
GRFV	3.26	0.52	3.24	0.52
GMax EMG	0.28	0.16	0.25	0.11
GMed EMG	0.34	0.13	0.35	0.11

Table 7. 5 Descriptive analysis for the SLL task

Descriptive Statistics				
	Right		Left	
Variables	Mean	Std. Deviation	Mean	Std. Deviation
Hip Flex Angle	54.31	13.68	53.38	15.00
Hip Flex Moment	-1.91	0.57	-1.76	0.53
Hip Int. Rot Angle	8.51	7.39	9.45	9.20
Hip Int. Rot Moment	-0.74	0.21	-0.97	0.24
Hip Add. Angle	6.21	6.20	5.65	4.95
Hip Add. Moment	-1.53	0.36	-1.93	0.35
Knee Abduction Angle	-2.49	5.10	-2.07	5.03
Knee Abduction Moment	0.31	0.28	0.12	0.17
Knee Flex. Angle	66.75	11.74	63.00	11.24
Knee Flex. Moment	2.66	0.55	2.46	0.54
GRFV	3.31	0.47	3.27	0.42
GMax EMG	0.29	0.13	0.26	0.11
GMed EMG	0.33	0.13	0.34	0.12

SLS task:

For the right SLS task, Table 7.6 reveals a strong negative correlation between knee abduction moment and hip abduction concentric strength ($r = -.50$, $p = .003$). In addition, hip abduction eccentric strength was negatively correlated with knee abduction moment and hip extension concentric strength ($r = -.44$, $p = .01$ and $\varrho = -.48$, $p = .004$, respectively). A moderate negative correlation was found between EMG activity of the G Med and the knee abduction angle ($\varrho = -.41$, $p = .01$), as shown in Table 7.6. However, during the left SLS task, no correlations were noted apart from G Med EMG activity and the internal hip rotation angle ($r = -.34$, $p = .04$), as shown in Table 7.7.

Table 7.6 Correlation between kinematics and kinetics with strength and EMG data during the right SLS task

Right SLS										
Variables	G Max	G Med	Ext. Con.	Ext. Ecc.	Ext. Con./BW	Ext. Ecc./BW	Abd. Con.	Abd. Ecc	Abd. Con./BW	Abd. Ecc./BW
Joint Angle (°)										
Hip Int. Rotation	r= .01 p= .95	ϱ= -.14 p= .40	ϱ= .29 p= .09	r= .06 p= .72	r= .29 p= .09	r= .06 p= .72	r= .26 p= .13	r= .25 p= .15	ϱ = .30 p = .07	r= .26 p= .12
Hip Adduction	r= -14 p = .40	ϱ= -.03 p= .83	ϱ= .09 p= .58	r= -.04 p= .81	r= .12 p= .47	r= .02 p= .89	r= .09 p= .57	r= .11 p= .52	ϱ = .17 p = .31	r= .20 p= .25
Knee Abduction	r=-.31 p = .07	ϱ= -.41 p= .01	ϱ= .15 p= .39	r= .04 p= .79	r= .15 p= .39	r= .05 p= .75	r= 32 p= .058	r = .29 p= .09	ϱ = .36 p = .06	r= .30 p= .08
Moments (Nm/Kg)										
Hip Int. Rotation	r= .02 p = .95	ϱ= -.18 p= .29	ϱ= -.12 p= .47	r= -.17 p= .45	r= -.05 p= .78	r= -.11 p= .51	r= -.21 p= .23	r= -.08 p= .64	ϱ = .16 p = .36	r= -.02 p= .90
Hip Adduction	r= .22 p= .19	ϱ= -.02 p= .90	ϱ= -.18 p= .29	r= -.07 p= .66	r= -.18 p= .29	r= -.05 p= .77	r= -.27 p= .11	r= -.27 p= .11	ϱ = .28 p= .10	r= -.28 p = .10
Knee Abduction	r= .17 p = .32	ϱ= .29 p= .09	ϱ= -.48 p= .004	r= -.26 p= .19	r= -.43 p= .01 R^2=0.18	r= -.23 p= .18	r= -.50 p=.003 R^2=0.25	r= -.44 p= .008 R^2=0.19	ϱ= -.46 p= .005	r= -.37 p= .02 R^2=0.13

- (ϱ) Spearman and (r) Pearson correlation coefficients, (R^2) coefficient of determination; correlation is significant at the level .05 (two-tailed) which has been highlighted. For hip flexion, knee flexion biomechanics and vertical ground reaction force results, see Sppendices (7).

Table 7.7 Correlation between kinematics and kinetics with strength and EMG data during the left SLS task

Left SLS										
Variables	G Max	G Med	Ext. Con.	Ext. Ecc.	Ext. Con./BW	Ext. Ecc./BW	Abd. Con.	Abd. Ecc	Abd. Con./BW	Abd. Ecc./BW
Joint Angle (°)										
Hip Int. Rotation	ϱ= -.16 p= .35	r= -.34 p= .04 R^2= 0.12	ϱ= .09 p= .85	r= .08 p = .64	r= .05 p = .74	r= .06 p = .73	r= -.25 p = .14	r= -.03 p = .84	ϱ= .28 p= .10	r= .06 p = .73
Hip Adduction	ϱ= -.16 p= .35	r= .12 p= .45	ϱ= -.31 p= .07	r= -.27 p= .11	r= -.26 p= .12	r= -.17 p= .30	r= -.26 p= .14	r= -.17 p= .32	ϱ= .15 p= .37	r= .02 p= .90
Knee Abduction	ϱ= -.05 p= .72	r= -.24 p= .17	ϱ= .09 p= .60	r= .07 p = .65	r= .14 p = .40	r= .14 p = .42	r= .09 p = .60	r= .26 p = .12	ϱ= .13 p= .45	r= .30 p = .07
Moments (Nm/Kg)										
Hip Int. Rotation	ϱ= -.15 p= .39	r= .03 p = .87	ϱ= .29 p= .08	r= -.29 p = .09	r= -.17 p = .31	r= .18 p = .28	r= -.31 p = .07	r= .15 p = .37	ϱ= -.14 p= .45	r= .03 p = .83
Hip Adduction	ϱ= .07 p= .68	r= .06 p= .70	ϱ= .10 p= .55	r= .02 p= .89	r= .10 p= .55	r= .03 p= .84	r= .02 p= 90	r= .01 p= .91	ϱ= .01 p= .85	r= .02 p= .87
Knee Abduction	ϱ= .07 p= .67	r= .24 p = .15	ϱ= -.13 p= .43	r= .05 p = .77	r= .06 p = .70	r= .08 p = .61	r= .07 p = .69	r= .12 p = .49	ϱ= .08 p= .61	r= .12 p = .47

- (ϱ) Spearman and (r) Pearson correlation coefficients, (R^2) coefficient of determination; correlation is significant at the level .05 (two-tailed) which has been highlighted. For hip flexion, knee flexion biomechanics and vertical ground reaction force results, see Sppendices (7).

FL task:

During the right FL, a strong negative correlation was noted between hip abduction eccentric strength and knee abduction moment (ϱ = -.65, p = .001), whereas a moderate negative correlation was found in hip abduction concentric strength and knee abduction moment (ϱ = -.47, p = .004), as shown in Table 7.8. During the left FL task, a moderate correlation was noted between G Med EMG activity and internal hip rotation moment (ϱ = .41, p = .02). The hip adduction angle was correlated with G Med EMG activity and hip abduction eccentric strength (r = .40 p = .01 and r = -.38 p = .02, respectively). Hip abduction eccentric strength also negatively correlated with knee abduction moment (r = -.48 p = .004). Furthermore, moderate negative correlations were found between the knee abduction angle and G Max (r = -.47, p = .004) and G Med (r = -.38, p = .01) EMG activity, as shown in Table 7.9.

Table 7.8 Correlation between kinematics and kinetics with strength and EMG data during the right FL task

Right FL										
Variables	G Max	G Med	Ext. Con.	Ext. Ecc.	Ext. Con./BW	Ext. Ecc./BW	Abd. Con.	Abd. Ecc	Abd. Con./BW	Abd. Ecc./BW
Joint Angle (°)										
Hip Int. Rotation	r= .04 p = .81	r= -.11 p = .51	ϱ= .26 p= .12	r= .02 p= .88	r= .16 p= .33	r= .08 p= .99	r= .20 p = .24	r= .15 p = .39	ϱ= .07 p= .62	r= .14 p = .40
Hip Adduction	r= .13 p= .46	r= .30 p= .07	ϱ= .16 p= .36	r= .10 p = .56	r= .32 p = .06	r= .21 p = .21	r= -.02 p= .88	r= -.16 p= .36	ϱ= .03 p= .83	r= -.09 p= .60
Knee Abduction	r= .09 p = .58	r= -.27 p = .11	ϱ= .23 p= .17	r= .06 p= .73	r= -.22 p= .12	r= .06 p= .65	r= .33 p = .055	r= .32 p = .06	ϱ= .12 p= .47	r= .32 p = .058
Moments (Nm/Kg)										
Hip Int. Rotation	r= .06 p = .71	r= .01 p = .95	ϱ= .12 p= .47	r= -.16 p= .35	r= -.15 p= .38	r= -.10 p= .56	r= .09 p = .58	r= .01 p = .97	ϱ= .07 p= .43	r= .05 p= .76
Hip Adduction	r= .16 p = .36	r= -.04 p = .80	ϱ= .00 p= .99	r= .11 p = .51	r= .11 p = .50	r= .07 p = .65	r= .11 p = .51	r= .21 p = .22	ϱ= .18 p= .67	r= .11 p = .53
Knee Abduction	ϱ= .07 p= .67	ϱ= .28 p= .10	ϱ= -.48 p= .004	ϱ= -.22 p= .19	ϱ= .07 p= .67	ϱ= .07 p= .67	ϱ= -.47 p= .005	ϱ= -.65 p= .001	ϱ= .07 p= .67	ϱ= .07 p= .67

- ((ϱ) Spearman and (r) Pearson correlation coefficients, (R^2) coefficient of determination; correlation is significant at the level .05 (two-tailed), which has been highlighted. For hip flexion, knee flexion biomechanics and vertical ground reaction force results, see Appendices (7).

Table 7.9 Correlation between kinematics and kinetics with strength and EMG data during the left FL task

Left FW										
Variables	G Max	G Med	Ext. Con.	Ext. Ecc.	Ext. Con./BW	Ext. Ecc./BW	Abd. Con.	Abd. Ecc	Abd. Con./BW	Abd. Ecc./BW
Joint Angle (°)										
Hip Int. Rotation	r= .07 p = .65	r= .07 p = .65	ϱ= .11 p= .51	r= .01 p = .92	r= .09 p = .60	r= .05 p = .67	r= .20 p = .25	r= .01 p = .98	ϱ= .17 p= .31	r= .05 p = .77
Hip Adduction	r= .07 p = .65	r= .40 p= .01 R^2= 0.16	ϱ= -.36 p= .03	r= -.31 p = .07	r= .26 p = .12	r= .23 p = .17	r= .23 p = .18	r= -.38 p= .02 R^2= 0.14	ϱ= .14 p= .40	r= -.25 p = .14
Knee Abduction	r= -.47 p= .004 R^2=0.22	r= -.38 p= .01 R^2= 0.14	ϱ= .13 p= .45	r= .17 p = .31	r= .22 p = .19	r= .25 p = .14	r= .06 p = .71	r= .33 p = .051	ϱ= .08 p= .64	r= .35 p= .03 R^2=0.12
Moments(Nm/Kg)										
Hip Int. Rotation	ϱ= .09 p= .61	ϱ= .02 p= .89	ϱ= .18 p= .30	ϱ= .09 p= .58	ϱ= .17 p= .31	ϱ= .03 p= .85	ϱ= .05 p= .67	ϱ= .05 p= .74	ϱ= .04 p= .79	ϱ= .18 p= .29
Hip Adduction	r= .07 p = .65	r= .07 p = .65	ϱ= .24 p= .16	r= .33 p = .052	r= .28 p = .10	r= .33 p = .054	r= .22 p = .19	r= -.48 p= .004 R^2= 0.23	ϱ= .17 p= .32	r= .42 p= .01 R^2=0.17
Knee Abduction	ϱ= .07 p= .67	ϱ= .07 p= .67	ϱ= .11 p= .52	ϱ= .05 p= .74	ϱ= .20 p= .24	ϱ= .09 p= .57	ϱ= .04 p= .67	ϱ= .13 p= .35	ϱ= .07 p= .67	ϱ= .16 p= .32

- (ϱ) Spearman and (r) Pearson correlation coefficients, (R^2) coefficient of determination; correlation is significant at the level .05 (two-tailed), which has been highlighted. For hip flexion, knee flexion biomechanics and vertical ground reaction force results, see Appendices (7).

SML task:

During the SML task, a moderate correlation was noted between the hip adduction angle and right G Med (r = .47, p = .005) and left G Med (r = .39, p = .02) EMG activity. During the right leg task, the knee abduction angle also correlated with hip extension concentric (ϱ = -.47, p = .004), hip abduction concentric (r = .33, p = .04) and hip abduction eccentric strength (r = .38, P =.04), as shown in Table 7.10. Eccentric strength of hip abduction also negatively correlated with knee abduction moment (ϱ = -.47, p = .01). In the left leg, the knee abduction angle strongly negatively correlated with hip abduction eccentric strength (r = -.51, p = .002) and moderately with abduction concentric strength (r = .35, p = .04), as shown in Tables 7.10 and 7.11.

Table 7.10 Correlation between kinematics and kinetics with strength and EMG data during the right SML task

Right SML										
Variables	G Max	G Med	Ext. Con.	Ext. Ecc.	Ext. Con./BW	Ext. Ecc./BW	Abd. Con.	Abd. Ecc	Abd. Con./BW	Abd. Ecc./BW
Joint Angle(°)										
Hip Int. Rotation	r= .26 p = .13	r= .07 p = .65	ϱ= .25 p= .14	r= .11 p = .57	r= .01 p = .92	r= .20 p = .25	r= .22 p = .19	r= .21 p = .24	ϱ= .17 p= .31	r= .25 p = .15
Hip Adduction	r= .19 p = .26	r= .47 p= .005 R^2=0.22	ϱ= .03 p= .86	r= .05 p = .74	r= -.31 p = .07	r= .23 p = .18	r= .19 p = .26	r= .19 p = .26	ϱ= .14 p= .40	r= .02 p = .87
Knee Abduction	r= .18 p = .28	r= -.20 p = .25	ϱ= .23 p= .18	r= .17 p = .33	r= .17 p = .31	r= .06 p = .71	r= .33 p= .04 R^2=0.11	r= .38 p= .02 R^2=0.16	ϱ= .39 p= .02	r= .41 p= .01 R^2=0.16
Moments(Nm/Kg)										
Hip Int. Rotation	r= .02 p = .90	r= .03 p = .68	ϱ= .22 p= .19	r= .25 p = .15	r= .17 p = .15	r= .12 p = .45	r= .25 p = .15	r= .18 p = .30	ϱ= .17 p= .31	r= .01 p = .15
Hip Adduction	r= .05 p = .75	r= .04 p = .75	ϱ= .03 p= .85	r= .02 p = .87	r= .02 p = .87	r= .07 p = .37	r= .13 p = .45	r= .03 p = .85	ϱ= .14 p= .40	r= .02 p = .87
Knee Abduction	ϱ= .08 p= .62	ϱ= .17 p= .31	ϱ= .23 p= .17	ϱ= .18 p= .29	ϱ= .31 p= .06	ϱ= .22 p= .19	ϱ= .15 p= .39	ϱ= -.47 p= .01	ϱ= .15 p= .39	ϱ= .39 p= .03

- (ϱ) Spearman and (r) Pearson correlation coefficients, (R^2) coefficient of determination; correlation is significant at the level .05 (two-tailed), which has been highlighted. For hip flexion, knee flexion biomechanics and vertical ground reaction force results, see Appendices (7).

Table 7.11 Correlation between kinematics and kinetics with strength and EMG data during the left SML task

Left SML										
Variables	G Max	G Med	Ext. Con.	Ext. Ecc.	Ext. Con./BW	Ext. Ecc./BW	Abd. Con.	Abd. Ecc	Abd. Con./BW	Abd. Ecc./BW
Joint Angle(°)										
Hip Int. Rotation	r= .19 p= .27	r= .09 p= .58	ϱ= .04 p= .80	r= .03 p = .86	r= .06 p = .67	r= .05 p = .77	r= .18 p = .28	r= .05 p = .77	ϱ = .19 p = .27	r= -.04 p = .80
Hip Adduction	r= .22 p= .20	r= .39 p= .02 R^2=0.15	ϱ= .31 p= .07	r= .21 p = .22	r= -.31 p = .06	r= -.20 p = .24	r= -.42 p= .01 R^2=0.18	r= -.51 p= .002 R^2=0.26	ϱ= .39 p= .02	r= -.47 p= .004 R^2=0.22
Knee Abduction	r= .06 p= .73	r= .07 p= .67	ϱ= .18 p= .300	r= .18 p = .28	r= .15 p = .36	r= .18 p = .30	r= .07 p = .68	r= .26 p = .13	ϱ = .02 p = .29	r= .18 p = .35
Moments(Nm/Kg)										
Hip Int. Rotation	ϱ= -.12 p= .47	ϱ = .01 p= .20	ϱ= .25 p= .14	ϱ= .18 p= .28	ϱ= .32 p= .06	ϱ= .42 p= .06	ϱ= .15 p= .39	ϱ= .32 p= .06	ϱ= .42 p= .06	ϱ= .15 p= .39
Hip Adduction	r= .13 p = .48	r= .02 p= .88	ϱ= .24 p= .15	r= .30 p = .08	r= .18 p = .29	r= .30 p = .08	r= .25 p = .14	r= .35 p= .04 R^2=0.12	ϱ = .16 p = .35	r= -.26 p = .13
Knee Abduction	ϱ= -.03 p= .58	ϱ= .22 p= .20	ϱ= .23 p= .18	ϱ= .18 p= .28	ϱ= .32 p= .06	ϱ= .42 p= .06	ϱ= .15 p= .39	ϱ= .03 p= .87	ϱ= .18 p= .28	ϱ= .32 p= .06

- (ϱ) Spearman and (r) Pearson correlation coefficients, (R^2) coefficient of determination; correlation is significant at the level .05 (two-tailed), which has been highlighted. For hip flexion, knee flexion biomechanics and vertical ground reaction force results, see Appendices (7).

SLL task:

During the right SLL task, it was noted that G Med EMG activity was correlated with the hip adduction angle ($r = .34$, $p = .04$). A negative moderate correlation was noted between hip extension concentric strength and the internal hip rotation angle ($\varrho = -.36$, $p = .01$). Furthermore, a negative moderate correlation was noted between internal hip rotation moment and hip extension eccentric strength ($r = -.37$, $p = .03$), hip abduction concentric ($r = -.46$ $p = .006$) and hip abduction eccentric strength ($r = 0.47$ $p = .005$), as shown in Table 7.12. During left side testing, the hip adduction angle strongly correlated with hip abduction eccentric strength ($r = -.51$, $p = .002$), and moderately correlated with hip abduction concentric strength ($r = -.38$, $p = .02$), while hip adduction moment moderately correlated with hip abduction eccentric ($r = .44$, $p = .008$) and negatively correlated with G Med ($r = -.39$, p .02) EMG activity, as shown in Tables 7.12 and 7.13.

.

Table 7. 12 Correlation between kinematics and kinetics with strength and EMG data during the right SLL task

Right SLL										
Variables	G Max	G Med	Ext. Con.	Ext. Ecc.	Ext. Con./BW	Ext. Ecc./BW	Abd. Con.	Abd. Ecc	Abd. Con./BW	Abd. Ecc./BW
Joint Angle (°)										
Hip Int. Rotation	r= .06 p = .73	r= .15 p = .38	ϱ= .19 p= .26	r= .02 p = .88	r= .21 p= .28	r= .07 p = .68	r= .16 p= .28	r= .15 p = .85	ϱ= .23 p= .28	r= .19 p = .27
Hip Adduction	r= .13 p = .45	r= .34 p= .04 R^2=0.12	ϱ= .07 p= .67	r= .04 p = .79	r= .20 p= .28	r= .19 p = .85	r= .18 p= .28	r= .19 p = .34	ϱ= .10 p= .28	r= .09 p = .85
Knee Abduction	r= .37 p= .07	r= .20 p = .23	ϱ= .32 p= .06	r= .11 p = .50	r= .25 p= .18	r= .09 p = .34	r= .34 p= .04 R^2=0.12	r= .37 p= .03 R^2=0.14	ϱ= .33 p= .05	r= .34 p= .04 R^2=0.12
Moments(Nm/Kg)										
Hip Int. Rotation	r= .03 p = .91	r= .14 p = .40	ϱ= -.36 p= .03	r= -.37 p= .03 R^2=0.14	r= -.32 p= .06	r= -.20 p = .25	r= -.46 p= .006 R^2=0.21	r= -.47 p= .005 R^2=0.22	ϱ= -.33 p= .055	r= .32 p = .06
Hip Adduction	ϱ= -.07 p= .69	ϱ= -.07 p= .66	ϱ= .01 p= .96	ϱ= .05 p= .77	ϱ= .06 p= .46	ϱ= .11 p= .13	ϱ= .15 p= .39	ϱ= .09 p= .54	ϱ= .29 p= .09	ϱ= .27 p= .11
Knee Abduction	ϱ= -.22 p= .19	ϱ= .11 p= .51	ϱ= -.41 p= .01	ϱ= .05 p= .77	ϱ= .06 p= .46	ϱ= .11 p= .13	ϱ= .15 p= .39	ϱ= .09 p= .54	ϱ= .29 p= .09	ϱ= .27 p= .11

- (ϱ) Spearman and (r) Pearson correlation coefficients, (R^2) Coefficient of determination; correlation is significant at the level .05 (2-tailed) which has been highlighted. For hip flexion, knee flexion biomechanics and vertical ground reaction force results, see appendices (7).

Table 7.13 Correlation between kinematics and kinetics with strength and EMG data during the left SLL task

Left SLL										
Variables	G Max	G Med	Ext. Con.	Ext. Ecc.	Ext. Con./BW	Ext. Ecc./BW	Abd. Con.	Abd. Ecc	Abd. Con./BW	Abd. Ecc./BW
Joint Angle (°)										
Hip Int. Rotation	r= .16 p = .34	r= .03 p = .84	ϱ = .12 p = .49	r= -.03 p = .84	r= -.03 p = .84	r= -.03 p = .84	r= .11 p= .57	r= .07 p= .86	ϱ = .12 p = .49	r= .07 p= .69
Hip Adduction	r= -.15 p= .37	r= .37 p= .03 R^2=0.13	ϱ = .12 p = .49	r= -.16 p= .34	r= -.16 p= .34	r= -.17 p= .32	r= -.38 p= .02 R^2=0.14	r= -.51 p= .002 R^2=0.26	ϱ= -.37 p= .03	r= -.49 p= .003 R^2=0.24
Knee Abduction	ϱ = .07 p = .62	ϱ = .06 p = .72	ϱ = .12 p = .49	ϱ = .03 p = .49	ϱ = .12 p = .49	ϱ = .12 p = .49	ϱ = .12 p = .49	ϱ = .12 p = .49	ϱ = .12 p = .49	ϱ = .36 p = .03
Moments(Nm/Kg)										
Hip Int. Rotation	r= -.28 p = .37	r= .19 p = .73	ϱ = .12 p = .49	r= -.11 p = .51	r= -.03 p = .84	r= -.03 p = .84	r= -.13 p = .84	r= .17 p= .33	r= .22 p= .19	r= .05 p= .76
Hip Adduction	r= .01 p= .91	r= -.39 p= .02 R^2=0.15	ϱ = .12 p = .49	r= -.11 p= .51	r= -.17 p= .32	r= -.17 p= .32	r= -.22 p= .32	r= .44 p= .008 R^2=0.2	ϱ = .12 p = .49	r= .45 p= .007 R^2=0.2
Knee Abduction	ϱ = .12 p = .49	ϱ = .12 p = .49	ϱ = .12 p = .49	ϱ = .07 p = .49	ϱ = .12 p = .49	ϱ = .12 p = .49	ϱ = .12 p = .49	ϱ = .12 p = .49	ϱ = .12 p = .49	ϱ = .12 p = .49

- (ϱ) Spearman and (r) Pearson correlation coefficients, (R^2) coefficient of determination; correlation is significant at the level .05 (two-tailed), which has been highlighted. For hip flexion, knee flexion biomechanics and vertical ground reaction force results, see Appendices (7).

Female Group:

During the right SLS task, a strong negative correlation was found between the hip adduction angle and G Med EMG activity (r = -.65, p = .005, R^2= 0.42) among the females. In addition, the hip adduction angle correlated with hip abduction eccentric strength (r = .59, p = .01, R^2= 0.35)). However, the knee abduction angle correlated with hip abduction concentric strength (r = .55, p = .02, R^2= 0.3). Hip adduction moment correlated with G Med EMG activity and hip abduction eccentric strength (r = .52, p = .03, R^2= 0.27 and r = .60, p = .01, R^2= 0.36, respectively). Moreover, knee abduction correlated with hip abduction concentric strength (r = .60, p = .01, R^2= 0.36). During the left SLS task, correlation was noted between the knee abduction angle and hip extension eccentric strength as normalised (r= = .48, p= .04, R^2= 0.23). Internal hip rotation moment correlated with hip extension concentric strength (ϱ= .60, p= .01), as shown in Appendices (5).

During the right FL, a very strong correlation was noted between hip abduction concentric strength and the knee abduction angle (ϱ = .75, p = .005), whereas knee abduction correlated with G Max EMG activity (ϱ = -.49, p = .04), hip abduction concentric strength (ϱ = -.49, p = .04) and hip abduction eccentric strength (ϱ = -.54, p = .02). During the left FL, G Max EMG activity negatively correlated with the knee abduction angle (r = -.56. p = .01, R^2= 0.31), as shown in Appendices (5).

During the right SML task, a moderate correlation was noted between the internal hip rotation angle and hip abduction concentric as normalised strength (r = .49, p = .04, R^2= 0.24) In addition, a strong correlation was observed between the knee abduction angle and hip abduction concentric strength (r = .55. p = .02, R^2= 0.3) and hip abduction eccentric strength (r = .56. p = .01, R^2= 0.31), as shown in Appendices (5).

Finally, during the right SLL task, strong correlations were noted between the knee abduction angle and hip abduction concentric strength (r = .73, p = .001 R^2= 0.53) and hip abduction eccentric (r = .68, p = .002 R^2= 0.46). Knee abduction moment correlated with hip extension concentric strength (ϱ = -.48, p = .004). However, no significant correlations were found during left SLL, as shown in Appendices (5).

Male Group:
During the right SLS task, a strong correlation was found between the hip adduction angle and hip extension concentric strength (r = .64, p = .005, R^2= 0.41), hip extension eccentric strength (r = .62, p = .007, R^2= .38) and hip abduction eccentric strength (r = .60, p = .01, R^2= 0.36). In addition, the hip adduction angle moderately correlated with hip abduction concentric strength (r = .48, p = .04, R^2= 0.23). Another large negative correlation was noted between the knee abduction angle and G Med EMG activity (ϱ = -.65, p = .004). However, the hip adduction angle and moment correlated with G Med EMG activity (r = -.56, p = .01, R^2= 0.31, r = .62, p = .008, R^2= 0.38, respectively), as shown in Appendices 5).

During FL in the male group, correlations were found only in the right leg. The hip adduction angle strongly correlated with hip abduction concentric (r = .60, p = .01, R^2= 0.36) and hip abduction eccentric strength (r = .52, p = .03, R^2= 0.27), as shown in Appendices (5).

During the right SML task, correlations were noted between the hip adduction angle and right G Med EMG activity (r = .55, p = .02, R^2= 0.3), hip abduction concentric (r = .56, p = .01, R^2= .31) and hip abduction eccentric strength (r = .64, p = .003, R^2= 0.41). Other correlations were found between internal hip rotation moment and right G Max EMG activity (ϱ = .49, p = .04) , hip abduction eccentric (r = -.59, p = .01) and hip extension eccentric strength (r = -.54, p = .02). On the other hand, during the left SML, G Med EMG activity strongly correlated with the hip adduction angle (ϱ = -.67, p = .009), as shown in Appendices (5).

Finally, during the right SLL task, the hip adduction angle strongly correlated with hip abduction concentric strength (r = .55, p = .02, R^2= 0.3) and abduction eccentric strength (r = .64, p = .001, R^2= 0.41). A negative strong correlation was noted between internal hip rotation moment and hip abduction eccentric strength (r = -.58, p = .01, R^2= 0.34). Furthermore, a negative moderate correlation was noted between G Med EMG activity and knee abduction moment (ϱ = -.48, p = .003). Hip abduction eccentric correlated with internal hip rotation moment (r= -.58, p= .01, R^2= 0.34). However, during the left leg SLL task, a strong correlation was found between the internal hip rotation angle and G Max EMG activity (r = .55 p = .02, R^2= 0.3). Furthermore, a correlation was found between the hip adduction angle and abduction eccentric strength (r = -.49 p = .04, R^2= 0.24), while hip abduction eccentric strength also correlated with hip abduction moment (r= .52, p= .03, R^2= 0.27).

7.4 Discussion:

The project's goal was to explore the relationship between strength and G Max and G Med EMG activity muscles and the lower limb biomechanical variables in active, healthy subjects during single-leg squats and multi-directional single-leg landing. The study found moderate to strong relationships between gluteal muscles strength and EMG activity, and lower limb biomechanical variables depending on the tasks ranging from (r= -0.51) to (r= 0.33), as presented previously in Tables 7.6 to 7.13. Some of the current findings were similar to the findings of previous studies, regardless of differences in the methodological tools and the participants, as most previous studies focused on female participants only (Hollman et al., 2009, 2013, Homan et al., 2013, Jacobs and Mattacola, 2005). Several studies have found a relationship between hip strength and knee valgus motion, with r ranging from -.10 during bilateral drop landing (Homan et al., 2013) to -0.61 during single-leg hop when females were tested separately (Jacobs and Mattacola, 2005). Other researchers reported a correlation between EMG amplitude and knee frontal plane motion, with r ranging from -0.28 to -0.45 during bilateral landing and step down tasks, respectively, among females (Hollman et al., 2009, Hollman et al., 2013), which will be discussed later in this chapter.

SLS Relationship:

During SLS, the current study noted significant correlations, specifically between the knee abduction angle and G Med EMG activity on the right side (ϱ= -.41, p = .01). However, the only correlation found on the left side was between G Med EMG activity and the internal hip rotation angle (r = -.34, p = .04 and R^2= 0.12). These findings support the study hypothesis that gluteal muscle EMG activity correlates with certain extent kinematic variables during SLS. However, in contrast to our hypothesis, EMG activity did not relate to kinetics. The differences between right and left legs could be explained by the differences in kinematics and kinetics between legs when participants performed the tasks, as explained earlier in Chapter 4. Moreover, differences in performance, and the difficulty in controlling squat depth between limbs, might have had an effect. The left internal rotation angle was significantly greater than the right (9.97 and 7.43°, p = .03). A difference was also found in the knee abduction angle, with the right limb greater than the left, albeit not significantly (-1.14 and -

.56, p = .31). The findings of this current study during SLS are different from those of a significant negative correlation between concentric hip abduction and the knee abduction angle (r= -0.37, R^2= 0.13) by Claiborne et al. (2006). However, Claiborne et al. (2006) used six 3D cameras to measure knee kinematics and did not include knee kinetics or muscle activity. In addition, the examiners used a standing position to test hip abduction strength isokinetically, which might have led to more effort in the contralateral side to stabilise the body (Jacobs and Mattacola, 2005). In addition, the reliability results of testing from this position were not reported. Hollman et al. (2014) examined the relationship between hip muscle strength and G Max and Med EMG activity in 41 females during a single-leg squat task. The study found that the gluteus maximus may modulate with knee frontal motion (partial r = 0.35). However, Hollman et al.'s (2014) study cannot be generalised, as it was carried out only on young females, and the study used isometric strength as the comparator, which might not reflect the nature of the strength interaction during dynamic tasks. Moreover, the use by Hollman et al. (2014) of a dominant leg only may affect the results, as the other side may differ in performance, as reported in the previous chapter.

When female data during SLS was analysed separately, G Med EMG activity and hip abduction eccentric strength correlated with the hip adduction angle (r= -0.65, and 0.59 with R^2 0.42 and 0.35 respectively), and the knee abduction angle correlated with hip abduction concentric strength (r = .55, R^2= 0.3) when testing the right side. A relationship was also found between the knee abduction angle and hip extension eccentric strength, normalised to body weight (r= = .48, R^2= 0.23). This was not surprising, as the females demonstrated greater hip adduction and knee abduction angles, with G Med trying to control this excessive motion. An additional explanation might be differences in the way SLS was performed in Hollman et al. (2014), as their participants completed five consecutive SLSs. In this situation, fatigue might affect the results. Another previous study reported a relationship between hip external rotation strength and the frontal plane projection angle (FPPA) during an SLS task (Willson et al., 2006). However, others found no relationship between hip external rotation strength and the knee abduction angle (Claiborne et al., 2006). The different results could be due to the use of a 2D camera by Willson et al. (2006) to measure knee kinematics, compared to the 3D cameras used in this study and Claiborne et al. (2006). In addition, Willson et al. (2006) used isometric strength in their study.

When examining the male group separately, it was found that during the right SLS, G Med

EMG contributed to controlling the knee abduction angle by approximately 42% of variance. Moreover, hip abduction and extension eccentric strength strongly correlated with the hip adduction angle (r= .62 and .60) with R^2 0.4 and 0.38). However, on the left side, G Med EMG contributed to controlling the hip adduction angle by approximately 31% of variance. It would therefore appear that the nature of movement in male subjects is less influenced by strength and activity than in females during SLS.

With regards to kinetics variables, the current study reported a correlation between knee abduction moment and hip abduction concentric and eccentric strength and extension concentric strength (r= -.50, -.44, and -.48 with R^2 0.25, 0.20 and 0.34, respectively) when examining both groups. No other study has examined the relationship between hip strength and/or gluteal muscle EMG activity with lower limb kinetic variables in active, healthy subjects, which makes direct comparison with others studies difficult. However, it has been reported in the literature that increased knee abduction moment may be a risk factors in ACL damage (Chappell, Yu, Kirkendall, and Garrett, 2002, Hewett et al., 2005). Clinically this could help in injury prevention. For example, if clinicians increase the eccentric strength of the hip abductors and extensors, knee abduction moment may decrease.

FL Relationship:

During the right FL, no correlations were found between kinematics variables in the frontal and transverse planes with gluteal EMG activity or hip abductor or extensor strength. However, when the left side was tested, the hip adduction angle correlated with G Med EMG activity and hip abduction eccentric strength (r = .40 and -.38 with R^2= 0.16 and 0.14, respectively). Furthermore, moderate negative correlations were found between the knee abduction angle and G Max (r = -.47 with R^2= 0.22) and G Med (r = -.38 with R^2= 0.14) EMG activity. Like the SLS task, the knee abduction angle correlated with EMG activity, thereby supporting the hypothesis that variance in EMG activity is associated with frontal knee motion.

Similar results were found in the literature in the work by Hollman et al. (2013), who examined hip extension strength and G Max recruitment on 40 females during double-leg landing. Both gluteus maximus strength and activation were associated with frontal knee motion (r = 0.21 and .13) (Hollman et al., 2013). In addition, the current study showed that the knee abduction angle strongly correlated with G Max EMG activity on the left side when females were tested separately (r = -.56 with R^2= 0.31). However, double-leg landing tasks were not representative of those activities during which ACL injuries actually occur, which might explain the low correlation found in the study by Hollman et al. (2013). Another drawback in the study by Hollman et al. (2013) was the isometric measurement of hip extension strength, as the difficulty of the task might increase the muscle function required to control lower limb alignment. Another study was conducted by Homan et al. (2013) to investigate the influence of hip strength on gluteal activation and knee kinematics in 82 healthy participants during double-leg jump landing tasks. The study stated that no differences were found between weak and strong groups in knee abduction motion, although the weaker group showed greater muscle activation (Homan et al., 2013). The study did not report any correlation which might also be explained by the task used. Another study with only female participants was conducted to measure knee kinematics during single-leg step downs (Hollman et al., 2009), finding that G Max activation has more of an effect than strength on knee valgus while stepping down. However, the study used 2D cameras to measure knee kinematics and did not use tasks representative of those during which ACL damage occurs. This implies that its findings cannot be compared to studies involving more challenging tasks such as single-leg medial or lateral landings.

With regards to kinetic variables, in the right leg a strong negative correlation was noted between hip abduction eccentric strength and knee abduction moment (ϱ = -.65), where a moderate negative correlation was found in hip abduction concentric strength and hip extension concentric with knee abduction moment (ϱ = -.47 and ϱ = -.48, respectively). Similar results found during the right SLS task indicate that hip abductor and extensor strength might decrease knee valgus, moment which in turn might help in ACL injury prevention, as increased knee valgus moment may be a risk factors in ACL injury (Chappell et al., 2002, Hewett et al., 2005). While testing the left side, correlations were found between hip abduction eccentric strength and hip adduction moment (ϱ = -.48), and between G Med EMG activity and internal hip rotation moment (ϱ = .41). Moreover, strength variables were negatively correlated with the ground reaction force in both legs, ranging from r= .34 to r= -.62. It has been reported in the literature that increased vertical ground reaction force may be a risk factors for ACL damage (Hewett et al., 2005). Therefore, the negative correlations between strength and reaction force might explain the importance of isokinetic hip abductors and extensors in reducing vertical ground reaction force during landing. It has been stated that 19% of body kinetics were absorbed when landing softly (Zazulak et al., 2005).

Furthermore, hip eccentric extension strength is responsible for absorbing 22% of kinetics (Devita and Skelly, 1992). As with SLS, there no other studies have examined the relationship between hip strength and/or gluteal muscle EMG activity with lower limb kinetic variables during forward single-leg landing in active, healthy subjects, which makes any comparison with other studies difficult, except for a study carried out recently by Malloy et al. (2016). The study reported a correlation between hip external rotators (not hip abductor muscles) with peak hip external rotation moments ($r = 0.47$, $p = 0.021$), greater hip frontal plane excursion ($r = 0.49$, $p = 0.017$), during single-leg landing and cutting tasks. This supported the work of Lawrence et al. (2008), who reported that females who generated greater hip external rotator and knee muscle strength significantly decreased vertical ground reaction force when landing from a 40-cm high step (Lawrence et al., 2008). The drawback of the Lawrence and Malloy studies, though was that they measured strength isometrically, which might not reflect the nature of dynamic tasks.

SML Relationship:

It has been proposed that single-leg landing is associated with ACL injuries (Boden et al., 2009). Many sports involve multi-directional motions controlled on a single-leg, which makes the investigation of biomechanical variables during medial and lateral side landing important. The current study found a moderate relationship between the hip adduction angle and G Med EMG activity ($r= .47$ with R^2 0.22) during a right SML task. On the left side, a similar correlation was noted between G Med data and the hip adduction angle ($r= .39$ with $R^2 = 0.15$). This was not surprising, as it was reported in the previous chapters that the hip adduction angle and G Med EMG activity were not different between the right and the left legs. Moreover, during the right SML, the knee abduction angle also correlated with hip extension concentric strength ($\varrho = -.47$), hip abduction concentric and eccentric strength ($r= .33$ and $.38$, respectively) with R^2 (0.11 and 0.14, respectively). Similar findings have been reported in the literature. Suzuki et al. (2015), for instance, used side medial landing from a 20-cm box to assess knee kinematics on 43 college basketball players (20 males and 23 females). The study reported that hip extension and hip abduction strength negatively correlated with the knee abduction angle ($r= -.48$ and $-.46$, respectively) on initial contact only but not when measuring the peaks. However, again, it would be better if isokinetic muscle strength had been measured instead of isometric, to give more understanding of how the muscles work concentrically and concentrically to control landing. In addition, an isometric strength test was done without external fixation rather than measuring concentric and eccentric strength, which makes these particular strength measures questionable. Furthermore, Suzuki et al. (2015) did not report any kinetic data and did not include EMG activity in their study. McCurdy et al. (2014) reported a negative relationship between hip extension strength and the knee abduction angle during unilateral drop landing and single-leg squats in females. In addition, the same finding was reported during an SLS in the literature (Stickler et al., 2015). However, both studies used isometric strength and measured only female participants. Stickler et al. (2015) used a frontal projection angle to measure knee kinematics. Both studies included only the dominant leg in the study, as different performance might be found between legs.

When female data were analysed separately, the current study found a strong correlation between the knee valgus angle and hip abduction concentric (r= .55) and eccentric strength (r= .56). It seemed that using appropriate muscle testing alongside more difficult tasks to perform explains the differences in the results. Moreover, the current study found that hip abduction eccentric strength negatively correlated with knee valgus moment (r = -.47 with R^2= 0.22), which was also found also during the right SLS and FL. Another study found no correlation between hip muscle strength and medial hopping and landing (Itoh et al., 2016). However, the study tested strength isometrically and did not include hip adduction motion in the kinematic variables, even though it is important in forming the dynamic knee valgus. Neither Suzuki et al. (2015) nor Itoh et al. (2016) measured muscle activity in their studies. The current study also found that, during the left SML, the hip adduction angle negatively correlated with hip abduction concentric and eccentric strength (r= -.42 and -.51 with R^2 0.17 and 0.26, respectively). This might be due to the need for more strength to prevent the excessive movement involved in hip adduction, which can be found also during the left FL and differences in biomechanical variables between legs according to the difficulty of the task. Moreover, it was noted while testing that participants had poor balance after side landing, possibly because of poor core and pelvic stability, as suggested by Powers (2010), who reported the link between lack of core and pelvic and lower limb injuries.

With regards to kinetics, the current study reported that left hip adduction moment moderately correlated with hip abduction eccentric strength (r= .35 and R^2= 0.12). It has been reported previously that hip adduction moment strongly correlated with knee abduction moment in subjects who suffered from ACL injury, but not with those who did not have injuries (Hewett et al., 2005). Limited studies are found in the literature regarding medial landing tasks, which makes any comparison difficult. Finally, there were negative correlations between hip strength and ground reaction force, apart from hip extension concentric strength during a right side SML. However, no correlations were found with the left SML. This might be due to landing on a single leg, which needs a smaller base of support and reduces stability. The study suggested that more research is needed to investigate the relationship between kinetics and strength and/or gluteal muscle activity during landing tasks.

SLL Relationship:

Almost similar results were found in both SML and SLL. The hip adduction angle was moderately correlated with G Med EMG activity (r= .34 and R^2= 0.11) during the right SLL task. However, on the left side, the hip adduction angle was negatively strongly correlated with hip abduction eccentric strength (r= -.51 and R^2= 0.26) and moderately correlated with hip abduction concentric strength (r= -.38 and R^2= 0.14). Moreover, during the right SLL, G Max EMG activity moderately correlated with the knee abduction angle (r= .37, R^2= 0.13) and hip abduction concentric and eccentric strength (r= .34 and r= .37 with R^2= 0.11 and 0.13). A negative moderate correlation was found between hip extension concentric strength and the internal hip rotation angle (ϱ= -.36 and R^2= 0.12). A comparison of our results with others was difficult, as only a single study analysed knee kinematics and kinetics during both side medial and side lateral landing and how they relate to hip extension and external rotation strength in male rugby players (Itoh et al., 2016). The study found that during side lateral landing, the knee abduction angle was significantly higher than the side medial. This finding is similar to ours, in that the knee valgus angle was higher in SLL for all tasks. Itoh et al. (2016) used Biodex to measure hip extension strength isokinetically and a hand-held dynamometer to measure external hip rotation strength, without measuring hip abduction strength. In addition, Itoh et al. (2016) did not include EMG in their study. When examining the relationship between strength and kinematics in males separately during both medial and lateral landing, the current study found strong correlations between hip abduction concentric and eccentric strength and the hip adduction angle, and the knee abduction angle strongly correlated with hip abduction concentric strength.

It seemed that conflicting results in the previous literature related to differences in the methods used, such as strength tests (isometric, concentric, and eccentric), population, leg involved, EMG activity and task used. In the current study, concentric and eccentric strength were assessed, because during functional tasks it is difficult to correlate isometric strength with dynamic movement (Sigward et al., 2008, Willson and Davis, 2008, Jacobs et al., 2007), especially in landing and squatting, as the gluteal muscles are responsible for working eccentrically to control the excessive adduction and internal rotation during landing (Neumann, 2010). Therefore, it is difficult to compare our results with others, as few studies have assessed concentric and eccentric strength (Claiborne et al., 2006, Jacobs and Mattacola, 2005) or investigated their relationship with lower limb biomechanics. None of the previous studies found significant correlations between eccentric strength and kinematics variables, which can be explained by the use of 2D to measure kinematics in the study by Jacobs and Mattacola (2005), and the measurement of hip muscle strength from a standing position in the study by Claiborne et al. (2006). Measuring from a standing position is still in the form of an open kinetic chain, as the non-stance (non-weight-bearing) leg will be tested. It has been reported that measuring hip abduction from a standing position will stress the hip muscles bilaterally and affect the validity of the test (Jacobs and Mattacola, 2005). To reduce all of these effects, it would be better to test hip muscle strength from a lying position, which might ensure that the upper trunk is steady against the testing chair. This would also reduce the load on the non-testing limb.

With regards to tasks, it would be better to use a task that is challenging enough and would reflect the scenario of injury during a sports competition. This was reported in Olsen et al.'s (2004) study, as video analysis showed side motion during landing on one leg. Using similar tasks in the study would increase the ability to control lower limb alignment, thus differentiating muscle function on lower limb biomechanics. Most of the previous studies used single-leg landing in a forward direction to investigate the relationship between hip muscle strength or gluteal function, apart from three studies (Itoh et al., 2016, Malloy et al., 2017, Suzuki et al., 2015). However, multi-directional motions are required in sport activities, and it is therefore important to investigate the factors that influence lower limb alignment. Results from investigating the relationship in frontal and transverse planes during FL, SML and SLL would be more relevant to sport activity tasks than bilateral landing or single-leg tests for the sagittal plane only, and so addressing this relationship might help in designing interventions to prevent these injuries in both genders.

With regards to EMG, Merletti and Parker (2004) stated that sampling EMG signals need to be at least double the frequency recorded, in order to help reduce noise. A bandpass filter with a high pass filter, which must be over 20 Hz, and a low pass filter, mainly around 400 to

450 Hz, is needed to smooth noise. Lesser signals contain unwanted artefacts in surface electrodes, so it is important to understand that a decreasing high frequency or an increasing low frequency may affect the EMG signal being collected. Therefore, the current study used 20-450 Hz as a bandpass filter. Moreover, the window used to smooth the raw signals might have an effect on variance in the literature, though it has been suggested that 100ms to 200ms is considered a typical window to use (Criswell, 2011). However, more research is needed on which window is better to use in smoothing EMG amplitude, especially in lower limb muscles during dynamic tasks. The current study recorded EMG activity starting from 100 milliseconds before initial contact and 2 seconds after landing, to make sure subjects were fully balanced on a single leg, taking into consideration data before heel strike, and also investigated the relationship between EMG capacity produced and lower limb biomechanical variables. Homan et al.'s (2013) study used a similar method, albeit the study recorded EMG activity only during the load phase. Moreover, using an appropriate task to examine the relationship could be vital, as muscle activity might be influenced by the task. Boudreau et al. (2009) stated that task used in a study actually influences G Max and G Med activity. The study reported higher peak EMG activity in SLSs that lunge and step up. Other studies have investigated gluteal muscle activity and how it differs between genders. Two studies found no significant differences in gluteus medius activity in this regard (Zazulak, et al. 2005, Russell, et al. 2006), while Zazulak et al. (2005) found differences in gluteus maximus activity in females during landing tasks. However, neither study used 3D motion analysis to examine the kinetics and kinematics, and so it would be difficult to state the effect of muscle activity on the knee joint.

The findings of our study suggest that a relationship may exist between hip extension and abduction strength, gluteal muscle EMG activity and lower limb biomechanical variables, depending on the tasks and on which sides they are performed. It seems that strength or activity are not the only factors affecting lower limb kinematics and kinetics, as different tasks produced different results. Trunk position during single-leg tasks might also have an effect on dynamic knee valgus. It has been observed that injured athletes had a 16° greater trunk lean that uninjured athletes (Hewett, Torg, and Boden, 2009), because the ground reaction force will follow the centre of mass, which will shift as a result of lateral trunk lean and produce greater knee abduction moment (Hewett and Myer, 2011). However, EMG activity was found to correlate with kinematics variables in most cases, and hip strength correlated mostly with kinetic variables. Correlations between gluteal muscles and lower limb biomechanical variables differed when each gender was examined separately. This can be explained by the differences between genders in performance – as reported in the previous chapters. The current study found that the knee abduction angle was correlated with hip abduction concentric (r= .55, R^2= 0.30) and eccentric strength (r= .56, R^2= 0.31) during a right SML. However, a correlation was not found on the left side, which can be explained by the differences in task performance between legs. During single-leg landing tasks, previous studies have also reported gender differences in the influence of hip muscle strength on knee kinematics (Leetun, Ireland, Willson, Ballantyne, and Davis, 2004, Suzuki et al., 2015) and suggested that the relationship between strength and knee injuries may differ depending on gender. Suzuki et al. (2015) reported the relationship between the knee abduction angle and hip strength in a female group, but a correlation was actually found between hip external rotation strength and the knee flexion angle.

Moreover, as stated in Chapter 4, gender differences have been noted in females also performing SLS tasks with a higher knee abduction angle, hip adduction angle and knee abduction moment. Moreover, there was no significant difference in G Max and G Med EMG activity in either leg in the female group during the SLS task, as mirrored by Zeller et al. (2003). Differences in knee abduction could be explained not only by the significant differences in muscle strength between genders, as reported in the previous chapter, but also because of the anatomical differences between the two sexes. During landing tasks, when examining each gender separately in our study, the main findings were strong negative correlations found between the hip adduction angle and G Med EMG activity (r = -.65, p = .005, R^2= 0.42) in the right side for the female group. In addition, the hip abduction angle was correlated with hip abduction eccentric strength (r = .59, p = .01, R^2= 0.34). However, the knee abduction angle correlated with hip abduction concentric strength (r = .55, p = .02, R^2= 0.30).

Current studies suggest that exercise and screening tasks would be better gender-based, as males demonstrated higher strength in all measures, even when taking body weight into consideration, except during right hip extension. Moreover, EMG activity differed between genders, especially gluteus medius data across all tasks. However, females usually demonstrated a higher percentage when normalised to MVC. This might be due to the need for greater muscle activity for weaker muscles to compensate for mechanical weakness. This finding has been reported by Homan et al. (2013), who stated that the weaker group demonstrated higher gluteal muscle activity. Another explanation for gender differences when performing tasks might be the effect of anatomical differences. As reported in the literature, female and male anatomical structures are different (Hewett et al., 2006). Furthermore, a higher Q angle might predict poor landing (Beutler et al., 2009), and females produce a higher Q angle with a higher pelvic tilt and a higher genu recurvatum than males (Beutler et al., 2009). It seems gender differences have an effect on the current findings, as differences exist between males and females in kinematic, kinetic, strength and gluteal muscle EMG activity markers. It has been theorised that weaker muscles produce greater activation to display better results as a way of compensating for any weakness exhibited (Enoka and Stuart, 1992). When the female group was considered as the weak group in the current study, due to the significant difference in strength, as previously reported in Table 6.4, it was noted that they produced a greater percentage of EMG activity across tasks than the male group. Greater muscle activation in the weaker group was also reported by Homan et al. (2013).

There are several limitations to the current study. First, it was carried out on active, healthy subjects aged between 18 and 35 years. Therefore, the study can only be generalised to this age group, but it is unclear if the activity level or age affected the results or if the findings are applicable to athletes who have experience in landing. Moreover, as stated earlier, it is unclear if these findings can be generalised to a population with lower limb pathologies. However, the findings of the current study would be clinically relevant to this population, as ACL injuries occur in this population and dynamic tasks that are relevant to ACL injury scenario were used in this study. The second limitation concerns EMG activity. As previously reported, it would be difficult to control movement artefacts that may affect EMG activity. However, an increase in EMG amplitude was noted while measuring the MVC for G Max and G Med. This was solved by carefully following SENIAM guidelines on applying surface EMG electrodes, ensuring they were in the proper position and making sure that data were normalised to MVC. However, regardless of the limitation stated, the current results indicate that gluteal muscle EMG amplitude may play a role in kinematics variables, and hip strength may play a role in shock absorption and moments, especially in hip and knee frontal and transverse plane motion. It has been stated that dynamic knee valgus is a modifiable biomechanical risk factor resulting in ACL injury, so clinically the findings would be important in preventing ACL injuries by using rehabilitation programmes to improve muscle strength of the hip and EMG amplitude of gluteal muscles. It would also be beneficial for future studies to carry out rehabilitation programmes especially for subjects who have poor lower limb biomechanics, and to investigate if improving strength and activation would modify kinematics and kinetics during multi-directional single-leg landing. Moreover, the current study found that differences exist between genders, which may explain why females have a higher rate of ACL injuries than males. Furthermore, differences between legs were noted in performances and relationships. Future research needs to examine if leg dominance does indeed play a role, as 31 subjects in the current study were right leg-dominant.

7.5 Conclusion:

In conclusion, in a healthy and active population, relationships exist between hip extension and abduction strength, as well as gluteal muscle EMG activity and lower limb biomechanical variables during SLS, FL, SML, and SLL tasks, although the findings were different between tasks and on which side they were performed. EMG activity was found to correlate with kinematics variables in most tasks, and gluteus medius EMG activity moderately correlated with the hip adduction angle in several landing tasks in both legs. Furthermore, strength usually correlated with moments and ground reaction force, depending on the task and the leg involved in the study. This can be explained by differences in kinetics data between tasks and between limbs.

Chapter 8

Summary, Conclusion and Recommendations

8.1 Summary:

ACL injuries are significant, affect both genders and can occur from non-contact mechanisms. One modifiable biomechanical risk factor that has been widely researched recently is dynamic knee valgus, which is a combination of hip adduction, internal hip rotation, knee abduction and external tibia rotation, which is believed to stress the ACL during landing tasks (Hewett et al., 2005). It has been hypothesised that gluteal maximus and medius strength can modify lower limb biomechanics by eccentrically controlling this excessive motion (Claiborne et al., 2006, Hollman et al., 2009).

From the literature, the relationship between gluteal muscles and lower limb biomechanics is still unclear and conflicting, regardless of the methodological tools used in previous studies. The systematic review in this research, presented in Chapter 2, revealed that most studies have investigated the relationship between the isometric strength of hip abductors and/or extensors and lower limb biomechanics. However, during dynamic tasks such as landing, the muscles are required also to work concentrically and eccentrically. Only a few studies have investigated the concentric and eccentric strength of hip muscles and their relationship to landing biomechanics, with the basis of these studies being hypothesis that strong hip musculature might work eccentrically to prevent excessive hip adduction and internal rotation, thus preventing ACL injury (Claiborne et al., 2006, Jacobs and Mattacola, 2005). However, athletes with strong hip musculature still sustain ACL injuries, thereby highlighting the need to investigate the relationship between the EMG activity of gluteal muscles and lower limb biomechanics, not just strength in isolation. This might be because the level of activation is more important than strength in predicting lower limb biomechanics during dynamic tasks such as landing. Therefore, it is important to investigate both the strength and activity data of the gluteus maximus and medius, as each factor might be important in the control of lower limb biomechanics. However, no study has looked at this subject during single-leg multi-directional landing and single-leg squats.

Therefore, the aim of this thesis was to explore the role of gluteal muscles during dynamic tasks, especially landing on a single leg from different directions, as this is a common scenario in different sporting activities and does lead to ACL damage. This unique work appears to be the first to use single-leg landing from a different direction whilst simultaneously recording the EMG activity of the gluteal muscles and also assessing the relationship to eccentric and concentric strength, to identify the risk factors in ACL injury. In order to achieve this aim, the current thesis had specific elements with specific aims:

1) To examine the within- and between-days reliability of the isokinetic muscle strength testing of hip abductors and extensors.
2) To determine the electromyography activity consistency of gluteal maximus and gluteus medius and biomechanical variables during single-leg squats and multi-directional single-leg landing tasks.
3) To investigate the kinetics and kinematic of lower limbs joints during single-leg squats and multi-directional single-leg landing tasks.
4) To investigate the electromyography activity of gluteus maximus and medius during single-leg squats and multi-directional single-leg landing tasks.
5) To investigate the concentric and eccentric strength of the gluteal maximus and medius muscles.
6) To explore the relationship between lower limb biomechanics and gluteal muscles during single-leg squats and multi-directional single-leg landing.

8.2 Conclusion:

Regarding the within- and between-day reliability of isokinetic muscle, the study found that the majority of the ICC values were good to excellent across all results. The ICC value for both hip extension and abduction were higher for the within-day (0.62 – 0.98) than the between-days (0.59 – 0.93) reliability. Concentric and eccentric tests did not correlate with isometric strength tests, so they were included in the study, which was not surprising, as the muscles work concentrically and eccentrically during functional tasks.

The second aim was to examine the consistency of lower limb biomechanical variables and the gluteus maximus and gluteus medius during single-leg squats and single-leg multi-directional landing. Across all tasks, the study showed good to excellent ICC values in kinematic variables. However, kinetic variables demonstrated higher levels of variability compared to kinematics, though their ICC values were fair to excellent. A possible explanation for this may be the dynamic nature of the tasks, as subjects must fully be balanced after landing, which might be affected by the trunk motion. On the other hand, EMG activity reliability was tested, to ensure correct electrode fixation and to determine the consistency of the EMG activity of the gluteus maximus and medius during the tasks. All ICC values during the landing tasks were excellent, albeit squatting showed less reliability. In the SLS task, high ICC was in the right gluteus medius (0.84), and the lowest was in the right gluteus maximus (0.60). This could be explained by dynamic instability, because of the associated movement while in the ascending and descending phases. The results of the first and second studies increased confidence in the ability to collect reliable data, when following the instructions for measurement described in Chapter 3, thereby making assessing relationships in the main study more likely to yield valid results. In addition, from the SEM values, it could be determined if the differences between limbs, tasks or genders were greater than the measurement error of the test, which gives a better understanding of the true differences between these elements.

In order to achieve the main aims, it was important to investigate how participants performed the tasks. Another reason was to determine whether there was symmetry between limbs, so one leg can define another's performance. In addition, if differences do exist, it might give a better clinical and biomechanical understanding of the influence of gluteal muscles, in order to control dynamic knee valgus. Therefore, the third and fourth studies' aimed to determine if there were differences between limbs and genders in kinetic and kinematic (Chapter 4) and EMG activity data (Chapter 5) variables when performing single-leg squats and single-leg multi-directional landing. The study found that differences exist between limbs, especially in knee abduction, hip adduction and internal hip rotation moments. This indicates that limb symmetry is not as important as previously reported, especially knee abduction, moment which was significantly different during SLS, SML and SLL. The right leg demonstrated greater knee valgus moments than the left across all landing tasks, apart from forward single-leg landing, which showed no difference. ACL injured players demonstrated higher knee abduction moment than uninjured counterparts during landing (Hewett et al., 2005). Unfortunately, because of the numbers of right-legged individuals, it was not possible to look properly at the influence of leg dominance.

Knee valgus moment was significantly higher during all landing tasks than SLS, indicating that researchers should utilise SML and SLL, in order to measure dynamic knee valgus and help predict future ACL injury risks, as SLS may provide a load of insufficient magnitude.

Differences were also found between genders, with females squatting and landing with greater knee valgus and hip adduction angles. Not surprisingly, these differences have been reported in the literature as risk factors in the higher incidence of ACL injury in females. However, different tasks were included in this thesis.

While investigating the EMG activity of the gluteus maximus and medius, no significant difference was found between the right and left limbs, apart from the EMG activity of gluteus maximus during SLS, whereby it was higher for the right than for the left, though EMG activity of the left gluteus medius was lower when compared to the right. Furthermore, gluteus medius EMG activity was significantly higher in females, which might explain the excessive hip adduction angles in females when performing landing or squatting tasks.

Regarding the fifth aim, the study conducted to investigate the concentric and eccentric strength of the gluteal maximus and medius muscles found no significant difference in the right or left lower limbs. However, significant differences were found between genders in peak torque, and similar results were found when peak torque was normalised to body weight, apart for right hip extension concentric and eccentric strength. However, differences in the left side were not significant. When considering the female group as the weaker group, this finding supported Homan et al.'s (2013) study, which found that the weaker group produced higher EMG activity levels during landing. However, more participants were needed in each group, in order to confirm this finding.

Finally, in order to answer the title of the thesis, the sixth aim of the study was to investigate the relationship between gluteal muscles and lower limb biomechanics, which demonstrated significant moderate correlations between gluteus medius EMG activity and hip adduction angles during all landing tasks, with R^2 ranging from 0.12 to 0.22 apart from during the right FL. Moreover, gluteus medius EMG activity moderately correlated with knee valgus angle during right SLS and with internal hip rotation angle during left SLS. However, gluteus maximus EMG activity moderately correlated with the knee valgus angle during left FL only and did not correlate with any transverse motion angles. However, when each group was examined separately, several moderate to strong correlations were found between gluteus maximus EMG activity and motion in the frontal and transverse planes. Another finding in the current study was the significant moderate to strong correlations between hip abductor and extensor strength and the knee valgus angle, the hip adduction angle, knee valgus moment, hip adduction moment and internal hip rotation moment. Moreover, strength negatively correlated moderately to strongly with the ground reaction force in both legs, which ranged from 0.34 to 0.62.

The results also showed that female participants performed the tasks with significantly higher hip adduction and knee valgus angles, both of which might predict ACL injury (Hewett et al., 2005) and may partly explain why females have higher ACL injury rates than males. When examined, the female participants were assessed separately, and higher correlations between hip adduction and knee abduction angles with G Max EMG activity/Med and abduction concentric and eccentric strength were found, which ranged between r= 0.55 and 0.75 for all tasks.

Consequently, targeting gluteal muscle to influence dynamic knee valgus during single-leg landing, neuromuscular training of the gluteal muscles may reduce movement contributing to dynamic knee valgus and possibly decrease injury risk. However, the relationship appears to be limb-, gender- and task-dependent, and the weak to moderate correlations found herein indicate that other factors might have an effect as well in controlling dynamic knee valgus, such as trunk and ankle motions.

There are number of limitations in the study. First, peak strength was measured during the strength assessment for both concentric and eccentric, but some subjects might produce sub-maximum strength. However, practice trials, motivation and rest periods were always offered to the participant, to make sure they produced maximum force. Second, the study did not include trunk motion or centre of mass, which might have a role in lower limb biomechanics during multi-directional single-leg landing, thus increasing the risk of ACL injury. Limited research in this regard, though, was found when starting this thesis. Third, the study included active, healthy subjects with different levels of sporting ability. It would be difficult to generalise the results to athletes in a specific sport or to injured subjects, as results from other populations may be differ. Moreover, using average performance may not give the full picture of performance, due to within-subject variability in performance for an individual task. Finally, it was difficult to compare between dominant and non-dominant legs, to determine the effect of leg dominancy, because more than 90% of the subjects had right-limb dominance.

8.3 Recommendations

8.3.1 Recommendation for Practice

The results of our study indicate that interventions targeting hip neuromuscular control may play a role in improving knee biomechanics, especially dynamic knee valgus. These programmes might include ACL prevention and rehabilitation strategies, particularly for those who have poor lower limb biomechanics that have been analysed during functional tasks. However, caution should be urged, as the results were task-, limb- and gender-dependent. Therefore, it would be appropriate to utilise hip neuromuscular control programmes with other programmes such as visual verbal feedback on landing strategies from different directions. A combination of these protocols might help in reducing ACL injury risks rather than traditional open-chain strengthening programmes. The inverse relationship found in sagittal plane motion (SLS and FL) between gluteus medius EMG activity and hip adduction and knee abduction suggests that interventions to improve muscle activation such as explosive training might be important. However, in frontal plane tasks (SML and SLL), almost similar gluteus medius EMG activity failed to reduce the hip adduction motion positively correlated with the hip adduction angle. This might suggest the need for more activity produced by muscle might help in mimicking and controlling the motion. More research is needed to confirm this notion.

Moreover, the results from Chapter 4 indicate that each limb should be examined separately, without using the other as a control, because knee abduction moment was significantly different between limbs, especially during SML and SLL. In addition, a high knee abduction angle and moment have been considered an important ACL injury risk factor during landing (Hewett et al., 2005). This finding might lead clinicians and researchers to use different tasks with different directions to screen the performance of limbs or genders, in order to predict ACL injury risk factors, as knee abduction and hip adduction biomechanics differed in most cases.

8.3.2 Recommendation for Further Studies

Based on the results of this thesis, several questions were raised which require further investigation.

- Having established that the level of gluteal muscle activity can account for almost 20% of the variance in dynamic knee valgus, future work should investigate what other factors contribute in dynamic knee valgus during single-leg landing. For example, if one considers a hierarchy of control lower limb motion during single-leg landing, it would be advisable to include the trunk, to maintain body stability. If the trunk moves laterally, the centre of mass will move with it, thereby resulting in a valgus position of the knee, because the ground reaction force will follow laterally to the knee joint. A prospective study investigating the influence of the trunk on dynamic knee valgus, including neuromuscular control of the trunk, might be useful in optimising risk screening and intervention programmes.

- Future research could concentrate on ankle eversion and foot pronation, as they are known to contribute to dynamic knee valgus during landing (Hewett et al., 2006). For ankle eversion to cause dynamic valgus, it can be hypothesised that abnormal movement would have to be initiated at the ankle and thus be the first joint to collapse in the kinetic chain. In this case, it would be expected that eversion sprains would accompany non-contact ACL injuries, given the superior mass of the upper body, thigh and leg segments in relation to the ankle.
- Future studies should include different athletes or injured populations. This would be helpful in revealing how they perform single-leg squatting and single-leg multi-directional landing tasks with respect to lower limb biomechanics, strength and EMG activity. It would also be useful to discover whether there are any differences between sports, as it would help identify those athletes who are at higher risk.
- More left-dominant participants should be included in future work, to balance right and left dominancy. This would help detect the effect of leg dominancy on joint angles or moments.
- Future work on the type of intervention is important to establish the effect of intervention on lower limb biomechanics. Considering that gluteal muscle factors are clearly not the only contributors to dynamic knee valgus, any factor which can influence this issue is worthy of investigation. Possible interventions include programmes that target individual factors such as hip strengthening, increasing dorsi-flexion ROM and improving balance, to establish whether they alone can improve individual landing strategies. Ultimately, this would allow for improved injury prevention strategies in those considered at high risk.

References:

Agel, J., Arendt, E. a, and Bershadsky, B. (2005). Anterior cruciate ligament injury in national collegiate athletic association basketball and soccer: a 13-year review. *The American Journal of Sports Medicine*, *33*(4), 524–530. http://doi.org/10.1177/0363546504269937

Alenezi, F., Herrington, L., Jones, P., and Jones, R. (2014). The reliability of biomechanical variables collected during single leg squat and landing tasks. *Journal of Electromyography and Kinesiology*, *24*(5), 718–721. http://doi.org/10.1016/j.jelekin.2014.07.007

Ardern, C. L., Taylor, N. F., Feller, J. a, and Webster, K. E. (2014). Fifty-five per cent return to competitive sport following anterior cruciate ligament reconstruction surgery: an updated systematic review and meta-analysis including aspects of physical functioning and contextual factors. *British Journal of Sports Medicine*, *48*(21), 1543–52. http://doi.org/10.1136/bjsports-2013-093398

Ardern, C. L., Webster, K. E., Taylor, N. F., and Feller, J. a. (2011). Return to the preinjury level of competitive sport after anterior cruciate ligament reconstruction surgery: two-thirds of patients have not returned by 12 months after surgery. *The American Journal of Sports Medicine*, *39*(3), 538–43. http://doi.org/10.1177/0363546510384798

Arokoski, M. H., Arokoski, J. P. A., Haara, M., Kankaanpää, M., Vesterinen, M., Niemitukia, L. H., and Helminen, H. J. (2002). Hip muscle strength and muscle cross sectional area in men with and without hip osteoarthritis. *The Journal of Rheumatology*, *29*(10), 2185–2195. Retrieved from http://www.jrheum.org/content/29/10/2185.full.pdf

Atkinson, G., and Nevill, A. M. (1998). Statistical methods for assessing measurement error (reliability) in variables relevant to sports medicine. *Sports medicine*, *26*(4), 217-238.

Augustsson, J., Thomeé, R., Linden, C., Folkesson, M., Tranberg, R., and Karlsson, J. (2006). Single-leg hop testing following fatiguing exercise: reliability and biomechanical analysis. *Scandinavian journal of medicine and science in sports*, *16*(2), 111-120.

Ayotte, N. W., Stetts, D. M., Keenan, G., and Greenway, E. H. (2007). Electromyographical analysis of selected lower extremity muscles during 5 unilateral weight-bearing exercises. *journal of orthopaedic and sports physical therapy*, *37*(2), 48-55.

Baker, R. (2006). Gait analysis methods in rehabilitation. *Journal of Neuroengineering and*

Rehabilitation, *3*, 4. http://doi.org/10.1186/1743-0003-3-4

Baldon, R. D. M., Lobato, D. F. M., Carvalho, L. P., Santiago, P. R. P., Benze, B. G., and Serrão, F. V. (2011). Relationship between eccentric hip torque and lower-limb kinematics: Gender differences. *Journal of Applied Biomechanics*, *27*(3), 223–232.

Barber-Westin, S. D., Noyes, F. R., & Galloway, M. (2006). Jump-land characteristics and muscle strength development in young athletes. *The American journal of sports medicine*, *34*(3), 375-384.

Barendrecht, M., Lezeman, H. C. A., Duysens, J., and Smits-Engelsman, B. C. M. (2011). Neuromuscular Training Improves Knee Kinematics, in Particular in Valgus Aligned Adolescent Team Handball Players of Both Sexes. Journal of Strength and Conditioning Research, 25(3), 575-584.

Barrios, J. A., Crossley, K. M., and Davis, I. S. (2010). Gait retraining to reduce the knee adduction moment through real-time visual feedback of dynamic knee alignment. *Journal of biomechanics*, *43*(11), 2208-2213.

Barton, C. J., Kennedy, A., Twycross-Lewis, R., Woledge, R., Malliaras, P., and Morrissey, D. (2014). Gluteal muscle activation during the isometric phase of squatting exercises with and without a Swiss ball. *Physical Therapy in Sport*, *15*(1), 39-46.

Barton, C. J., Lack, S., Malliaras, P., and Morrissey, D. (2012). Gluteal muscle activity and patellofemoral pain syndrome: a systematic review. *Br J Sports Med*, bjsports-2012.

Batterham, A. M., and George, K. P. (2003). Reliability in evidence-based clinical practice: a primer for allied health professionals☆. *Physical therapy in Sport*, *4*(3), 122-128.

Bell, D. R., Padua, D. A., and Clark, M. A. (2008). Muscle Strength and Flexibility Characteristics of People Displaying Excessive Medial Knee Displacement. *Archives of Physical Medicine and Rehabilitation*, *89*(7), 1323–1328. http://doi.org/10.1016/j.apmr.2007.11.048

Bell L, Brand A, and Pedersen, R. (1989). Prediction of hip-joint center location from external landmarks. *Hum Move Sci, 8*, 3-16.

Bendjaballah, M., Shirazi-Adl, A., and Zukor, D. J. (1997). Finite element analysis of human knee joint in varus-valgus. *Clinical biomechanics*, *12*(3), 139-148.

Benoit, D. L., Lamontagne, M., Cerulli, G., and Liti, A. (2003). The clinical significance of electromyography normalisation techniques in subjects with anterior cruciate ligament injury during treadmill walking. *Gait and posture*, *18*(2), 56-63.

Berns, G. S., Hull, M. L., and Patterson, H. A. (1992). Strain in the anteromedial bundle of the anterior cruciate ligament under combination loading. *Journal of Orthopaedic Research*, *10*(2), 167-176.

Besier, T. F., Lloyd, D. G., Cochrane, J. L. and Ackland, T. R. 2001. External loading of the knee joint during running and cutting maneuvers. Medicine and Science in Sports and Exercise 33(7), pp.1168-1175.

Beutler, A. I., de la Motte, S. J., Marshall, S. W., Padua, D. A., and Boden, B. P. (2009). Muscle strength and qualitative jump-landing differences in male and female military cadets: The jump-ACL study. *Journal of Sports Science and Medicine*, *8*(4), 663–671.

Beynnon, B. D. and Fleming, B. C. 1998. Anterior cruciate ligament strain in-vivo: a review of previous work. Journal of Biomechanics 31(6), pp.519-525.

Biodex Medical Systems, Inc. (1999). Balance system operations and service manual #945–300. Shirley, NY: Author.

Blackburn, J. T., and Padua, D. A. (2008). Influence of trunk flexion on hip and knee joint kinematics during a controlled drop landing. *Clinical Biomechanics*, *23*(3), 313-319.

Blackburn, J. T., and Padua, D. A. (2009). Sagittal-plane trunk position, landing forces, and quadriceps electromyographic activity. *Journal of athletic training*, *44*(2), 174-179.

Boden BP, Torg JS, Knowles SB, H. TE. (2009). Video Analysis of Anterior Cruciate Ligament Injury. *American Journal of SPorts Medicine*, *37*, 252–259. http://doi.org/10.1177/0363546508328107

Boden, B. P., Dean, G. S., Feagin, J. A., and Garrett, W. E. (2000). Mechanisms of anterior cruciate ligament injury. *Orthopedics*, *23*(6), 573-578.

Boden, B. P., Griffin, L. Y., and Garrett Jr, W. E. (2000). Etiology and prevention of noncontact ACL injury. *The Physician and sportsmedicine*, *28*(4), 53-60.

Bolgla, L. A., and Uhl, T. L. (2005). Electromyographic analysis of hip rehabilitation exercises in a group of healthy subjects. *Journal of Orthopaedic and Sports Physical Therapy*, *35*(8), 487-494.

Bolgla, L. A., Malone, T. R., Umberger, B. R., and Uhl, T. L. (2010). Reliability of electromyographic methods used for assessing hip and knee neuromuscular activity in females diagnosed with patellofemoral pain syndrome. *Journal of Electromyography and Kinesiology*, *20*(1), 142-147.

Bolgla, L. A., Malone, T. R., Umberger, B. R., and Uhl, T. L. (2011). Comparison of hip and knee strength and neuromuscular activity in subjects with and without patellofemoral pain syndrome. *International journal of sports physical therapy*, *6*(4), 285.

Boling, M. C., Padua, D. A., and Creighton, R. A. (2009). Concentric and eccentric torque of the hip musculature in individuals with and without patellofemoral pain. *Journal of Athletic Training*, *44*(1), 7–13. http://doi.org/10.4085/1062-6050-44.1.7

Borsa, P. A., Lephart, S. M., Irrgang, J. J., Safran, M. R., and Fu, F. H. (1997). The effects of joint position and direction of joint motion on proprioceptive sensibility in anterior cruciate ligament-deficient athletes. *The American journal of sports medicine*, *25*(3), 336-340.

Boudreau, Samantha N, Dwyer, M. K., Mattacola, C. G., Lattermann, C., Uhl, T. L., and McKeon, J. (2009). Hip-Muscle Activation during the Lunge, Single-Leg Squat, and Step-Up-and-Over Exercises. *Journal of Sport Rehabilitation*, *18*(1), 91–103. Retrieved from http://journals.humankinetics.com/doi/abs/10.1123/jsr.18.1.91

Bouillon, L. E., Wilhelm, J., Eisel, P., Wiesner, J., Rachow, M., and Hatteberg, L. (2012). Electromyographic assessment of muscle activity between genders during unilateral weight-bearing tasks using adjusted distances. *International Journal of Sports Physical Therapy*, *7*(6), 595–605. Retrieved from http://www.pubmedcentral.nih.gov/articlerender.fcgi?artid=3537460andtool=pmcentrezandrendertype=abstract

Brophy, R., Silvers, H. J., Gonzales, T., and Mandelbaum, B. R. (2010). Gender influences: the role of leg dominance in ACL injury among soccer players. *British journal of sports medicine*, bjsports51243.

Brown, T. N., Palmieri-Smith, R. M., & McLean, S. G. (2009). Sex and limb differences in hip and knee kinematics and kinetics during anticipated and unanticipated jump landings: implications for anterior cruciate ligament injury. *British journal of sports medicine*, *43*(13), 1049-1056.

Brown, T. N., McLean, S. G., and Palmieri-Smith, R. M. (2014). Associations between lower limb muscle activation strategies and resultant multi-planar knee kinetics during single leg landings. *Journal of Science and Medicine in Sport / Sports Medicine Australia*, *17*(4), 408–413.

Buoncristiani, A. M., Tjoumakaris, F. P., Starman, J. S., Ferretti, M. and Fu, F. H. 2006. Anatomic double-bundle anterior cruciate ligament reconstruction. Arthroscopy: The

Journal of Arthroscopic and Related Surgery 22(9), pp.1000-1006.

Butler, D. L., Noyes, F. R., and Grood, E. S. (1980). Ligamentous restraints to anterior-posterior drawer in the human knee. *J Bone Joint Surg Am*, *62*(2), 259-270.

Butler, D., Oster, D., Feder, S., Grood, E. and Noyes, F. (1991). Effects of gamma irradiation on the biomechanics of patellar tendon allografts of the ACL in the goat. *Trans Orthop Res Soc,* 16**,** 205.

Byrne, D., Mulhall, K., and Baker, J. (2010). Anatomy and Biomechanics of the Hip. The Open Sports Medicine Journal, 4(1), 51-57. http://dx.doi.org/10.2174/1874387001004010051

Cahalan, T. D., Johnson, M. E., Liu, S., and Chao, E. Y. S. (1989). Quantitative measurements of hip strength in different age groups. *Clinical orthopaedics and related research*, *246*, 136-145.

Cappozzo, A., Catani, F., Leardini, A., Benedetti, M. G., and Della Croce, U. (1996). Position and orientation in space of bones during movement: experimental artefacts. *Clinical biomechanics*, *11*(2), 90-100.

Cashman, G. E. (2012). The effect of weak hip abductors or external rotators on knee valgus kinematics in healthy subjects: a systematic review. *Journal of Sport Rehabilitation*, *21*(3), 273–84. Retrieved from http://www.ncbi.nlm.nih.gov/pubmed/22894982

Chandrashekar, N., Slauterbeck, J., and Hashemi, J. (2005). Sex-based differences in the anthropometric characteristics of the anterior cruciate ligament and its relation to intercondylar notch geometry. *The American journal of sports medicine*, *33*(10), 1492-1498.

Chappell, J. D., and Limpisvasti, O. (2008). Effect of a neuromuscular training program on the kinetics and kinematics of jumping tasks. *The American Journal of Sports Medicine*, *36*(6), 1081–6. http://doi.org/10.1177/0363546508314425

Chappell, J. D., Herman, D. C., Knight, B. S., Kirkendall, D. T., Garrett, W. E., and Yu, B. (2005). Effect of fatigue on knee kinetics and kinematics in stop-jump tasks. *The American journal of sports medicine*, *33*(7), 1022-1029.

Chappell, J. D., Yu, B., Kirkendall, D. T., and Garrett, W. E. (2002). A comparison of knee kinetics between male and female recreational athletes in stop-jump tasks. *The American Journal of Sports Medicine*, *30*(2), 261–267. http://doi.org/10.1177/03635465020300021901

Chomiak, J., Junge, A., Peterson, L., and Dvorak, J. (2000). Severe injuries in football players. *The American journal of sports medicine*, *28*(5_suppl), 58-68.

Claiborne, T. L., Armstrong, C. W., Gandhi, V., and Pincivero, D. M. (2006). Relationship between hip and knee strength and knee valgus during a single leg squat. *Journal of Applied Biomechanics*, *22*(1), 41–50. http://doi.org/10.1097/00005768-200305001-01931

Claiborne, T. L., Timmons, M. K., and Pincivero, D. M. (2009). Test–retest reliability of cardinal plane isokinetic hip torque and EMG. *Journal of Electromyography and Kinesiology*, *19*(5), e345–e352. Retrieved from http://ac.els-cdn.com/S1050641108001168/1-s2.0-S1050641108001168-main.pdf?_tid=855628b6-dae2-11e5-8cd7-00000aacb35dandacdnat=1456310394_5da915ec1c3da2fe75d06b2b48c2efe7

Coppieters, M, Stappaerts, K, Janssens, K, and Jull, G. (2002). Reliability of detecting 'onset of pain' and 'submaximal pain' during neural provocation testing of the upper quadrant. *Physiotherapy Research International, 7*(3), 146-156.

Criswell, E. (2011). *Cram's Introduction to Surface Electromyography*. *Training*. http://doi.org/10.1002/car.1158

Csintalan, R. P., Inacio, M. C. S., and Funahashi, T. T. (2008). Incidence rate of anterior cruciate ligament reconstructions. *The Permanente Journal*, *12*(3), 17–21. Retrieved from http://www.pubmedcentral.nih.gov/articlerender.fcgi?artid=3037119andtool=pmcentrezandrendertype=abstract

Culvenor, a. G., Cook, J. L., Collins, N. J., and Crossley, K. M. (2012). Is patellofemoral joint osteoarthritis an under-recognised outcome of anterior cruciate ligament reconstruction? A narrative literature review. *British Journal of Sports Medicine*, *47*(3), 66–73. http://doi.org/10.1136/bjsports-2012-091490

Dargel, J., Gotter, M., Mader, K., Pennig, D., Koebke, J., and Schmidt-Wiethoff, R. (2007). Biomechanics of the anterior cruciate ligament and implications for surgical reconstruction. *Strategies in Trauma and Limb Reconstruction*, *2*(1), 1-12.

DeAngelis, J. P., and Spindler, K. P. (2010). Traumatic Bone Bruises in the Athlete's Knee. *Sports Health: A Multidisciplinary Approach*. http://doi.org/10.1177/1941738110377745

Decker, M. J., Torry, M. R., Wyland, D. J., Sterett, W. I., and Steadman, J. R. (2003). Gender differences in lower extremity kinematics, kinetics and energy absorption during landing. *Clinical biomechanics*, *18*(7), 662-669.

DeMorat, G., Weinhold, P., Blackburn, T., Chudik, S., & Garrett, W. (2004). Aggressive quadriceps loading can induce noncontact anterior cruciate ligament injury. *The American journal of sports medicine*, *32*(2), 477-483.

Denegar, C. R., and Ball, D. W. (1993). Assessing reliability and precision of measurement: an introduction to intraclass correlation and standard error of measurement. *Journal of sport rehabilitation*, *2*(1), 35-42.

Devita, P., and Skelly, W. a. (1992). Effect of landing stiffness on joint kinetics and energetics in the lower extremity. *Medicine and Science in Sports and Exercise*, *24*(1), 108–115. http://doi.org/10.1249/00005768-199201000-00018

DiCesare, C. A., Bates, N. A., Barber Foss, K. D., Thomas, S. M., Wordeman, S. C., Sugimoto, D., ... and Ford, K. R. (2015). Reliability of 3-Dimensional Measures of Single-Leg Cross Drop Landing Across 3 Different Institutions: Implications for Multicenter Biomechanical and Epidemiological Research on ACL Injury Prevention. *Orthopaedic journal of sports medicine*, *3*(12), 2325967115617905.

Dienst, M., Burks, R. T. and Gries, P. E. (2002). Anatomy and biomechanics of the anterior cruciate ligament. *Orthopedic Clinics of North America,* 33**,** 605-620.

Dierks, T. a., Manal, K. T., Hamill, J., and Davis, I. S. (2008). Proximal and Distal Influences on Hip and Knee Kinematics in Runners With Patellofemoral Pain Syndrome During a Prolonged Run. *Journal of Orthopaedic and Sports Physical Therapy*, *38*(8), 448–456. http://doi.org/10.2519/jospt.2008.2490

DiMattia, M. A., Livengood, A. L., Uhl, T. L., Mattacola, C. G., and Malone, T. R. (2005). What Are the Validity of the Single-Leg-Squat Test and Its Relationship to Hip-Abduction Strength? *Journal of Sport Rehabilitation*, *14*(2), 108–123. http://doi.org/10.1123/jsr.14.2.108

Distefano, L. J., Blackburn, J. T., Marshall, S. W., and Padua, D. A. (2009). Gluteal muscle activation during common therapeutic exercises. *journal of orthopaedic and sports physical therapy*, *39*(7), 532-540.

Domnick, C., Raschke, M. J., and Herbort, M. (2016). Biomechanics of the anterior cruciate ligament: Physiology, rupture and reconstruction techniques. *World Journal of Orthopedics*, *7*(2), 82. http://doi.org/10.5312/wjo.v7.i2.82

Downs, S. H., and Black, N. (1998). The feasibility of creating a checklist for the assessment of the methodological quality both of randomised and non-randomised studies of health care interventions. *Journal of Epidemiology and Community Health*, *52*(6), 377-384.

Dwyer, M. K., Boudreau, S. N., Mattacola, C. G., Uhl, T. L., and Lattermann, C. (2010). Comparison of lower extremity kinematics and hip muscle activation during rehabilitation tasks between sexes. *Journal of Athletic Training*, *45*(2), 181–190. http://doi.org/10.4085/1062-6050-45.2.181

Edwards, S., Steele, J. R., and McGhee, D. E. (2010). Does a drop landing represent a whole skill landing and is this moderated by fatigue?. *Scandinavian journal of medicine and science in sports*, *20*(3), 516-523.

Edwards, S., Steele, J. R., Cook, J. L., Purdam, C. R., and McGhee, D. E. (2012). Lower limb movement symmetry cannot be assumed when investigating the stop-jump landing. *Medicine and Science in Sports and Exercise*, *44*(6), 1123–1130. http://doi.org/10.1249/MSS.0b013e31824299c3

Ekegren, C. L., Miller, W. C., Celebrini, R. G., Eng, J. J., and Macintyre, D. L. (2009). Reliability and validity of observational risk screening in evaluating dynamic knee valgus. *The Journal of Orthopaedic and Sports Physical Therapy*, *39*(9), 665–74. http://doi.org/10.2519/jospt.2009.3004

Ekstrand, J., Gillquist, J., and Liljedahl, S. O. (1983). Prevention of soccer injuries: supervision by doctor and physiotherapist. *The American Journal of Sports Medicine*, *11*(3), 116-120.

Eng, J. J., Kim, C. M., and MacIntyre, D. L. (2002). Reliability of lower extremity strength measures in persons with chronic stroke. *Archives of Physical Medicine and Rehabilitation*, *83*(3), 322–328. http://doi.org/10.1053/apmr.2002.29622

Engelen-van Melick, N., van Cingel, R. E., van Tienen, T. G., and Nijhuis-van der Sanden, M. W. (2015). Functional performance 2–9 years after ACL reconstruction: cross-sectional comparison between athletes with bone–patellar tendon–bone, semitendinosus/gracilis and healthy controls. *Knee Surgery, Sports Traumatology, Arthroscopy*, 1-12.

Enoka, R. M., and Stuart, D. G. (1992). Neurobiology of muscle fatigue. *Journal of applied physiology*, *72*(5), 1631-1648.

Faude, O., Junge, a, Kindermann, W., and Dvorak, J. (2006). Risk factors for injuries in elite female soccer players. *British Journal of Sports Medicine*, *40*(9), 785–790. http://doi.org/10.1136/bjsm.2006.027540

Ferber, R., Davis, I. M., and Williams Iii, D. S. (2003). Gender differences in lower extremity mechanics during running. *Clinical biomechanics*, *18*(4), 350-357.

Ford, K. R., DiCesare, C. A., Myer, G. D., and Hewett, T. E. (2014). Real-time biofeedback to target risk of anterior cruciate ligament injury: a technical report for injury prevention and rehabilitation. *J Sport Rehabil*, 2013-0138.

Ford, K. R., McLean, S. G., Hewett, T. E., Myer, G. D., Ford, K. R., Heidt, R. S., … Succop, P. (2005). Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes a prospective study. *The American Journal of Sports Medicine*, *33*(4), 492–501. http://doi.org/10.1177/0363546504269591

Ford, K. R., Myer, G. D., and Hewett, T. E. (2003). Valgus knee motion during landing in high school female and male basketball players. *Medicine and Science in Sports and Exercise*, *35*(10), 1745–1750. http://doi.org/10.1249/01.MSS.0000089346.85744.D9

Ford, K. R., Myer, G. D., and Hewett, T. E. (2007). Reliability of landing 3D motion analysis: implications for longitudinal analyses. *Medicine and Science in Sports and Exercise*, *39*(11), 2021–2028. Retrieved from http://graphics.tx.ovid.com/ovftpdfs/FPDDNCGCHFLEDO00/fs047/ovft/live/gv031/00005768/00005768-200711000-00018.pdf

Ford, K. R., Myer, G. D., Smith, R. L., Vianello, R. M., Seiwert, S. L., and Hewett, T. E. (2006). A comparison of dynamic coronal plane excursion between matched male and female athletes when performing single leg landings. *Clin Biomech (Bristol, Avon), 21*(1), 33-40. doi: 10.1016/j.clinbiomech.2005.08.010

Ford, K. R., Shapiro, R., Myer, G. D., Van Den Bogert, A. J., and Hewett, T. E. (2010). Longitudinal sex differences during landing in knee abduction in young athletes. *Med Sci Sports Exerc, 42*(10), 1923-1931. doi: 10.1249/MSS.0b013e3181dc99b1

Frigo C., Crenna P. (2009). Multichannel SEMG in clinical gait analysis: a review and state-of-the-art. Clin. Biomech. 24, 236–245.

Fukuda, Y., Woo, S. L. Y., Loh, J. C., Tsuda, E., Tang, P., McMahon, P. J., and Debski, R. E. (2003). A quantitative analysis of valgus torque on the ACL: A human cadaveric study. *Journal of Orthopaedic Research*, *21*(6), 1107–1112.

http://doi.org/10.1016/S0736-0266(03)00084-6

Fung, D. T., Hendrix, R. W., Koh, J. L., and Zhang, L. Q. (2007). ACL impingement prediction based on MRI scans of individual knees. *Clin Orthop Relat Res, 460*, 210-218. doi: 10.1097/BLO.0b013e31804d2339

Gabriel, M. T., Wong, E. K., Woo, S. L. Y., Yagi, M., & Debski, R. E. (2004). Distribution of in situ forces in the anterior cruciate ligament in response to rotatory loads. *Journal of Orthopaedic Research*, *22*(1), 85-89.

Gage, B. E., Mcilvain, N. M., Collins, C. L., Fields, S. K., and Comstock, R. D. (2012). Presenting to United States Emergency Departments From 1999 Through 2008, 378–385. http://doi.org/10.1111/j.1553-2712.2012.01315.x

Gagnon, D., Nadeau, S., Gravel, D., Robert, J., Bélanger, D., and Hilsenrath, M. (2005). Reliability and validity of static knee strength measurements obtained with a chair-fixed dynamometer in subjects with hip or knee arthroplasty. *Archives of physical medicine and rehabilitation*, *86*(10), 1998-2008.

Gopalakrishnan, S., and Ganeshkumar, P. (2013). Systematic Reviews and Meta-analysis: Understanding the Best Evidence in Primary Healthcare. *Journal of Family Medicine and Primary Care*, *2*(1), 9–14. http://doi.org/10.4103/2249

Graci, V., Van Dillen, L. R., and Salsich, G. B. (2012). Gender differences in trunk, pelvis and lower limb kinematics during a single leg squat. *Gait and Posture*, *36*(3), 461–466. http://doi.org/10.1016/j.gaitpost.2012.04.006

Griffin, L. (2006). Understanding and Preventing Noncontact Anterior Cruciate Ligament Injuries. *The American Journal of Sports Medicine*, *34*(9), 1512–1532. http://doi.org/10.1177/0363546506286866

Gross, K. D., Niu, J., Zhang, Y. Q., Felson, D. T., McLennan, C., Hannan, M. T., ... and Hunter, D. J. (2007). Varus foot alignment and hip conditions in older adults. *Arthritis and Rheumatology*, *56*(9), 2993-2998.

Gwinn, D. E., Wilckens, J. H., McDevitt, E. R., Ross, G., and Kao, T. C. (2000). The relative incidence of anterior cruciate ligament injury in men and women at the United States Naval Academy. *American Journal of Sports Medicine*, *28*(1), 98–102.

Hamilton, R. T., Shultz, S. J., Schmitz, R. J., and Perrin, D. H. (2008). Triple-hop distance as a valid predictor of lower limb strength and power. *Journal of athletic training*, *43*(2), 144-151.

Hanson, A. M., Padua, D. A., Blackburn, J. T., Prentice, W. E., and Hirth, C. J. (2008). Muscle activation during side-step cutting maneuvers in male and female soccer athletes. *Journal of Athletic Training*, *43*(2), 133–143. http://doi.org/10.4085/1062-6050-43.2.133

Harner, C. D., Paulos, L. E., Greenwald, A. E., Rosenberg, T. D., and Cooley, V. C. (1994). Detailed analysis of patients with bilateral anterior cruciate ligament injuries. *The American Journal of Sports Medicine*, *22*(1), 37–43. http://doi.org/10.1177/036354659402200107

Hart, H. F., Culvenor, A. G., Collins, N. J., Ackland, D. C., Cowan, S. M., Machotka, Z., and Crossley, K. M. (2016). Knee kinematics and joint moments during gait following anterior cruciate ligament reconstruction: a systematic review and meta-analysis. *British Journal of Sports Medicine*, *50*(10), 597–612. http://doi.org/10.1136/bjsports-2015-094797

Hart, J. M., Craig Garrison, J., Casey Kerrigan, D., Palmieri-Smith, R., and Ingersoll, C. D. (2007). Gender differences in gluteus medius muscle activity exist in soccer players performing a forward jump. *Research in sports medicine*, *15*(2), 147-155.

Heidt, R. S., Sweeterman, L. M., Carlonas, R. L., Traub, J. A., and Tekulve, F. X. (2000). Avoidance of soccer injuries with preseason conditioning. *American Journal of Sports Medicine, 28*(5), 659-662.

Herman, D. C., Oñate, J. A., Weinhold, P. S., Guskiewicz, K. M., Garrett, W. E., Yu, B., and Padua, D. A. (2009). The effects of feedback with and without strength training on lower extremity biomechanics. *The American journal of sports medicine*, *37*(7), 1301-1308.

Herrington, L. (2010). The effects of 4 weeks of jump training on landing knee valgus and crossover hop performance in female basketball players. *Journal of Strength and Conditioning Research / National Strength and Conditioning Association*, *24*(12), 3427–3432. http://doi.org/10.1519/JSC.0b013e3181c1fcd8

Herrington, L. (2011). Knee valgus angle during landing tasks in female volleyball and basketball players. *J Strength Cond Res*, *25*(1), 262–266. http://doi.org/10.1519/JSC.0b013e3181b62c77

Herrington, L. (2013). Effect of a SERF strap on pain and knee-valgus angle during unilateral squat and step landing in patellofemoral patients. *J Sport Rehabil*, *22*(1), 27–32.

Herrington, L. and Munro, A. 2014. A preliminary investigation to establish the criterion validity of a qualitative scoring system of limb alignment during single-leg squat and landing. *J Exerc Sports Orthop,* 1**,** 1-6.

Herrington, L., and Munro, A. (2010). Drop jump landing knee valgus angle; normative data in a physically active population. *Physical Therapy in Sport*, *11*(2), 56–59. http://doi.org/10.1016/j.ptsp.2009.11.004

Hewett, T. E., and Myer, G. D. (2011). The Mechanistic Connection between the Trunk, Knee, and ACL Injury. *Exercise and Sport Sciences Reviews*, 1. http://doi.org/10.1097/JES.0b013e3182297439

Hewett, T. E., Ford, K. R., Hoogenboom, B. J., and Myer, G. D. (2010). Understanding and preventing acl injuries: current biomechanical and epidemiologic considerations - update 2010. *North American Journal of Sports Physical Therapy : NAJSPT*, *5*(4), 234–251. Retrieved from http://www.ncbi.nlm.nih.gov/pubmed/21655382%0Ahttp://www.pubmedcentral.nih.gov/articlerender.fcgi?artid=PMC3096145

Hewett, T. E., Lindenfeld, T. N., Riccobene, J. V., and Noyes, F. R. (1999). The effect of neuromuscular training on the incidence of knee injury in female athletes. A prospective study. *Am J Sports Med, 27*(6), 699-706.

Hewett, T. E., Myer, G. D. and Ford, K. R. (2006). Anterior cruciate ligament injuries in female athletes part 1, mechanisms and risk factors. *The American journal of sports medicine,* 34**,** 299-311.

Hewett, T. E., Myer, G. D., and Ford, K. R. (2004). Decrease in neuromuscular control about the knee with maturation in female athletes. *The Journal of Bone and Joint Surgery. American Volume*, *86–A*(September), 1601–1608. http://doi.org/86/8/1601 [pii]

Hewett, T. E., Myer, G. D., Ford, K. R., Heidt, R. S., Colosimo, A. J., McLean, S. G., … Succop, P. (2005). Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes: a prospective study. *The American Journal of Sports Medicine*, *33*(4), 492–501. http://doi.org/10.1177/0363546504269591

Hewett, T. E., Paterno, M. V, and Myer, G. D. (2002). Strategies for enhancing proprioception and neuromuscular control of the knee. *Clinical Orthopaedics and Related Research*, (402), 76–94. http://doi.org/10.1097/01.blo.0000026962.51742.99

Hewett, T. E., Torg, J. S., and Boden, B. P. (2009). Video analysis of trunk and knee motion during non-contact anterior cruciate ligament injury in female athletes: lateral trunk and knee abduction motion are combined components of the injury mechanism. *British Journal of Sports Medicine*, *43*(6), 417–22. http://doi.org/10.1136/bjsm.2009.059162

Hollman, J. H., Galardi, C. M., Lin, I. H., Voth, B. C., and Whitmarsh, C. L. (2014). Frontal and transverse plane hip kinematics and gluteus maximus recruitment correlate with frontal plane knee kinematics during single-leg squat tests in women. *Clinical Biomechanics*, *29*(4), 468–474. http://doi.org/10.1016/j.clinbiomech.2013.12.017

Hollman, J. H., Ginos, B. E., Kozuchowski, J., Vaughn, A. S., Krause, D. A., and Youdas, J. W. (2009). Relationships between knee valgus, hip-muscle strength, and hip-muscle recruitment during a single-limb step-down. *Journal of Sport Rehabilitation*, *18*(1), 104.

Hollman, J. H., Hohl, J. M., Kraft, J. L., Strauss, J. D., and Traver, K. J. (2013). Modulation of frontal-plane knee kinematics by hip-extensor strength and gluteus maximus recruitment during a jump-landing task in healthy women. *Journal of Sport Rehabilitation*, *22*(3), 184–190.

Homan, K. J., Norcross, M. F., Goerger, B. M., Prentice, W. E., and Blackburn, J. T. (2013). The influence of hip strength on gluteal activity and lower extremity kinematics. *Journal of Electromyography and Kinesiology*, *23*(2), 411–415. Retrieved from http://ac.els-cdn.com/S105064111200212X/1-s2.0-S105064111200212X-main.pdf?_tid=87aa6330-dae1-11e5-bf9f-00000aab0f26andacdnat=1456309968_5ba4f9126b87b77fc9364828bdb4d9b3

Hootman, J. M., Dick, R., and Agel, J. (2007). Epidemiology of collegiate injuries for 15 sports: summary and recommendations for injury prevention initiatives. *Journal of Athletic Training*, *42*(2), 311. Retrieved from http://www.ncbi.nlm.nih.gov/pmc/articles/PMC1941297/pdf/i1062-6050-42-2-311.pdf

Hopkins, W. G. (2000). Measures of reliability in sports medicine and science. *Sports medicine,* 30**,** 1-15.

Hopkins, W. G., Marshall, S. W., Batterham, A. M., and Hanin, J. (2009). Progressive statistics for studies in sports medicine and exercise science. *Med Sci Sports Exerc, 41*(1), 3-13. doi: 10.1249/MSS.0b013e31818cb278

Ireland, M. L. (1999). Anterior cruciate ligament injury in female athletes: epidemiology.

Journal of Athletic Training, *34*(2), 150–154. Retrieved from http://eutils.ncbi.nlm.nih.gov/entrez/eutils/elink.fcgi?dbfrom=pubmedandid=16558558andretmode=refandcmd=prlinks%5Cnpapers3://publication/uuid/2DA6A832-473F-457C-952B-F298459BF4BA

Irmischer, B. S., Harris, C., Pfeiffer, R. P., DeBeliso, M. A., Adams, K. J., and Shea, K. G. (2004). Effects of a knee ligament injury prevention exercise program on impact forces in women. *The Journal of Strength and Conditioning Research*, *18*(4), 703-707.

Itoh, H., Takiguchi, K., Shibata, Y., Okubo, S., Yoshiya, S., and Kuroda, R. (2016). Correlation between hip function and knee kinematics evaluated by three-dimensional motion analysis during lateral and medial side-hopping, 2461–2467. http://doi.org/10.1589/jpts.28.2461

Jacobs, C. A., Uhl, T. L., Mattacola, C. G., Shapiro, R., and Rayens, W. S. (2007). Hip abductor function and lower extremity landing kinematics: sex differences. *Journal of Athletic Training*, *42*(1), 76.

Jacobs, C., and Mattacola, C. (2005). Sex differences in eccentric hip-abductor strength and knee-joint kinematics when landing from a jump. *Journal of Sport Rehabilitation*, *14*(4), 346.

Järvinen, M., Natri, A., Laurila, S., & Kannus, P. (1994). Mechanisms of anterior cruciate ligament ruptures in skiing. *Knee Surgery, Sports Traumatology, Arthroscopy*, *2*(4), 224-228.

Julia, M., Dupeyron, A., Laffont, I., Parisaux, J.-M., Lemoine, F., Bousquet, P.-J., and Hérisson, C. (2010). Reproducibility of isokinetic peak torque assessments of the hip flexor and extensor muscles. *Annals of Physical and Rehabilitation Medicine*, *53*(5), 293–305. Retrieved from http://ac.els-cdn.com/S1877065710001089/1-s2.0-S1877065710001089-main.pdf?_tid=91d45414-dae2-11e5-bd8f-00000aacb361andacdnat=1456310414_92c2ba7cd44424045270669140e63f83

Kadaba, M. P., Ramakrishnan, H. K., Wooten, M. E., Gainey, J., Gorton, G., and Cochran, G. V. B. (1989). Repeatability of Kinematic, Kinetic, and EMG activity in Normal Adult Gait.pdf. *Journal of Orthopaedic Research*, *7*(6), 849–860. http://doi.org/10.1002/jor.1100070611

Kanamori, A., Zeminski, J., Rudy, T. W., Li, G., Fu, F. H., & Woo, S. L. Y. (2002). The effect of axial tibial torque on the function of the anterior cruciate ligament: a biomechanical study of a simulated pivot shift test. *Arthroscopy: The Journal of Arthroscopic & Related Surgery*, *18*(4), 394-398.

Kernozek, T. W., Torry, M. R., and Iwasaki, M. (2007). Gender Differences in Lower Extremity Landing Mechanics Caused by Neuromuscular Fatigue. *The American Journal of Sports Medicine*, *36*(3), 554–565. http://doi.org/10.1177/0363546507308934

Kiapour, A. M., Quatman, C. E., Goel, V. K., Wordeman, S. C., Hewett, T. E., & Demetropoulos, C. K. (2014). Timing sequence of multi-planar knee kinematics revealed by physiologic cadaveric simulation of landing: implications for ACL injury mechanism. *Clinical Biomechanics*, *29*(1), 75-82.

Kiriyama, S., Sato, H., and Takahira, N. (2009). Gender differences in rotation of the shank during single-legged drop landing and its relation to rotational muscle strength of the knee. *The American Journal of Sports Medicine*, *37*(1), 168–174. http://doi.org/10.1177/0363546508324692

Koga, H., Nakamae, A., Shima, Y., Iwasa, J., Myklebust, G., Engebretsen, L., ... & Krosshaug, T. (2010). Mechanisms for noncontact anterior cruciate ligament injuries: knee joint kinematics in 10 injury situations from female team handball and basketball. *The American journal of sports medicine*, *38*(11), 2218-2225.

Kweon, C., Lederman, E. S., and Chhabra, A. (2013). Anatomy and biomechanics of the cruciate ligaments and their surgical implications. In *The multiple ligament injured knee* (pp. 17-27). Springer New York.

Lanshammar, K., and 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*(2), 76–79. http://doi.org/10.1016/j.ptsp.2010.10.004

Lawrence, R. K., Kernozek, T. W., Miller, E. J., Torry, M. R., Reuteman, P., Lawrence 3rd, R. K., … Reuteman, P. (2008). Influences of hip external rotation strength on knee mechanics during single-leg drop landings in females. *Clinical Biomechanics*, *23*(6), 806–813. Retrieved from http://ac.els-cdn.com/S0268003308000533/1-s2.0-S0268003308000533-main.pdf?_tid=965dd3c8-dae9-11e5-af23-00000aab0f26andacdnat=1456313429_352e4513fa2982176e49f069b51bfae7

LaRoche, D. P., Lussier, M. V., and Roy, S. J. (2008). Chronic stretching and voluntary muscle force. *The Journal of Strength and Conditioning Research*, *22*(2), 589-596.

Leetun, D. T., Ireland, M. L., Willson, J. D., Ballantyne, B. T., and Davis, I. M. (2004). Core stability measures as risk factors for lower extremity injury in athletes. *Medicine and Science in Sports and Exercise*, *36*(6), 926–934. http://doi.org/10.1249/01.MSS.0000128145.75199.C3

Lephart, S. M., Abt, J. P., Ferris, C. M., Sell, T. C., Nagai, T., Myers, J. B., and Irrgang, J. J. (2005). Neuromuscular and biomechanical characteristic changes in high school athletes: a plyometric versus basic resistance program. *British journal of sports medicine*, *39*(12), 932-938.

Levine, J. W., Kiapour, A. M., Quatman, C. E., Wordeman, S. C., Goel, V. K., Hewett, T. E., & Demetropoulos, C. K. (2013). Clinically relevant injury patterns after an anterior cruciate ligament injury provide insight into injury mechanisms. *The American journal of sports medicine*, *41*(2), 385-395.

Li, G., Rudy, T. W., Sakane, M., Kanamori, A., Ma, C. B., and Woo, S. L. Y. (1999). The importance of quadriceps and hamstring muscle loading on knee kinematics land in-situ forces in the ACL. *Journal of Biomechanics, 32*(4), 395-400.

Liberati, A., Altman, D. G., Tetzlaff, J., Mulrow, C., Gøtzsche, P. C., Ioannidis, J. P. A., . . . Moher, D. (2009). The PRISMA Statement for Reporting Systematic Reviews and Meta-Analyses of Studies That Evaluate Health Care Interventions: Explanation and Elaboration. *PLOS Medicine, 6*(7), e1000100. doi:10.1371/journal.pmed.1000100

Lloyd, D. G., Buchanan, T. S., and Besier, T. F. (2005). Neuromuscular biomechanical modeling to understand knee ligament loading. In *Medicine and Science in Sports and Exercise* (Vol. 37, pp. 1939–1947). http://doi.org/10.1249/01.mss.0000176676.49584.ba

Lohmander, L. S., Ostenberg, a, Englund, M., and Roos, H. (2004). High prevalence of knee osteoarthritis, pain, and functional limitations in female soccer players twelve years after anterior cruciate ligament injury. *Arthritis and Rheumatism*, *50*(10), 3145–52. http://doi.org/10.1002/art.20589

Lorenz, D., and Reiman, M. (2011). The role and implementation of eccentric training in athletic rehabilitation: tendinopathy, hamstring strains, and acl reconstruction. *International journal of sports physical therapy*, *6*(1), 27.

Lund, H., Søndergaard, K., Zachariassen, T., Christensen, R., Bülow, P., Henriksen, M., ... and Bliddal, H. (2005). Learning effect of isokinetic measurements in healthy subjects, and reliability and comparability of Biodex and Lido dynamometers. *Clinical physiology and functional imaging*, *25*(2), 75-82.

Macauley, D. (2007). Oxford handbook of sport and exercise medicine.'. *Journal of sports science and medicine,* 6**,** 151.

Maffiuletti, N. A., Bizzini, M., Desbrosses, K., Babault, N., & Munzinger, U. 2007. Reliability of knee extension and flexion measurements using the Con-Trex isokinetic dynamometer. Clinical physiology and functional imaging, 27(6), pp.346-353.

Malloy, P., Morgan, A., Meinerz, C., Geiser, C. F., Atc, M. P., and Kipp, K. (2017). Hip external rotator strength is associated with better dynamic control of the lower extremity during landing tasks, *30*(1), 282–291. http://doi.org/10.1519/JSC.0000000000001069

Manal, K., McClay, I., Stanhope, S., Richards, J., and Galinat, B. (2000). Comparison of surface mounted markers and attachment methods in estimating tibial rotations during walking: an in vivo study. *Gait and posture*, *11*(1), 38-45.

Marieb, E. N. (2004). The endocrine system. *Human Anatomy and Physiology*, 603-643.

Markolf, K. L., Burchfield, D. M., Shapiro, M. M., Shepard, M. F., Finerman, G. A., and Slauterbeck, J. L. (1995). Combined knee loading states that generate high anterior cruciate ligament forces. *Journal of Orthopaedic Research*, *13*(6), 930-935.

Martin, H. J., Yule, V., Syddall, H. E., Dennison, E. M., Cooper, C., and Aihie Sayer, A. (2006). Is hand-held dynamometry useful for the measurement of quadriceps strength in older people? A comparison with the gold standard biodex dynamometry. *Gerontology*, *52*(3), 154–159. http://doi.org/10.1159/000091824

Mather III, R. C., Koenig, L., Kocher, M. S., Dall, T. M., Gallo, P., Scott, D. J., ... and MOON Knee Group. (2013). Societal and economic impact of anterior cruciate ligament tears. *The Journal of bone and joint surgery. American volume*, *95*(19), 1751.

Maynard, V., Bakheit, A. M. O., Oldham, J., & Freeman, J. (2003). Intra-rater and inter-rater reliability of gait measurements with CODA mpx30 motion analysis system. *Gait & posture*, *17*(1), 59-67.

McElveen, M. T., Riemann, B. L., & Davies, G. J. (2010). Bilateral comparison of propulsion mechanics during single-leg vertical jumping. *The Journal of Strength & Conditioning Research*, *24*(2), 375-381.

McGinley, J. L., Baker, R., Wolfe, R., and Morris, M. E. (2009). The reliability of three-dimensional kinematic gait measurements: a systematic review. *Gait and posture*, *29*(3), 360-369.

McLean, S. G., Lipfert, S. W., and van den Bogert, A. J. (2004). Effect of gender and defensive opponent on the biomechanics of sidestep cutting. *Medicine and Science in Sports and Exercise*, *36*(6), 1008–1016.

McNitt-Gray, J. L. (1993). Kinetics of the lower extremities during drop landings from three heights. *Journal of biomechanics*, *26*(9), 1037-1046.

Merletti, R., and Parker, P. A. (Eds.). (2004). *Electromyography: physiology, engineering, and non-invasive applications* (Vol. 11). John Wiley and Sons.

Meyer, C., Corten, K., Wesseling, M., Peers, K., Simon, J.-P., Jonkers, I., and Desloovere, K. (2013). Test-retest reliability of innovated strength tests for hip muscles. *PloS One*, *8*(11), e81149.

Meyer, E. G., & Haut, R. C. (2005). Excessive compression of the human tibio-femoral joint causes ACL rupture. *Journal of biomechanics*, *38*(11), 2311-2316.

Meyer, E. G., & Haut, R. C. (2008). Anterior cruciate ligament injury induced by internal tibial torsion or tibiofemoral compression. *Journal of biomechanics*, *41*(16), 3377-3383.

Mihata, L. C. S., Beutler, A. I., and Boden, B. P. (2006). Comparing the Incidence of Anterior Cruciate Ligament Injury in Collegiate Lacrosse, Soccer, and Basketball Players: Implications for Anterior Cruciate Ligament Mechanism and Prevention. *American Journal of Sports Medicine*, *34*(6), 899–904. http://doi.org/10.1177/0363546505285582

Miller, M. D., Thompson, S. R. and Hart, J. A. 2012. *Review of orthopaedics*, Elsevier Health Sciences.

Milner, C. E., Fairbrother, J. T., Srivatsan, A., and Zhang, S. (2012). Simple verbal instruction improves knee biomechanics during landing in female athletes. *The Knee*, *19*(4), 399–403.

Miyasaka, T., Matsumoto, H., Suda, Y., Otani, T., & Toyama, Y. (2002). Coordination of the anterior and posterior cruciate ligaments in constraining the varus–valgus and internal–external rotatory instability of the knee. *Journal of orthopaedic science*, *7*(3), 348-353.

Moher, D., Liberati, A., Tetzlaff, J., and Altman, D. G. (2009). Reprint--preferred reporting items for systematic reviews and meta-analyses: the PRISMA statement. *Physical Therapy*, *89*(9), 873–880. http://doi.org/10.1136/bmj.b2535

Morris, A. D., Kemp, G. J., Lees, A., and Frostick, S. P. (1998). A study of the reproducibility of three different normalisation methods in intramuscular dual fine wire electromyography of the shoulder. *Journal of Electromyography and Kinesiology*, *8*(5), 317-322.

Munn, J., Sullivan, S. J., and Schneiders, A. G. (2010). Evidence of sensorimotor deficits in functional ankle instability: a systematic review with meta-analysis. *Journal of Science and Medicine in Sport*, *13*(1), 2-12.

Myer, G. D., Chu, D. A., Brent, J. L., and Hewett, T. E. (2008). Trunk and hip control neuromuscular training for the prevention of knee joint injury. *Clinics in sports medicine*, *27*(3), 425-448.

Myer, G. D., Ford, K. R., and Hewett, T. E. (2004). Methodological approaches and rationale for training to prevent anterior cruciate ligament injuries in female athletes. *Scandinavian Journal of Medicine and Science in Sports*, *14*(5), 275–285. http://doi.org/10.1111/j.1600-0838.2004.00410.x

Myer, G. D., Ford, K. R., and Hewett, T. E. (2011). New method to identify athletes at high risk of ACL injury using clinic-based measurements and freeware computer analysis. *British Journal of Sports Medicine*, *45*(4), 238–244. http://doi.org/10.1136/bjsm.2010.072843

Myer, G. D., Ford, K. R., Foss, K. D. B., Goodman, A., Ceasar, A., Rauh, M. J., ... and Hewett, T. E. (2010). The incidence and potential pathomechanics of patellofemoral pain in female athletes. *Clinical biomechanics*, *25*(7), 700-707.

Myer, G. D., Sugimoto, D., Thomas, S., and Hewett, T. E. (2013). The influence of age on the effectiveness of neuromuscular training to reduce anterior cruciate ligament injury in female athletes: a meta-analysis. *The American Journal of Sports Medicine*, *41*(1), 203–15. http://doi.org/10.1177/0363546512460637

Myklebust, G., Engebretsen, L., Braekken, I. H., Skjølberg, A., Olsen, O.-E., and Bahr, R. (2003). Prevention of anterior cruciate ligament injuries in female team handball players: a prospective intervention study over three seasons. *Clinical Journal of Sport Medicine*, *13*(2), 71–78. http://doi.org/10.1097/00042752-200303000-00002

Nakagawa, T. H. (2012). Trunk, Pelvis, Hip, and Knee Kinematics, Hip Strength, and Gluteal Muscle Activation During a Single Leg Squat in Males and Females With and Without Patellofemoral Pain Syndrome. *Journal of Orthopaedic and Sports Physical Therapy*, *42*(6), 491–501. http://doi.org/10.2519/jospt.2012.3987

Nakagawa, T. H., Moriya, É. T. U., Maciel, C. D., and Serrão, F. V. (2014). Test–retest reliability of three-dimensional kinematics using an electromagnetic tracking system during single-leg squat and stepping maneuver. *Gait and posture*, *39*(1), 141-146.

Negrete, R. J., Schick, E. A., and Cooper, J. P. (2007). Lower-limb dominance as a possible etiologic factor in noncontact anterior cruciate ligament tears. *Journal of Strength and Conditioning Research*, *21*(1), 270.

Neumann, D. A. (2010). Kinesiology of the hip: a focus on muscular actions. *The Journal of Orthopaedic and Sports Physical Therapy*, *40*(2), 82–94.

Neumann, D. A., Cook, T. M., Sholty, R. L., and Sobush, D. C. (1992). An electromyographic analysis of hip abductor muscle activity when subjects are carrying loads in one or both hands. *Physical therapy*, *72*(3), 207-217.

Nguyen, A.-D., Shultz, S. J., Schmitz, R. J., Luecht, R. M., and Perrin, D. H. (2011). A preliminary multifactorial approach describing the relationships among lower extremity alignment, hip muscle activation, and lower extremity joint excursion. *Journal of Athletic Training*, *46*(3), 246. Retrieved from http://www.ncbi.nlm.nih.gov/pmc/articles/PMC3419552/pdf/i1062-6050-46-3-246.pdf

Norcross, M. F., Blackburn, J. T., and Goerger, B. M. (2010). Reliability and interpretation of single leg stance and maximum voluntary isometric contraction methods of electromyography normalization. *Journal of Electromyography and Kinesiology*, *20*(3), 420-425.

Norcross, M., Halverson, S., Hawkey, T., Blackburn, J., and Padua, D. (2009). Evaluation of the lateral step-down test as a clinical assessment of hip musculature strength. Athletic Training and Sports Health Care. 1: 272-278. doi: 10.3928/19425864-20091019-0

Oh, Y. K., Lipps, D. B., Ashton-Miller, J. A., and Wojtys, E. M. (2012). What strains the anterior cruciate ligament during a pivot landing? *Am J Sports Med, 40*(3), 574- 583. doi: 10.1177/0363546511432544

Olsen, O., Myklebust, G., Engebretsen, L., and Bahr, R. (2004). Injury Mechanisms for Anterior Cruciate Ligament Injuries in Team Handball. *Americen Journal of Sports Medicine*, *32*(4), 1002–1012. http://doi.org/10.1177/0363546503261724

Oñate, J. A., Guskiewicz, K. M., Marshall, S. W., Giuliani, C., Yu, B., and Garrett, W. E. (2005). Instruction of jump-landing technique using videotape feedback. *The American journal of sports medicine*, *33*(6), 831-842.

Orishimo, K. F., Kremenic, I. J., Pappas, E., Hagins, M., and Liederbach, M. (2009). Comparison of landing biomechanics between male and female professional dancers. *The American Journal of Sports Medicine*, *37*(11), 2187–2193. http://doi.org/10.1177/0363546509339365

Ortiz, A., Olson, S., Libby, C. L., Kwon, Y. H., and Trudelle-Jackson, E. (2007). Kinematic and kinetic reliability of two jumping and landing physical performance tasks in young adult women. *North American journal of sports physical therapy: NAJSPT*, *2*(2), 104.

Ortiz, A., Olson, S., Trudelle-Jackson, E., Rosario, M., and Venegas, H. L. (2011). Landing mechanics during side hopping and crossover hopping maneuvers in noninjured women and women with anterior cruciate ligament reconstruction. *PMandR*, *3*(1), 13-20.

Padua, D. a, Marshall, S. W., Boling, M. C., Thigpen, C. a, Garrett, W. E., and Beutler, A. I. (2009). The Landing Error Scoring System (LESS) Is a valid and reliable clinical assessment tool of jump-landing biomechanics: The JUMP-ACL study. *The American Journal of Sports Medicine*, *37*(10), 1996–2002. http://doi.org/10.1177/0363546509343200

Padua, D. A., Carcia, C. R., Arnold, B. L., and Granata, K. P. (2005). Gender Differences in Leg Stiffness and Stiffness Recruitment Strategy During Two-Legged Hopping. *Journal of Motor Behavior*, *37*(2), 111–126. http://doi.org/10.3200/JMBR.37.2.111-126

Pandy, M. G., and Shelburne, K. B. (1997). Dependence of cruciate-ligament loading on muscle forces and external load. *Journal of Biomechanics*, *30*(10), 1015–1024. http://doi.org/10.1016/S0021-9290(97)00070-5

Pappas, E., Hagins, M., Sheikhzadeh, A., Nordin, M., and Rose, D. (2007). Biomechanical differences between unilateral and bilateral landings from a jump: gender differences. *Clinical Journal of Sport Medicine : Official Journal of the Canadian Academy of Sport Medicine*, *17*(4), 263–268. http://doi.org/10.1097/JSM.0b013e31811f415b

Patrek, M. F., Kernozek, T. W., Willson, J. D., Wright, G. A., and Doberstein, S. T. (2011). Hip-abductor fatigue and single-leg landing mechanics in women athletes. *Journal of*

Athletic Training, *46*(1), 31. Retrieved from http://www.ncbi.nlm.nih.gov/pmc/articles/PMC3017486/pdf/i1062-6050-46-1-31.pdf

Patrek, M. F., Kernozek, T. W., Willson, J. D., Wright, G. A., and Doberstein, S. T. (2011). Hip-abductor fatigue and single-leg landing mechanics in women athletes. J Athl Train, 46(1), 31-42. doi:10.4085/1062-6050-46.1.31

PEASE, W. S. and LEW, H. L. 2007. *Johnson's practical electromyography*, Lippincott Williams and Wilkins Philadelphia:.

Perrin DH. *Isokinetic Exercise and Assessment* (1993). Human Kinetics, Champaign, IL, 212 pp.

Petersen, W., and Zantop, T. (2007). Anatomy of the anterior cruciate ligament with regard to its two bundles. *Clinical orthopaedics and related research*, *454*, 35-47.

Pfeiffer, R. P., Shea, K. G., Roberts, D., Grandstrand, S., and Bond, L. (2006). Lack of effect of a knee ligament injury prevention program on the incidence of noncontact anterior cruciate ligament injury. *J Bone Joint Surg Am, 88*(8), 1769-1774. doi: 10.2106/JBJS.E.00616

PINCIVERO, D. M., GANDAIO, C. B. and ITO, Y. 2003. Gender-specific knee extensor torque, flexor torque, and muscle fatigue responses during maximal effort contractions. *European journal of applied physiology,* 89**,** 134-141.

Piva, S. R., Goodnite, E. A., and Childs, J. D. (2005). Strength around the hip and flexibility of soft tissues in individuals with and without patellofemoral pain syndrome. *Journal of Orthopaedic and Sports Physical Therapy, 35*(12), 793-801.

Pollard, C. D., Sigward, S. M., Ota, S., Langford, K., and Powers, C. M. (2006). The influence of in-season injury prevention training on lower-extremity kinematics during landing in female soccer players. *Clinical Journal of Sport Medicine*, *16*(3), 223-227.

Powers, C. M. (2003). The influence of altered lower-extremity kinematics on patellofemoral joint dysfunction: a theoretical perspective. *Journal of Orthopaedic and Sports Physical Therapy*, *33*(11), 639–646.

Powers, C. M. (2010). The influence of abnormal hip mechanics on knee injury: a biomechanical perspective. *The Journal of Orthopaedic and Sports Physical Therapy*, *40*(2), 42–51. http://doi.org/10.2519/jospt.2010.3337

Presswood, L., Cronin, J., Keogh, J. W. L., and Whatman, C. (2008). Gluteus Medius:

Applied Anatomy, Dysfunction, Assessment, and Progressive Strengthening. *Strength and Conditioning Journal, 30*(5), 41-53. doi:10.1519/SSC.0b013e318187f19a

Quatman, C. E., Kiapour, A. M., Demetropoulos, C. K., Kiapour, A., Wordeman, S. C., Levine, J. W., … Hewett, T. E. (2014). Preferential loading of the ACL compared with the MCL during landing: a novel in sim approach yields the multiplanar mechanism of dynamic valgus during ACL injuries. *The American Journal of Sports Medicine*, *42*(1), 177–86. http://doi.org/10.1177/0363546513506558

Quatman, C. E., Quatman-Yates, C. C., and Hewett, T. E. (2010). A "plane" explanation of anterior cruciate ligament injury mechanisms: A systematic review. *Sports Medicine*, *40*(9), 729–746. http://doi.org/10.2165/11534950-000000000-00000

Quatman-Yates, C. C., Quatman, C. E., Meszaros, A. J., Paterno, M. V., & Hewett, T. E. (2012). A systematic review of sensorimotor function during adolescence: a developmental stage of increased motor awkwardness?. *Br J Sports Med*, *46*(9), 649-655.

Queen, R. M., Gross, M. T., and Liu, H. Y. (2006). Repeatability of lower extremity kinetics and kinematics for standardized and self-selected running speeds. *Gait and posture*, *23*(3), 282-287.

Rabin, A., and Kozol, Z. (2010). Measures of Range of Motion and Strength Among Healthy Women With Differing Quality of Lower Extremity Movement During the Lateral Step-Down Test. *Journal of Orthopaedic and Sports Physical Therapy*, *40*(12), 792–800. http://doi.org/10.2519/jospt.2010.3424

Rankin, G, and Stokes, M. (1998). Reliability of assessment tools in rehabilitation: an illustration of appropriate statistical analyses. *Clinical Rehabilitation, 12*, 187-199.

Reiman, M. P., Bolgla, L. A., and Loudon, J. K. (2012). A literature review of studies evaluating gluteus maximus and gluteus medius activation during rehabilitation exercises. *Physiother Theory Pract,* *28*(4), 257-268. doi:10.3109/09593985.2011.604981

Requiao, L. F., Nadeau, S., Milot, M. H., Gravel, D., Bourbonnais, D. and Gagnon, D. 2005. Quantification of level of effort at the plantarflexors and hip extensors and flexor muscles in healthy subjects walking at different cadences. Journal of Electromyography and Kinesiology 15(4), pp.393-405.

Resende, R. A., Deluzio, K. J., Kirkwood, R. N., Hassan, E. A., and Fonseca, S. T. (2015). Increased unilateral foot pronation affects lower limbs and pelvic biomechanics during walking. *Gait and posture*, *41*(2), 395-401.

Ross S, Guskiewicz K, Prentice W, Schneider R, Yu B. (2004). Comparison of biomechanical factors between the kicking and stance limbs. J Sports Rehabil, 13:135-150.

Ruedl, G., Webhofer, M., Helle, K., Strobl, M., Schranz, A., Fink, C., … Burtscher, M. (2012). Leg dominance is a risk factor for noncontact anterior cruciate ligament injuries in female recreational skiers. *The American Journal of Sports Medicine*, *40*(6), 1269–73. http://doi.org/10.1177/0363546512439027

Russell, K. A., Palmieri, R. M., Zinder, S. M., and Ingersoll, C. D. (2006). Sex differences in valgus knee angle during a single-leg drop jump. *Journal of athletic training*, *41*(2), 166.

Saenz, A., Avellanet, M., Hijos, E., Chaler, J., Garreta, R., Pujol, E., ... and Farreny, A. (2010). Knee isokinetic test-retest: a multicentre knee isokinetic test-retest study of a fatigue protocol. *European journal of physical and rehabilitation medicine*, *46*(1), 81-88.

Scattone Silva, R., and Serrão, F. V. (2014). Sex differences in trunk, pelvis, hip and knee kinematics and eccentric hip torque in adolescents. *Clinical Biomechanics*, *29*(9), 1063–1069. http://doi.org/10.1016/j.clinbiomech.2014.08.004

Schmitz, R. J., Kulas, A. S., Perrin, D. H., Riemann, B. L., and Shultz, S. J. (2007). Sex differences in lower extremity biomechanics during single leg landings. *Clinical Biomechanics*, *22*(6), 681–688. http://doi.org/10.1016/j.clinbiomech.2007.03.001

Selkowitz, D. M., Beneck, G. J., and Powers, C. M. (2013). Which exercises target the gluteal muscles while minimizing activation of the tensor fascia lata? Electromyographic assessment using fine-wire electrodes. *J Orthop Sports Phys Ther, 43*(2), 54-64. doi:10.2519/jospt.2013.4116

Sell, T. C., Chu, Y., Abt, J. P., Nagai, T., Deluzio, J., Mcgrail, M. A., Rowe, R. S. and Lephart, S. M. 2010. Minimal additional weight of combat equipment alters air assault soldiers' landing biomechanics. *Military medicine,* 175**,** 41-47.

SENIAM. (2011). http://www.seniam.org/.

Shelbourne, K. D., Davis, T. J., and Klootwyk, T. E. (1998). The relationship between intercondylar notch width of the femur and the incidence of anterior cruciate ligament tears. A prospective study. *The American Journal of Sports Medicine*, *26*(3), 402–8. http://doi.org/10.1177/03635465980260031001

Shimokochi, Y., and Shultz, S. J. (2008). Mechanisms of noncontact anterior cruciate ligament injury. *Journal of Athletic Training*, *43*(4), 396–408. http://doi.org/10.4085/1062-6050-43.4.396

Shimokochi, Y., Yong Lee, S., Shultz, S. J., and Schmitz, R. J. (2009). The relationships among sagittal-plane lower extremity moments: implications for landing strategy in anterior cruciate ligament injury prevention. *Journal of Athletic Training*, *44*(1), 33-38.

Shin, C. S., Chaudhari, A. M., and Andriacchi, T. P. (2009). The effect of isolated valgus moments on ACL strain during single-leg landing: A simulation study. *Journal of Biomechanics*, *42*(3), 280–285. http://doi.org/10.1016/j.jbiomech.2008.10.031

Shin, C. S., Chaudhari, A. M., and Andriacchi, T. P. (2011). Valgus plus internal rotation moments increase anterior cruciate ligament strain more than either alone. *Medicine and Science in Sports and Exercise*, *43*(8), 1484-1491.

Shultz, S. J., and Schmitz, R. J. (2009). Effects of transverse and frontal plane knee laxity on hip and knee neuromechanics during drop landings. *American Journal of Sports Medicine, 37*(9), 1821-1830.

Shultz, S. J., Carcia, C. R., and Perrin, D. H. (2004). Knee joint laxity affects muscle activation patterns in the healthy knee. *Journal of Electromyography and Kinesiology, 14*(4), 475-483.

Shultz, S. J., Nguyen, A. D., Leonard, M. D., and Schmitz, R. J. (2009). Thigh strength and activation as predictors of knee biomechanics during a drop jump task. *Medicine and Science in Sports and Exercise*, *41*(4), 857–866. http://doi.org/10.1249/MSS.0b013e3181e3b3f

Sigward, S. M., Ota, S., and Powers, C. M. (2008). Predictors of Frontal Plane Knee Excursion During a Drop Land in Young Female Soccer Players. *Journal of Orthopaedic and Sports Physical Therapy*, *38*(11), 661–667. http://doi.org/10.2519/jospt.2008.2695

Sinsurin, K., Srisangboriboon, S., and Vachalathiti, R. (2017). Side-to-side differences in lower extremity biomechanics during multi-directional jump landing in volleyball athletes. *European Journal of Sport Science*, *0*(0), 1–11.

http://doi.org/10.1080/17461391.2017.1308560

Sinsurin, K., Vachalathiti, R., Jalayondeja, W., and Limroongreungrat, W. (2013). Altered peak knee valgus during jump-landing among various directions in basketball and volleyball athletes. *Asian Journal of Sports Medicine*, *4*(3), 195–200. http://doi.org/10.5812/asjsm.34258

Smith, B. A., Livesay, G. A. and Woo, S. L. 1993. Biology and biomechanics of the anterior cruciate ligament. Clinics in sports medicine 12(4), pp.637-670.

Smith, H. C., Johnson, R. J., Shultz, S. J., Tourville, T., Holterman, L. A., Slauterbeck, J., … Beynnon, B. D. (2012). A Prospective Evaluation of the Landing Error Scoring System (LESS) as a Screening Tool for Anterior Cruciate Ligament Injury Risk. *American Journal of Sports Medicine*, *40*(3), 521–526. http://doi.org/Doi 10.1177/0363546511429776

Smith, H. C., Vacek, P., Johnson, R. J., Slauterbeck, J. R., Hashemi, J., Shultz, S., and Beynnon, B. D. (2012). Risk Factors for Anterior Cruciate Ligament Injury. *Sports Health*, *4*(2), 155–161. http://doi.org/10.1177/1941738111428282

Sorenson, B., Kernozek, T. W., Willson, J. D., Ragan, R., and Hove, J. (2015). Two- and Three-Dimensional Relationships Between Knee and Hip Kinematic Motion Analysis: Single-Leg Drop-Jump Landings. *J Sport Rehabil, 24*(4), 363-372. doi:10-1123/jsr.2014-0206

Sonnery-Cottet, B., Archbold, P., Cucurulo, T., Fayard, J. M., Bortolletto, J., Thaunat, M., ... & Chambat, P. (2011). The influence of the tibial slope and the size of the intercondylar notch on rupture of the anterior cruciate ligament. *J Bone Joint Surg Br*, *93*(11), 1475-1478.

Souza, R. B., and Powers, C. M. (2009). Differences in hip kinematics, muscle strength, and muscle activation between subjects with and without patellofemoral pain. *Journal of Orthopaedic and Sports Physical Therapy*, *39*(1), 12–19.

Stearns, K. M., and Powers, C. M. (2014). Improvements in hip muscle performance result in increased use of the hip extensors and abductors during a landing task. *The American Journal of Sports Medicine*, *42*(3), 602–609. Retrieved from http://ajs.sagepub.com/content/42/3/602.full.pdf

Stearns, K. M., Keim, R. G., and Powers, C. M. (2013). Influence of relative hip and knee extensor muscle strength on landing biomechanics. *Medicine and Science in Sports and Exercise*, *45*(5), 935-941.

Stensrud, S., Myklebust, G., Kristianslund, E., Bahr, R., and Krosshaug, T. (2011). Correlation between two-dimensional video analysis and subjective assessment in evaluating knee control among elite female team handball players. *British Journal of Sports Medicine*, *45*(7), 589–95. http://doi.org/10.1136/bjsm.2010.078287

Stickler, L., Finley, M., and Gulgin, H. (2015). Relationship between hip and core strength and frontal plane alignment during a single leg squat. *Physical Therapy in Sport*, *16*(1), 66–71. http://doi.org/10.1016/j.ptsp.2014.05.002

Sugimoto, D., Alentorn-Geli, E., Mendiguchía, J., Samuelsson, K., Karlsson, J., and Myer, G. D. (2015). Biomechanical and neuromuscular characteristics of male athletes: implications for the development of anterior cruciate ligament injury prevention programs. *Sports Medicine (Auckland, N.Z.)*, *45*(6), 809–22. http://doi.org/10.1007/s40279-015-0311-1

Sugimoto, D., Mattacola, C. G., Mullineaux, D. R., Palmer, T. G., and Hewett, T. E. (2014). Comparison of isokinetic hip abduction and adduction peak torques and ratio between sexes. *Clinical Journal of Sport Medicine : Official Journal of the Canadian Academy of Sport Medicine*, *24*(5), 422–8. http://doi.org/10.1097/JSM.0000000000000059

Suzuki, H., Omori, G., Uematsu, D., Nishino, K., and Endo, N. (2015). the Influence of Hip Strength on Knee Kinematics During a Single-Legged Medial Drop Landing Among Competitive Collegiate Basketball Players. *International Journal of Sports Physical Therapy*, *10*(5), 592–601. Retrieved from http://www.pubmedcentral.nih.gov/articlerender.fcgi?artid=4595912andtool=pmcentrezandrendertype=abstract

Swearingen, J., Lawrence, E., Stevens, J., Jackson, C., Waggy, C., and Davis, D. S. (2011). Correlation of single leg vertical jump, single leg hop for distance, and single leg hop for time. *Phys Ther Sport, 12*(4), 194-198. doi: 10.1016/j.ptsp.2011.06.001

Taylor, K. A., Terry, M. E., Utturkar, G. M., Spritzer, C. E., Queen, R. M., Irribarra, L. A., … Defrate, L. E. (2011). Measurement of in vivo anterior cruciate ligament strain during dynamic jump landing. *Journal of Biomechanics*, *44*(3), 365–371. http://doi.org/10.1016/j.jbiomech.2010.10.028

Uhorchak, J. M., Scoville, C. R., Williams, G. N., Arciero, R. A., St. Pierre, P., and Taylor, D. C. (2003). Risk Factors Associated with Noncontact Injury of the Anterior Cruciate Ligament. A Prospective Four-Year Evaluation of 859 West Point Cadets. *American Journal of Sports Medicine*, *31*(6), 831–842. http://doi.org/0363-

5465/103/3131-0831$02.00/0

Utturkar, G. M., Irribarra, L. A., Taylor, K. A., Spritzer, C. E., Taylor, D. C., Garrett, W. E., and DeFrate, L. E. (2013). The effects of a valgus collapse knee position on in vivo ACL elongation. *Annals of biomedical engineering*, *41*(1), 123-130.

Wall, S. J., Rose, D. M., Sutter, E. G., Belkoff, S. M., & Boden, B. P. (2012). The role of axial compressive and quadriceps forces in noncontact anterior cruciate ligament injury: a cadaveric study. *The American journal of sports medicine*, *40*(3), 568-573.

Webster, K. E., McClelland, J. A., Wittwer, J. E., Tecklenburg, K., and Feller, J. A. (2010). Three dimensional motion analysis of within and between day repeatability of tibial rotation during pivoting. *The Knee*, *17*(5), 329-333.

Weeks, B. K., Carty, C. P., and Horan, S. A. (2012). Kinematic predictors of single-leg squat performance: a comparison of experienced physiotherapists and student physiotherapists. *BMC Musculoskeletal Disorders*, *13*, 207. http://doi.org/10.1186/1471-2474-13-207

Widler, K. S., Glatthorn, J. F., Bizzini, M., Impellizzeri, F. M., Munzinger, U., Leunig, M., and Maffiuletti, N. A. (2009). Assessment of hip abductor muscle strength. A validity and reliability study. *The Journal of Bone and Joint Surgery. American Volume*, *91*(11), 2666–72. http://doi.org/10.2106/JBJS.H.01119

Wikholm, J. B., & Bohannon, R. W. (1991). Hand-held dynamometer measurements: tester strength makes a difference. *Journal of Orthopaedic & Sports Physical Therapy*, *13*(4), 191-198.

Willson, J. D., and Davis, I. S. (2008). Utility of the frontal plane projection angle in females with patellofemoral pain. *Journal of Orthopaedic and Sports Physical Therapy*, *38*(10), 606-615.

Willson, J. D., Ireland, M. L., and Davis, I. (2006). Core strength and lower extremity alignment during single leg squats. *Medicine and Science in Sports and Exercise*, *38*(5), 945–52. http://doi.org/10.1249/01.mss.0000218140.05074.fa

Willy, R. W., and Davis, I. S. (2011). The effect of a hip-strengthening program on mechanics during running and during a single-leg squat. *The Journal of Orthopaedic and Sports Physical Therapy*, *41*(9), 625–632.

Willy, R. W., Scholz, J. P., and Davis, I. S. (2012). Mirror gait retraining for the treatment of patellofemoral pain in female runners. *Clinical Biomechanics*, *27*(10), 1045-1051.

Withrow, T. J., Huston, L. J., Wojtys, E. M., and Ashton-Miller, J. A. (2006). The effect of an impulsive knee valgus moment on in vitro relative ACL strain during a simulated jump landing. *Clinical Biomechanics*, *21*(9), 977–983. http://doi.org/10.1016/j.clinbiomech.2006.05.001

Woo, S. L. Y., Abramowitch, S. D., Kilger, R., and Liang, R. (2006). Biomechanics of knee ligaments: Injury, healing, and repair. *Journal of Biomechanics*, *39*(1), 1–20. http://doi.org/10.1016/j.jbiomech.2004.10.025

Woods, K., Bishop, P. and Jones, E. 2007. Warm-up and stretching in the prevention of muscular injury. Sports Medicine 37(12), pp.1089-1099.

Wyatt, M. P., and Edwards, A. M. (1981). Comparison of quadriceps and hamstring torque values during isokinetic exercise. *Journal of Orthopaedic and Sports Physical Therapy*, *3*(2), 48-56.

Yamazaki, J., Muneta, T., Ju, Y. J., and Sekiya, I. (2009). Differences in kinematics of single leg squatting between anterior cruciate ligament-injured patients and healthy controls. *Knee Surgery, Sports Traumatology, Arthroscopy*, *18*(1), 56–63. http://doi.org/10.1007/s00167-009-0892-z

Yeow, C. H., Lee, P. V. S., and Goh, J. C. H. (2009). Effect of landing height on frontal plane kinematics, kinetics and energy dissipation at lower extremity joints. *Journal of biomechanics*, *42*(12), 1967-1973.

Yu, B., Gabriel, D., Noble, L., and An, K. N. (1999). Estimate of the optimum cutoff frequency for the Butterworth low-pass digital filter. *Journal of Applied Biomechanics*, *15*(3), 318-329.

Zabala, M. E., Favre, J., and Andriacchi, T. P. (2015). Relationship Between Knee Mechanics and Time Since Injury in ACL-Deficient Knees Without Signs of Osteoarthritis. *The American Journal of Sports Medicine*, *43*(5), 1189–1196. http://doi.org/10.1177/0363546514567296

Zahradnik, D., Jandacka, D., Uchytil, J., Farana, R., and Hamill, J. (2015). Lower extremity mechanics during landing after a volleyball block as a risk factor for anterior cruciate ligament injury. *Physical Therapy in Sport*, *16*(1), 53–58. http://doi.org/10.1016/j.ptsp.2014.04.003

Zantop, T., Eggers, A. K., Musahl, V., Weimann, A., and Petersen, W. (2005). Cyclic testing of flexible all-inside meniscus suture anchors. *The American journal of sports medicine*, *33*(3), 388-394.

Zazulak, B. T., Ponce, P. L., Straub, S. J., Medvecky, M. J., Avedisian, L., and Hewett, T. E.

(2005). Gender comparison of hip muscle activity during single-leg landing. *The Journal of Orthopaedic and Sports Physical Therapy*, *35*(5), 292–299. http://doi.org/10.2519/jospt.2005.1734

Zebis, M. K., Andersen, L. L., Bencke, J., Kjær, M., and Aagaard, P. (2009). Identification of Athletes at Future Risk of Anterior Cruciate Ligament Ruptures by Neuromuscular Screening. *The American Journal of Sports Medicine*, *37*(10), 1967–1973. http://doi.org/10.1177/0363546509335000

Zeller, B. L., McCrory, J. L., Kibler, W. Ben, and Uhl, T. L. (2003). Differences in Kinematics and Electromyographic Activity Between Men and Women during the Single-Legged Squat*. *The American Journal of Sports Medicine*, *31*(3), 449–456. Retrieved from http://ajs.sagepub.com/content/31/3/449.full.pdf

Zhang, S., Derrick, T. R., Evans, W., and Yu, Y. (2008). Shock and impact reduction in moderate and strenuous landing activities. *Sports Biomechanics / International Society of Biomechanics in Sports*, *7*(2), 296–309. http://doi.org/10.1080/14763140701841936

Appendices:

Appendices (1)

Patients/selection bias	YES (published)	Unable to Determine	NO
1) Is the hypothesis/aim/objective of the study clearly described?			
2) Are the characteristics of the patients included in the study clearly described?			
3) Is the patient sample representative of patients treated in routine clinical practice?			
4) Is there information on possibility of selection bias present in study? *For example: were participants recruited from same population; recruited over same time period; randomized to group; was allocation concealed*			
Comparison			
5) Was a comparison group identified and clearly defined?			
Outcomes			
6) Are the main outcomes to be measured clearly described in the Introduction or Methods section? *If the main outcomes are first mentioned in the Results section, the question should be answered no.*			
7) Were the main outcome measures used accurate (valid and reliable)? *For studies where the outcome measures are clearly described, the question should be answered yes. For studies which refer to other work or that demonstrates the outcome measures are accurate, the question should be answered as yes.*			
8) Was an attempt made to blind those measuring the main outcomes of the intervention?			
Reported findings/statistical analysis			
9) Are the main findings of the study clearly described? *Simple outcome data (including denominators and numerators) should be reported for all major findings so that the reader can check the major analyses and conclusions (This question does not cover statistical tests which are considered below).*			
10) Does the study provide estimates of the random variability in the data for the main outcomes? *In non-normally distributed data the inter-quartile range of results should be reported. In normally distributed data the standard error, standard deviation or confidence intervals should be reported. If the distribution of the data is not described, it must be assumed that the estimates used were appropriate and the question should be answered yes.*			
11) Were the statistical tests used to assess the main outcomes appropriate? *The statistical techniques used must be appropriate to the data. For example nonparametric methods should be used for small sample sizes. Where little statistical analysis has been undertaken but where there is no evidence of bias, the question, should be answered yes. If the distribution of the data (normal or not) is not described it must be assumed that the estimates used were appropriate and the question should be answered yes* .			
Confounding			

12) Are the distributions of principal confounders in each group of subjects to be compared clearly described? *(e.g. age, sex, height, weight, activity level, sporting activity, player position, dominance, duration symptoms)*			
13) Was there adequate adjustment for confounding in the analyses from which the main findings were drawn?			
Power			
14) Was a sample size calculation reported? 15) Did the study have sufficient power to detect a clinically important effect where the probability value for a difference being due to chance is less than 5%? *Sample sizes have been calculated to detect a difference of x% and y%.*			

Study	1	2	3	4	5	6	7	8	9	10	11	12	13	14	15	**Total**	Methodology Quality
Hollaman et al. (2014)	1	1	0	0	1	1	0	1	1	0	1	1	1	0	1	**10**	Moderate
Hollaman et al. (2013)	1	1	0	0	1	1	0	1	1	0	1	1	1	0	1	**11**	Moderate
Homan et al. (2013	1	1	0	0	1	1	1	1	1	0	1	1	1	0	0	**10**	Moderate
Nguyen et al. (2001)	1	1	1	1	1	1	0	0	1	1	1	1	1	0	0	**11**	Moderate

Note. A score of ≥12 indicates high methodological quality, a score 10 or 11 indicates moderate quality, and a score ≤9 indicates low quality

Supplementary material: Methodological quality rating scores with the Modified Downs and Black Scale

Appendices (2)

University of Salford MANCHESTER

Research, Innovation and Academic Engagement Ethical Approval Panel

College of Health & Social Care
AD 101 Allerton Building
University of Salford
M6 6PU

T +44(0)161 295 2280
HSresearch@salford.ac.uk

www.salford.ac.uk/

28 May 2015

Dear Ziyad,

RE: ETHICS APPLICATION HSCR 15-19 – The Relationship of Gluteal Muscles Ability Performance to Knee Valgus During Functional Tasks

Based on the information you provided, I am pleased to inform you that application HSCR15-19 has been approved.

If there are any changes to the project and/ or its methodology, please inform the Panel as soon as possible by contacting HSresearch@salford.ac.uk

Yours sincerely,

Sue McAndrew
Chair of the Research Ethics Panel

University of
Salford
MANCHESTER

Research, Innovation and Academic Engagement Ethical Approval Panel

College of Health & Social Care
AD 101 Allerton Building
University of Salford
M6 6PU

T +44(0)161 295 2280
HSresearch@salford.ac.uk

www.salford.ac.uk/

13 July 2015

Dear Ziyad,

RE: ETHICS APPLICATION HSCR 15-47 – The Relationship Between Gluteal Muscles Strength Performance and Knee Valgus

Based on the information you provided, I am pleased to inform you that application HSCR15-47 has been approved.

If there are any changes to the project and/ or its methodology, please inform the Panel as soon as possible by contacting HSresearch@salford.ac.uk

Yours sincerely,

Sue McAndrew
Chair of the Research Ethics Panel

<u>Informed Consent Form</u>

1. <u>The researcher</u>, is a postgraduate research student at the University of Salford, has requested my participation in a research study. My involvement in the study and its purpose has been fully explained to me.

2. My participation in this research will involve a number of muscle power tests, which include hip abductors, extensors and lateral rotators and muscles.

3. I have been informed that I will not be compensated for my participation.

4. I understand that the results of this research may be published, but my name or identity will not be revealed at any time. In order to keep my records confidential, <u>the researcher</u> will store all the data as numbered codes in a computer that will only be accessed by him.

5. I have been informed that the researcher will answer any further questions that I have at any time concerning the research or my participation and I can contact him at his e-mail address.

6. I understand that I may withdraw my consent and participation at any time without objection from the researcher.

7. I understand that if I withdraw from the study, all the information about me will be destroyed and not to be used in the study at all.

8. I have read and understand the participation information sheet and have had the chance to ask questions.

Name: .... **Signed:** .... **Date:** ……......

Appendix 4

Tick which type of exercise activity the subject will be participating in:

Maximal exercise ☐ Submaximal exercise ☐ other ☐……………………
(Please specify)

1. **Personal information**

Surname: …………………………… Forename(s): …………………………...

Date of birth: ……………................... Age: ……………………….…………………..

Height (cm): ………………………..…. Weight (kg): ……………..……………………

2. **Additional information**

a. Please state when you last had something to eat / drink……………………...

b. Tick the box that relates to your present level of activity:

Inactive ☐ moderately active ☐ highly active ☐

c. Give an example of a typical weeks exercise:

……………………………………………………………………………………...

d. If you smoke, approximately how many cigarettes do you smoke a day (   )

3.	Are you currently taking any medication that might affect your ability to participate in the test as outlined?	YES	NO
4.	Do you suffer, or have you ever suffered from, cardiovascular disorders? e.g. Chest pain, heart trouble, cholesterol etc.	YES	NO
5.	Do you suffer, or have you ever suffered from, high/low blood pressure?	YES	NO
6.	Has your doctor said that you have a condition and that you should only do physical activity recommended by a doctor?	YES	NO
7.	Have you had a cold or feverish illness in the last 2 weeks?	YES	NO
8.	Do you ever lose balance because of dizziness, or do you ever lose consciousness?	YES	NO
9.	Do you suffer, or have you ever suffered from, respiratory disorders? e.g. Asthma, bronchitis etc.	YES	NO

10	Are you currently receiving advice from a medical advisor i.e. GP or Physiotherapist **not** to participate in physical activity because of back pain or any musculoskeletal (muscle, joint or bone) problems?	YES	NO
11	Do you suffer, or have you ever suffered from diabetes?	YES	NO
12	Do you suffer, or have you ever suffered from epilepsy/seizures?	YES	NO
13	Do you know of any reason, not mentioned above, why you should not exercise? e.g. Head injury (within 12 months), pregnant or new mother, hangover, eye injury or anything else.	YES	NO
14	Do you have any allergies, athletic tape or sticking plasters?	YES	NO

15 Health Questionnaire/Exclusion Criteria:

Are you suffering from, or have you ever suffered any of the following in the last 6 months:

- History of heart problems.
- Diabetes mellitus.
- Asthma, breathing or lung problems.
- Allergies.
- Cancer.
- Seizures, Seizure medication, neurological problems or dizziness.
- High blood pressure.
- Back problems.
- Lower limb joint or muscular disorders.
- Recent surgery.
- Hernia or any condition that may be aggravated by exercises.
- Skeletal injuries: Back, neck, head, knee, and hip.
- If female: are you or is there any chance you may be pregnant.

Please note: if you answered YES to any of the above questions, you will be excluded from the study.

Appendices 3

Shapiro-Wilk Tests of Normality for SLS task

Variables	Right			Left		
	Statistic	df	Sig.	Statistic	df	Sig.
Hip Flexion Angle	.988	34	.967	.986	34	.939
Hip Flexion Moment	.980	34	.771	.982	34	.837
Hip Int. Rot. Angle	.966	34	.357	.972	34	.521
Hip Int. Rot. Moment	.972	34	.518	.978	34	.704
Hip Adduction Angle	.939	34	.059	.985	34	.902
Hip Adduction Moment	.971	34	.486	.977	34	.663
Knee Valgus Angle	.958	34	.211	.963	34	.304
Knee Valgus Moment	.949	34	.115	.903	34	.005
Knee Flexion Angle	.972	34	.531	.954	34	.167
Knee Flexion Moment	.966	34	.354	.978	34	.696
GRFV	.834	34	.000	.906	34	.007

Shapiro-Wilk Tests of Normality for FL task

Variables	Right			Left		
	Statistic	df	Sig.	Statistic	df	Sig.
Hip Flexion Angle	.940	34	.061	.956	34	.185
Hip Flexion Moment	.973	34	.538	.948	34	.105
Hip Int. Rot. Angle	.980	34	.767	.974	34	.572
Hip Int. Rot. Moment	.952	34	.137	.911	34	.009
Hip Adduction Angle	.952	34	.138	.949	34	.116
Hip Adduction Moment	.960	34	.248	.977	34	.689
Knee Valgus Angle	.945	34	.087	.975	34	.608
Knee Valgus Moment	.928	34	.027	.926	34	.023
Knee Flexion Angle	.981	34	.817	.934	34	.041
Knee Flexion Moment	.966	34	.353	.960	34	.250
GRFV	.953	34	.152	.953	34	.152

Shapiro-Wilk Tests of Normality for SML task

Variables	Right			Left		
	Statistic	df	Sig.	Statistic	df	Sig.
Hip Flexion Angle	.962	34	.275	.982	34	.822
Hip Flexion Moment	.906	34	.007	.972	34	.505
Hip Int. Rot. Angle	.988	34	.969	.982	34	.831
Hip Int. Rot. Moment	.982	34	.836	.876	34	.001
Hip Adduction Angle	.964	34	.325	.958	34	.213
Hip Adduction Moment	.984	34	.879	.979	34	.732
Knee Valgus Angle	.973	34	.543	.963	34	.290
Knee Valgus Moment	.897	34	.004	.786	34	.000
Knee Flexion Angle	.967	34	.372	.935	34	.045
Knee Flexion Moment	.958	34	.212	.981	34	.811
GRFV	.956	34	.182	.932	34	.035

Shapiro-Wilk Tests of Normality for SLL task

Variables	Right			Left		
	Statistic	df	Sig.	Statistic	df	Sig.
Hip Flexion Angle	.968	34	.410	.980	34	.784
Hip Flexion Moment	.950	34	.121	.938	34	.055
Hip Int. Rot. Angle	.982	34	.841	.979	34	.752
Hip Int. Rot. Moment	.943	34	.077	.973	34	.558
Hip Adduction Angle	.975	34	.610	.989	34	.980
Hip Adduction Moment	.920	34	.017	.964	34	.321
Knee Valgus Angle	.986	34	.932	.921	34	.018
Knee Valgus Moment	.825	34	.000	.900	34	.004
Knee Flexion Angle	.972	34	.509	.955	34	.169
Knee Flexion Moment	.970	34	.468	.966	34	.367
GRFV	.963	34	.300	.988	34	.971

Shapiro-Wilk Tests of Normality for Gluteus Maximus and Gluteus Medius

Variables	Right			Left		
	Statistic	df	Sig.	Statistic	df	Sig.
SLS						
Gluteus Maximus	.974	34	.564	.872	34	.001
Gluteus Medius	.976	34	.644	.982	34	.821
FL						
Gluteus Maximus	.942	34	.071	.959	34	.227
Gluteus Medius	.969	34	.422	.970	34	.453
SML						
Gluteus Maximus	.941	34	.065	.953	34	.156
Gluteus Medius	.986	34	.935	.965	34	.330
SLL						
Gluteus Maximus	.968	34	.414	.971	34	.497
Gluteus Medius	.979	34	.753	.968	34	.413

Shapiro-Wilk Tests of Normality for Hip Abductors and Extensors tests

Variables	Right			Left		
	Statistic	df	Sig.	Statistic	df	Sig.
Extension						
Concentric	.936	34	.048	.929	34	.030
Eccentric	.948	34	.109	.959	34	.221
Extension / BW						
Concentric	.974	34	.577	.957	34	.196
Eccentric	.940	34	.062	.954	34	.162
Abduction						
Concentric	.980	34	.758	.938	34	.054
Eccentric	.949	34	.115	.945	34	.085
Abduction / BW						
Concentric	.969	34	.430	.915	34	.012
Eccentric	.971	34	.489	.917	34	.214

Appendices 4

t tests - Correlation: Point biserial model

Analysis:	A priori: Compute required sample size		
Input:	Tail(s)	=	Two
	Effect size $\lvert\rho\rvert$	=	0.4472136
	α err prob	=	0.05
	Power (1-β err prob)	=	0.80
Output:	Noncentrality parameter δ	=	2.9154760
	Critical t	=	2.0369333
	Df	=	32
	Total sample size	=	34
	Actual power	=	0.8070367

Appendices 5

Female Rt. SLS											
		Gmax EMG	Gmed EMG	Ext CON.	Ext. Ecc.	Abd. Con.	AbdEcc	Ext CON. / BW	Ext. Ecc. / BW	Abd. Con. / BW	AbdEcc. / BW
Hip Flex Angle	Correlation	-.531*	0.189	0.025	0.027	0.307	0.346	0.114	0.088	0.420	0.330
	Sig. (2-tailed)	0.028	0.467	0.925	0.919	0.231	0.173	0.662	0.738	0.094	0.195
Hip Flex Moment	Correlation	0.277	-0.416	0.017	0.029	-0.094	-0.270	-0.036	0.008	-0.241	-0.242
	Sig. (2-tailed)	0.283	0.097	0.947	0.913	0.720	0.294	0.890	0.975	0.352	0.350
Hip IntRot Angle	Correlation	-0.234	0.007	0.102	-0.111	0.458	0.315	0.022	-0.247	.510*	0.182
	Sig. (2-tailed)	0.366	0.978	0.697	0.672	0.064	0.218	0.933	0.339	0.037	0.484
Hip IntRot Moment	Correlation	0.096	-0.375	-0.263	-0.249	-0.335	-0.092	-0.295	-0.238	-0.311	-0.046
	Sig. (2-tailed)	0.714	0.138	0.307	0.335	0.188	0.725	0.251	0.358	0.225	0.862
Hip Add Angle	Correlation	-.653**	-0.392	-0.026	-0.170	0.273	.590*	0.063	-0.120	0.406	.553*
	Sig. (2-tailed)	0.004	0.120	0.922	0.515	0.289	0.013	0.812	0.645	0.106	0.021
Hip Add Moment	Correlation	.526*	0.141	-0.178	-0.022	-0.389	-.606**	-0.219	0.002	-.483*	-0.473
	Sig. (2-tailed)	0.030	0.588	0.494	0.932	0.123	0.010	0.398	0.993	0.050	0.055
Knee Valgus Angle	Correlation	-0.290	0.005	0.137	-0.043	.551*	0.168	0.190	-0.055	.603*	0.138
	Sig. (2-tailed)	0.259	0.985	0.601	0.869	0.022	0.520	0.466	0.835	0.010	0.599
Knee Valgus Moment	Correlation	0.372	0.222	-0.300	-0.019	-.617**	-0.328	-0.269	0.086	-.561*	-0.182
	Sig. (2-tailed)	0.142	0.392	0.242	0.943	0.008	0.198	0.297	0.744	0.019	0.484
Knee Flex Angle	Correlation	-0.187	0.122	0.182	0.339	-0.315	0.087	0.314	0.461	-0.310	0.094
	Sig. (2-tailed)	0.473	0.642	0.486	0.183	0.218	0.739	0.219	0.063	0.225	0.718
Knee Flex Moment	Correlation	0.384	0.032	0.314	0.424	-0.436	-0.272	0.180	0.305	-.630**	-0.405
	Sig. (2-tailed)	0.128	0.902	0.220	0.089	0.080	0.290	0.489	0.233	0.007	0.107
GRFV	Correlation	0.233	0.421	0.309	0.342	0.225	0.147	0.108	0.121	0.104	-0.132
	Sig. (2-tailed)	0.368	0.092	0.227	0.179	0.386	0.574	0.679	0.644	0.690	0.615

*. Correlation is significant at the 0.05 level (2-tailed).

**. Correlation is significant at the 0.01 level (2-tailed).
Highlighted Variables = Non-parametric data

Female Lt. SLS											
		Gmax EMG	Gmed EMG	Ext CON.	Ext. Ecc.	Abd. Con.	AbdEcc	Ext CON. / BW	Ext. Ecc. / BW	Abd. Con. / BW	AbdEcc. / BW
Hip Flex Angle	Correlation	-0.269	-0.288	-0.130	0.104	0.153	0.316	-0.215	0.076	0.133	0.322
	Sig. (2-tailed)	0.297	0.263	0.620	0.690	0.557	0.216	0.408	0.772	0.612	0.207
Hip Flex Moment	Correlation	.533*	0.235	-0.122	-0.166	-0.210	-0.344	0.238	0.159	0.071	-0.119
	Sig. (2-tailed)	0.028	0.365	0.641	0.525	0.419	0.177	0.357	0.541	0.785	0.651
Hip IntRot Angle	Correlation	0.445	0.203	0.300	0.170	-0.013	0.204	.514*	0.336	0.055	0.264
	Sig. (2-tailed)	0.074	0.434	0.241	0.515	0.959	0.432	0.035	0.188	0.833	0.305
Hip IntRot Moment	Correlation	0.035	0.160	-0.095	-0.266	-0.285	-0.343	0.117	-0.097	-0.117	-0.198
	Sig. (2-tailed)	0.893	0.538	0.717	0.303	0.267	0.178	0.654	0.710	0.655	0.447
Hip Add Angle	Correlation	0.384	0.145	.647**	.625**	.488*	.605*	.689**	.631**	0.385	.521*
	Sig. (2-tailed)	0.128	0.578	0.005	0.007	0.047	0.010	0.002	0.007	0.127	0.032
Hip Add Moment	Correlation	-0.049	-0.074	-0.396	-0.357	-0.453	-.545*	-0.290	-0.224	-0.300	-0.408
	Sig. (2-tailed)	0.852	0.777	0.115	0.159	0.068	0.024	0.259	0.388	0.243	0.104
Knee Valgus Angle	Correlation	-0.106	-0.401	-0.302	-0.245	-0.257	-0.044	-0.188	-0.106	-0.124	0.085
	Sig. (2-tailed)	0.686	0.110	0.239	0.343	0.319	0.867	0.470	0.684	0.637	0.746
Knee Valgus Moment	Correlation	-0.403	-0.065	-0.269	-0.270	-0.117	-0.233	-0.464	-0.441	-0.160	-0.261
	Sig. (2-tailed)	0.109	0.805	0.296	0.295	0.655	0.367	0.060	0.076	0.541	0.311
Knee Flex Angle	Correlation	-0.308	-0.095	-.630**	-0.476	-0.197	-0.183	-0.362	-0.198	0.180	0.161
	Sig. (2-tailed)	0.229	0.718	0.007	0.053	0.449	0.482	0.153	0.445	0.489	0.537
Knee Flex Moment	Correlation	-0.109	-0.054	-.555*	-0.259	-0.301	-0.376	-0.456	-0.098	-0.107	-0.213
	Sig. (2-tailed)	0.678	0.837	0.021	0.315	0.240	0.137	0.066	0.707	0.684	0.413
GRFV	Correlation	-0.158	-0.080	-0.067	0.076	0.034	-0.041	-0.366	-0.169	-0.160	-0.219
	Sig. (2-tailed)	0.544	0.760	0.800	0.772	0.896	0.874	0.149	0.518	0.540	0.399

*. Correlation is significant at the 0.05 level (2-tailed).

**. Correlation is significant at the 0.01 level (2-tailed).
Highlighted Variables = Non-parametric data

Female Rt. FL											
		Gmax EMG	Gmed EMG	Ext CON.	Ext. Ecc.	Abd. Con.	AbdEcc	Ext CON. / BW	Ext. Ecc. / BW	Abd. Con. / BW	AbdEcc. / BW
Hip Flex Angle	Correlation	0.363	0.186	0.399	0.264	0.209	0.424	0.433	0.144	0.100	0.363
	Sig. (2-tailed)	0.152	0.474	0.113	0.305	0.420	0.090	0.083	0.581	0.703	0.152
Hip Flex Moment	Correlation	-0.096	-0.256	-0.047	0.242	0.130	-0.182	-0.288	0.194	0.009	-0.096
	Sig. (2-tailed)	0.714	0.321	0.857	0.349	0.618	0.484	0.263	0.456	0.972	0.714
Hip IntRot Angle	Correlation	0.303	-0.029	0.083	0.426	0.210	-0.050	-0.325	0.440	0.079	0.303
	Sig. (2-tailed)	0.237	0.913	0.753	0.088	0.419	0.849	0.202	0.077	0.762	0.237
Hip IntRot Moment	Correlation	-0.130	0.002	-0.223	-0.174	0.102	-0.021	0.023	-0.020	0.271	-0.130
	Sig. (2-tailed)	0.619	0.994	0.389	0.504	0.697	0.935	0.932	0.938	0.292	0.619
Hip Add Angle	Correlation	0.235	0.125	0.291	0.249	0.285	0.391	0.231	0.290	0.247	0.235
	Sig. (2-tailed)	0.364	0.633	0.258	0.334	0.268	0.121	0.372	0.259	0.339	0.364
Hip Add Moment	Correlation	0.050	-0.162	-0.059	0.120	0.064	-0.202	-0.069	0.034	-0.062	0.050
	Sig. (2-tailed)	0.850	0.535	0.821	0.645	0.806	0.438	0.794	0.897	0.812	0.850
Knee Valgus Angle	Correlation	0.445	0.066	0.146	.528*	0.338	0.160	-0.162	.579*	0.251	0.445
	Sig. (2-tailed)	0.074	0.800	0.576	0.029	0.184	0.539	0.534	0.015	0.332	0.074
Knee Valgus Moment	Correlation	-.483*	-0.264	-0.202	-.484*	-.561*	-0.349	-0.087	-.500*	-.525*	-.483*
	Sig. (2-tailed)	0.049	0.306	0.437	0.049	0.019	0.169	0.739	0.041	0.031	0.049
Knee Flex Angle	Correlation	0.225	0.128	0.108	-0.153	0.079	0.289	0.390	-0.157	0.192	0.225
	Sig. (2-tailed)	0.385	0.624	0.679	0.558	0.763	0.261	0.122	0.546	0.461	0.385
Knee Flex Moment	Correlation	-0.228	0.053	-0.082	-0.399	0.050	0.060	0.207	-0.426	0.202	-0.228
	Sig. (2-tailed)	0.378	0.840	0.755	0.113	0.850	0.818	0.426	0.088	0.438	0.378
GRFV	Correlation	-.509*	-0.215	-0.478	-.488*	-0.305	-0.397	-0.129	-0.352	-0.104	-.509*
	Sig. (2-tailed)	0.037	0.408	0.052	0.047	0.233	0.115	0.621	0.166	0.691	0.037

*. Correlation is significant at the 0.05 level (2-tailed).

**. Correlation is significant at the 0.01 level (2-tailed).
Highlighted Variables = Non-parametric data

Female Lt. FL											
		Gmax EMG	Gmed EMG	Ext CON.	Ext. Ecc.	Abd. Con.	AbdEcc	Ext CON. / BW	Ext. Ecc. / BW	Abd. Con. / BW	AbdEcc. / BW
Hip Flex Angle	Correlation	-0.230	-0.239	0.413	0.463	0.212	0.065	0.411	0.440	0.065	-0.160
	Sig. (2-tailed)	0.375	0.355	0.100	0.061	0.414	0.804	0.101	0.077	0.803	0.539
Hip Flex Moment	Correlation	0.206	0.051	-0.104	-0.108	0.051	0.349	-0.218	-0.243	0.038	0.247
	Sig. (2-tailed)	0.427	0.847	0.692	0.679	0.845	0.170	0.401	0.346	0.884	0.339
Hip IntRot Angle	Correlation	-0.145	-0.008	-0.309	-0.421	-0.189	0.050	-0.359	-.486*	-0.082	0.231
	Sig. (2-tailed)	0.579	0.975	0.227	0.092	0.469	0.850	0.157	0.048	0.754	0.372
Hip IntRot Moment	Correlation	0.056	-0.035	-0.346	-0.281	-0.275	0.012	-0.193	-0.085	0.006	0.390
	Sig. (2-tailed)	0.832	0.894	0.174	0.274	0.285	0.962	0.459	0.745	0.982	0.122
Hip Add Angle	Correlation	-0.149	0.331	0.028	-0.089	0.187	0.044	-0.037	-0.180	0.159	0.026
	Sig. (2-tailed)	0.567	0.194	0.914	0.734	0.473	0.866	0.887	0.489	0.542	0.920
Hip Add Moment	Correlation	-0.023	-0.306	-0.061	0.059	-0.236	0.098	-0.163	-0.020	-0.322	-0.016
	Sig. (2-tailed)	0.930	0.232	0.815	0.823	0.362	0.709	0.533	0.940	0.208	0.951
Knee Valgus Angle	Correlation	-.568*	-0.344	-0.219	-0.214	-0.176	-0.061	-0.051	-0.033	0.080	0.264
	Sig. (2-tailed)	0.017	0.177	0.398	0.409	0.499	0.815	0.846	0.900	0.762	0.306
Knee Valgus Moment	Correlation	0.327	0.261	-0.031	0.089	0.160	0.059	-0.158	-0.007	0.076	-0.119
	Sig. (2-tailed)	0.200	0.312	0.905	0.733	0.539	0.822	0.545	0.979	0.773	0.649
Knee Flex Angle	Correlation	-0.313	-0.253	0.268	0.382	0.051	-0.035	0.432	.552*	0.149	0.054
	Sig. (2-tailed)	0.222	0.327	0.299	0.131	0.845	0.894	0.083	0.022	0.569	0.838
Knee Flex Moment	Correlation	-0.081	0.037	0.087	0.114	0.075	-0.163	0.226	0.250	0.249	0.012
	Sig. (2-tailed)	0.757	0.887	0.739	0.664	0.774	0.532	0.382	0.333	0.335	0.964
GRFV	Correlation	0.305	0.310	-.484*	-.525*	-0.050	-0.412	-0.459	-0.447	0.101	-0.191
	Sig. (2-tailed)	0.233	0.225	0.049	0.031	0.849	0.101	0.064	0.072	0.699	0.463

*. Correlation is significant at the 0.05 level (2-tailed).

**. Correlation is significant at the 0.01 level (2-tailed).
Highlighted Variables = Non-parametric data

Female Rt. SML											
		Gmax EMG	Gmed EMG	Ext CON.	Ext. Ecc.	Abd. Con.	AbdEcc	Ext CON. / BW	Ext. Ecc. / BW	Abd. Con. / BW	AbdEcc. / BW
Hip Flex Angle	Correlation	0.369	0.218	0.208	0.217	0.184	0.135	0.159	0.167	0.069	0.048
	Sig. (2-tailed)	0.145	0.401	0.424	0.402	0.480	0.606	0.543	0.521	0.791	0.853
Hip Flex Moment	Correlation	-0.159	0.167	-0.136	-0.063	-0.055	0.030	-0.145	-0.025	-0.107	0.058
	Sig. (2-tailed)	0.542	0.523	0.602	0.809	0.835	0.909	0.579	0.924	0.683	0.826
Hip IntRot Angle	Correlation	0.379	-0.022	0.094	-0.098	0.466	0.196	-0.003	-0.245	.498*	0.078
	Sig. (2-tailed)	0.133	0.934	0.720	0.709	0.059	0.452	0.991	0.343	0.042	0.767
Hip IntRot Moment	Correlation	-0.240	0.207	-0.235	-0.254	-0.259	0.150	-0.052	-0.080	-0.065	0.291
	Sig. (2-tailed)	0.354	0.424	0.364	0.325	0.316	0.564	0.842	0.759	0.805	0.257
Hip Add Angle	Correlation	0.113	0.088	0.123	0.118	-0.092	0.033	0.243	0.200	-0.004	0.104
	Sig. (2-tailed)	0.667	0.737	0.638	0.651	0.724	0.899	0.347	0.442	0.989	0.693
Hip Add Moment	Correlation	0.050	0.134	0.025	0.090	0.153	0.131	-0.161	-0.051	-0.001	-0.047
	Sig. (2-tailed)	0.848	0.609	0.925	0.732	0.559	0.616	0.538	0.846	0.998	0.858
Knee Valgus Angle	Correlation	0.281	0.234	0.134	-0.092	.553*	.567*	0.220	-0.087	.664**	0.473
	Sig. (2-tailed)	0.275	0.366	0.608	0.726	0.021	0.018	0.396	0.739	0.004	0.055
Knee Valgus Moment	Correlation	0.010	-0.095	0.070	0.095	-0.105	-0.215	-0.124	-0.049	-0.302	-0.295
	Sig. (2-tailed)	0.971	0.716	0.790	0.716	0.688	0.407	0.636	0.853	0.238	0.251
Knee Flex Angle	Correlation	0.122	0.025	0.065	0.163	-0.121	-0.024	0.106	0.225	-0.174	0.015
	Sig. (2-tailed)	0.641	0.925	0.805	0.531	0.644	0.927	0.686	0.384	0.503	0.955
Knee Flex Moment	Correlation	-0.263	-0.180	0.127	0.129	-0.276	-0.238	0.102	0.127	-0.390	-0.211
	Sig. (2-tailed)	0.307	0.489	0.627	0.622	0.284	0.357	0.698	0.627	0.122	0.417
GRFV	Correlation	-.539*	-0.195	-0.281	-0.132	-.536*	-.514*	-0.286	-0.117	-0.429	-0.415
	Sig. (2-tailed)	0.026	0.453	0.275	0.614	0.026	0.035	0.266	0.656	0.086	0.097

*. Correlation is significant at the 0.05 level (2-tailed).

**. Correlation is significant at the 0.01 level (2-tailed).
Highlighted Variables = Non-parametric data

Female Lt. SML											
		Gmax EMG	Gmed EMG	Ext CON.	Ext. Ecc.	Abd. Con.	AbdEcc	Ext CON. / BW	Ext. Ecc. / BW	Abd. Con. / BW	AbdEcc. / BW
Hip Flex Angle	Correlation	0.154	-0.066	0.196	0.321	0.118	0.216	0.282	0.221	0.025	0.203
	Sig. (2-tailed)	0.554	0.801	0.450	0.209	0.653	0.405	0.273	0.395	0.926	0.434
Hip Flex Moment	Correlation	-0.196	0.012	0.239	0.224	0.148	.641**	0.157	0.135	0.032	0.358
	Sig. (2-tailed)	0.451	0.963	0.355	0.387	0.570	0.006	0.548	0.606	0.903	0.158
Hip IntRot Angle	Correlation	0.074	-0.029	-0.206	-0.280	0.034	0.194	-0.257	-0.373	-0.007	0.235
	Sig. (2-tailed)	0.779	0.911	0.428	0.277	0.896	0.456	0.319	0.141	0.978	0.363
Hip IntRot Moment	Correlation	0.201	0.179	-0.126	-0.050	-0.116	-0.072	-0.194	-0.169	-0.245	-0.321
	Sig. (2-tailed)	0.439	0.492	0.629	0.848	0.656	0.783	0.456	0.516	0.343	0.209
Hip Add Angle	Correlation	-0.223	0.027	0.284	0.400	0.076	0.314	0.319	0.355	-0.078	0.206
	Sig. (2-tailed)	0.390	0.918	0.268	0.112	0.772	0.220	0.213	0.162	0.765	0.428
Hip Add Moment	Correlation	0.064	-0.020	-0.190	-0.293	0.119	-0.158	-0.051	-0.213	0.309	0.191
	Sig. (2-tailed)	0.808	0.940	0.465	0.254	0.649	0.544	0.844	0.411	0.228	0.462
Knee Valgus Angle	Correlation	0.027	-0.275	0.386	0.452	0.234	0.224	0.419	0.392	0.159	0.154
	Sig. (2-tailed)	0.918	0.286	0.126	0.068	0.366	0.387	0.094	0.119	0.541	0.554
Knee Valgus Moment	Correlation	-0.137	-0.402	0.221	0.174	-0.005	-0.039	0.135	0.142	0.012	-0.169
	Sig. (2-tailed)	0.599	0.110	0.395	0.504	0.985	0.881	0.606	0.586	0.963	0.516
Knee Flex Angle	Correlation	-0.157	-0.100	-0.077	-0.224	-0.090	-0.369	-0.066	-0.125	0.154	-0.196
	Sig. (2-tailed)	0.548	0.701	0.768	0.387	0.733	0.145	0.801	0.633	0.554	0.451
Knee Flex Moment	Correlation	0.154	-0.066	0.196	0.321	0.118	0.216	0.282	0.221	0.025	0.203
	Sig. (2-tailed)	0.554	0.801	0.450	0.209	0.653	0.405	0.273	0.395	0.926	0.434
GRFV	Correlation	-0.196	0.012	0.239	0.224	0.148	.641**	0.157	0.135	0.032	0.358
	Sig. (2-tailed)	0.451	0.963	0.355	0.387	0.570	0.006	0.548	0.606	0.903	0.158

*. Correlation is significant at the 0.05 level (2-tailed).

**. Correlation is significant at the 0.01 level (2-tailed).
Highlighted Variables = Non-parametric data

Female Rt. SLL											
		Gmax EMG	Gmed EMG	Ext CON.	Ext. Ecc.	Abd. Con.	AbdEcc	Ext CON. / BW	Ext. Ecc. / BW	Abd. Con. / BW	AbdEcc. / BW
Hip Flex Angle	Correlation	.650**	0.395	0.391	0.384	0.148	0.227	0.473	.549*	0.147	0.150
	Sig. (2-tailed)	0.005	0.117	0.121	0.128	0.570	0.381	0.055	0.022	0.573	0.567
Hip Flex Moment	Correlation	-0.159	0.167	-0.136	-0.063	-0.055	0.030	-0.145	-0.025	-0.107	0.058
	Sig. (2-tailed)	0.542	0.523	0.602	0.809	0.835	0.909	0.579	0.924	0.683	0.826
Hip IntRot Angle	Correlation	0.287	0.083	0.268	0.262	0.321	0.233	0.186	0.105	0.297	0.078
	Sig. (2-tailed)	0.264	0.751	0.298	0.309	0.209	0.368	0.474	0.687	0.248	0.765
Hip IntRot Moment	Correlation	-0.270	-0.108	-0.326	-0.358	-0.194	0.083	-0.213	-0.206	-0.071	0.194
	Sig. (2-tailed)	0.295	0.680	0.202	0.158	0.456	0.750	0.411	0.428	0.786	0.456
Hip Add Angle	Correlation	0.380	0.025	0.221	0.113	0.076	0.204	0.397	0.365	0.152	0.341
	Sig. (2-tailed)	0.133	0.926	0.393	0.666	0.772	0.433	0.115	0.149	0.560	0.181
Hip Add Moment	Correlation	-0.047	-0.022	-.494*	-0.430	0.036	0.386	-.510*	-0.441	0.059	0.240
	Sig. (2-tailed)	0.859	0.933	0.044	0.085	0.892	0.126	0.037	0.076	0.823	0.353
Knee Valgus Angle	Correlation	0.341	0.069	0.306	0.238	.738**	0.262	0.346	0.061	.689**	0.127
	Sig. (2-tailed)	0.181	0.794	0.232	0.358	0.001	0.309	0.174	0.815	0.002	0.626
Knee Valgus Moment	Correlation	-0.154	-0.010	-0.488*	-0.433	-0.246	0.011	-0.588*	-0.341	-0.164	0.125
	Sig. (2-tailed)	0.554	0.970	0.047	0.083	0.340	0.966	0.013	0.181	0.529	0.633
Knee Flex Angle	Correlation	.522*	0.088	0.197	0.273	0.112	0.193	0.350	0.473	0.159	0.169
	Sig. (2-tailed)	0.032	0.736	0.449	0.288	0.670	0.459	0.168	0.055	0.541	0.516
Knee Flex Moment	Correlation	.650**	0.395	0.391	0.384	0.148	0.227	0.473	.549*	0.147	0.150
	Sig. (2-tailed)	0.005	0.117	0.121	0.128	0.570	0.381	0.055	0.022	0.573	0.567
GRFV	Correlation	-.797**	-0.238	-0.358	-0.380	-0.471	-0.405	-0.392	-0.360	-.485*	-0.201
	Sig. (2-tailed)	0.000	0.358	0.159	0.132	0.056	0.107	0.119	0.155	0.048	0.439

*. Correlation is significant at the 0.05 level (2-tailed).

**. Correlation is significant at the 0.01 level (2-tailed).
Highlighted Variables = Non-parametric data

Female Lt. SLL											
		Gmax EMG	Gmed EMG	Ext CON.	Ext. Ecc.	Abd. Con.	AbdEcc	Ext CON. / BW	Ext. Ecc. / BW	Abd. Con. / BW	AbdEcc. / BW
Hip Flex Angle	Correlation	-0.052	-0.049	0.261	0.347	0.083	0.124	0.303	0.393	0.004	0.014
	Sig. (2-tailed)	0.844	0.852	0.312	0.173	0.751	0.636	0.237	0.118	0.989	0.957
Hip Flex Moment	Correlation	-0.090	-0.043	0.067	0.089	0.153	0.228	0.035	0.049	0.093	0.072
	Sig. (2-tailed)	0.733	0.896	0.799	0.735	0.558	0.378	0.893	0.851	0.723	0.784
Hip IntRot Angle	Correlation	-0.166	-0.103	-0.136	-0.226	-0.052	0.228	-0.165	-0.280	0.027	0.348
	Sig. (2-tailed)	0.524	0.694	0.602	0.383	0.842	0.378	0.527	0.277	0.918	0.171
Hip IntRot Moment	Correlation	-0.467	-0.453	-0.061	-0.086	-0.128	-0.181	0.094	0.053	0.055	0.026
	Sig. (2-tailed)	0.059	0.068	0.816	0.743	0.625	0.487	0.718	0.841	0.835	0.921
Hip Add Angle	Correlation	0.022	0.115	0.102	0.058	0.061	-0.113	-0.022	-0.084	-0.076	-0.286
	Sig. (2-tailed)	0.932	0.660	0.697	0.824	0.816	0.666	0.934	0.748	0.772	0.266
Hip Add Moment	Correlation	-0.262	-0.461	0.062	0.161	-0.177	0.321	0.062	0.161	-0.235	0.244
	Sig. (2-tailed)	0.310	0.063	0.813	0.536	0.496	0.209	0.814	0.536	0.364	0.346
Knee Valgus Angle	Correlation	-0.118	0.150	-0.232	0.137	-0.043	0.186	0.168	0.190	-0.055	.306
	Sig. (2-tailed)	0.653	0.567	0.371	0.601	0.869	0.474	0.520	0.466	0.835	0.538
Knee Valgus Moment	Correlation	0.118	-0.289	0.034	-0.300	-0.019	-.143	-0.328	-0.269	0.086	-0.231
	Sig. (2-tailed)	0.653	0.260	0.896	0.242	0.943	0.583	0.198	0.297	0.744	0.431
Knee Flex Angle	Correlation	-0.091	-0.216	0.227	0.344	0.025	0.189	0.343	0.470	0.055	0.223
	Sig. (2-tailed)	0.728	0.406	0.380	0.177	0.925	0.468	0.177	0.057	0.834	0.390
Knee Flex Moment	Correlation	-0.108	-0.223	-0.014	0.037	-0.159	-0.482	0.019	0.083	-0.105	-0.442
	Sig. (2-tailed)	0.680	0.390	0.958	0.888	0.541	0.050	0.943	0.752	0.687	0.075
GRFV	Correlation	-0.022	0.218	-0.427	-0.428	-0.215	-0.358	-.518*	-0.473	-0.203	-0.350
	Sig. (2-tailed)	0.933	0.400	0.087	0.086	0.408	0.159	0.033	0.055	0.435	0.168

*. Correlation is significant at the 0.05 level (2-tailed).

**. Correlation is significant at the 0.01 level (2-tailed).
Highlighted Variables = Non-parametric data

Male Rt. SLS											
		Gmax EMG	Gmed EMG	Ext CON.	Ext. Ecc.	Abd. Con.	AbdEcc	Ext CON. / BW	Ext. Ecc. / BW	Abd. Con. / BW	AbdEcc. / BW
Hip Flex Angle	Correlation	-0.269	-0.288	-0.130	0.104	0.153	0.316	-0.215	0.076	0.133	0.322
	Sig. (2-tailed)	0.297	0.263	0.620	0.690	0.557	0.216	0.408	0.772	0.612	0.207
Hip Flex Moment	Correlation	.533*	0.235	-0.122	-0.166	-0.210	-0.344	0.238	0.159	0.071	-0.119
	Sig. (2-tailed)	0.028	0.365	0.641	0.525	0.419	0.177	0.357	0.541	0.785	0.651
Hip IntRot Angle	Correlation	0.445	0.203	0.300	0.170	-0.013	0.204	.514*	0.336	0.055	0.264
	Sig. (2-tailed)	0.074	0.434	0.241	0.515	0.959	0.432	0.035	0.188	0.833	0.305
Hip IntRot Moment	Correlation	0.035	0.160	-0.095	-0.266	-0.285	-0.343	0.117	-0.097	-0.117	-0.198
	Sig. (2-tailed)	0.893	0.538	0.717	0.303	0.267	0.178	0.654	0.710	0.655	0.447
Hip Add Angle	Correlation	0.384	0.145	.647**	.625**	.488*	.605*	.689**	.631**	0.385	.521*
	Sig. (2-tailed)	0.128	0.578	0.005	0.007	0.047	0.010	0.002	0.007	0.127	0.032
Hip Add Moment	Correlation	-0.049	-0.074	-0.396	-0.357	-0.453	-.545*	-0.290	-0.224	-0.300	-0.408
	Sig. (2-tailed)	0.852	0.777	0.115	0.159	0.068	0.024	0.259	0.388	0.243	0.104
Knee Valgus Angle	Correlation	-0.106	-0.659**	-0.302	-0.245	-0.257	-0.044	-0.188	-0.106	-0.124	0.085
	Sig. (2-tailed)	0.686	0.004	0.239	0.343	0.319	0.867	0.470	0.684	0.637	0.746
Knee Valgus Moment	Correlation	-0.403	-0.065	-0.269	-0.270	-0.117	-0.233	-0.464	-0.441	-0.160	-0.261
	Sig. (2-tailed)	0.109	0.805	0.296	0.295	0.655	0.367	0.060	0.076	0.541	0.311
Knee Flex Angle	Correlation	-0.308	-0.095	-.630**	-0.476	-0.197	-0.183	-0.362	-0.198	0.180	0.161
	Sig. (2-tailed)	0.229	0.718	0.007	0.053	0.449	0.482	0.153	0.445	0.489	0.537
Knee Flex Moment	Correlation	-0.109	-0.054	-.555*	-0.259	-0.301	-0.376	-0.456	-0.098	-0.107	-0.213
	Sig. (2-tailed)	0.678	0.837	0.021	0.315	0.240	0.137	0.066	0.707	0.684	0.413
GRFV	Correlation	-0.158	-0.080	-0.067	0.076	0.034	-0.041	-0.366	-0.169	-0.160	-0.219
	Sig. (2-tailed)	0.544	0.760	0.800	0.772	0.896	0.874	0.149	0.518	0.540	0.399

*. Correlation is significant at the 0.05 level (2-tailed).

**. Correlation is significant at the 0.01 level (2-tailed).
Highlighted Variables = Non-parametric data

Male Lt. SLS											
		Gmax EMG	Gmed EMG	Ext CON.	Ext. Ecc.	Abd. Con.	AbdEcc	Ext CON. / BW	Ext. Ecc. / BW	Abd. Con. / BW	AbdEcc. / BW
Hip Flex Angle	Correlation	-0.433	-0.390	0.014	0.174	0.438	0.235	-0.066	0.162	0.356	0.169
	Sig. (2-tailed)	0.082	0.122	0.957	0.505	0.078	0.364	0.802	0.535	0.161	0.516
Hip Flex Moment	Correlation	0.148	0.327	-0.420	-0.365	-0.194	-0.136	-0.057	-0.002	0.078	0.217
	Sig. (2-tailed)	0.570	0.200	0.094	0.149	0.455	0.602	0.827	0.993	0.767	0.402
Hip IntRot Angle	Correlation	0.175	-0.381	0.261	0.334	-0.416	-0.174	0.268	0.335	-0.453	-0.251
	Sig. (2-tailed)	0.502	0.131	0.312	0.190	0.097	0.503	0.298	0.188	0.068	0.331
Hip IntRot Moment	Correlation	0.336	0.286	-0.214	-0.178	-0.420	-0.172	-0.008	0.007	-0.226	0.039
	Sig. (2-tailed)	0.187	0.266	0.409	0.493	0.093	0.510	0.976	0.978	0.384	0.883
Hip Add Angle	Correlation	-0.140	-.560*	-0.148	0.056	-0.001	0.061	0.029	0.287	0.089	0.187
	Sig. (2-tailed)	0.591	0.019	0.569	0.832	0.998	0.816	0.911	0.264	0.733	0.472
Hip Add Moment	Correlation	0.183	.621**	0.043	-0.022	-0.190	-0.086	0.066	-0.033	-0.168	-0.076
	Sig. (2-tailed)	0.483	0.008	0.871	0.934	0.466	0.742	0.801	0.899	0.519	0.772
Knee Valgus Angle	Correlation	.492*	-0.032	-0.019	0.058	-0.210	0.135	0.101	0.185	-0.131	0.209
	Sig. (2-tailed)	0.045	0.902	0.942	0.826	0.418	0.605	0.700	0.477	0.616	0.420
Knee Valgus Moment	Correlation	-0.119	0.431	-0.025	-0.107	0.132	-0.017	-0.095	-0.193	0.094	-0.056
	Sig. (2-tailed)	0.650	0.084	0.925	0.683	0.615	0.947	0.718	0.458	0.720	0.832
Knee Flex Angle	Correlation	-0.384	-0.050	-0.435	-0.135	0.054	-0.132	-0.191	0.200	0.265	0.175
	Sig. (2-tailed)	0.128	0.850	0.081	0.607	0.836	0.614	0.462	0.442	0.303	0.502
Knee Flex Moment	Correlation	-0.449	0.179	-0.413	-0.249	0.184	-0.096	-0.228	0.003	0.327	0.138
	Sig. (2-tailed)	0.071	0.493	0.100	0.335	0.480	0.715	0.378	0.989	0.201	0.598
GRFV	Correlation	-0.327	0.022	0.180	0.131	0.263	0.036	-0.077	-0.115	0.099	-0.156
	Sig. (2-tailed)	0.200	0.935	0.489	0.618	0.308	0.891	0.770	0.661	0.707	0.549

*. Correlation is significant at the 0.05 level (2-tailed).

**. Correlation is significant at the 0.01 level (2-tailed).
Highlighted Variables = Non-parametric data

Male Rt. FL											
		Gmax EMG	Gmed EMG	Ext CON.	Ext. Ecc.	Abd. Con.	AbdEcc	Ext CON. / BW	Ext. Ecc. / BW	Abd. Con. / BW	AbdEcc. / BW
Hip Flex Angle	Correlation	0.191	0.108	0.136	0.340	-0.023	0.222	0.347	.567*	0.076	0.316
	Sig. (2-tailed)	0.464	0.680	0.604	0.182	0.931	0.392	0.172	0.018	0.771	0.217
Hip Flex Moment	Correlation	0.335	0.228	-0.070	-0.077	-0.006	-0.115	-0.252	-0.219	-0.137	-0.226
	Sig. (2-tailed)	0.189	0.379	0.790	0.768	0.983	0.661	0.328	0.398	0.600	0.383
Hip IntRot Angle	Correlation	-0.372	-0.218	0.208	0.186	-0.059	0.137	0.386	0.352	-0.014	0.175
	Sig. (2-tailed)	0.142	0.401	0.424	0.475	0.822	0.601	0.126	0.165	0.957	0.501
Hip IntRot Moment	Correlation	0.477	0.093	-0.222	-0.203	-0.057	-0.133	-0.401	-0.338	-0.131	-0.187
	Sig. (2-tailed)	0.053	0.722	0.391	0.436	0.829	0.611	0.111	0.185	0.616	0.472
Hip Add Angle	Correlation	-0.208	-0.058	.603*	.525*	0.202	0.148	.793**	.660**	0.188	0.121
	Sig. (2-tailed)	0.423	0.826	0.010	0.031	0.437	0.572	0.000	0.004	0.469	0.643
Hip Add Moment	Correlation	0.379	0.351	-0.076	0.052	-0.050	0.123	-0.172	-0.003	-0.133	0.066
	Sig. (2-tailed)	0.134	0.167	0.771	0.842	0.848	0.638	0.509	0.990	0.612	0.802
Knee Valgus Angle	Correlation	-0.262	-0.178	-0.181	-0.155	-0.255	-0.081	-0.022	0.015	-0.127	0.028
	Sig. (2-tailed)	0.310	0.494	0.487	0.551	0.324	0.758	0.933	0.955	0.627	0.917
Knee Valgus Moment	Correlation	0.403	0.309	-0.056	0.038	-0.166	-0.303	-0.310	-0.167	-0.348	-0.465
	Sig. (2-tailed)	0.109	0.227	0.830	0.886	0.523	0.238	0.225	0.521	0.172	0.060
Knee Flex Angle	Correlation	0.148	-0.206	-0.001	0.007	-0.004	0.119	0.256	0.237	0.188	0.295
	Sig. (2-tailed)	0.571	0.428	0.996	0.980	0.987	0.649	0.322	0.361	0.470	0.251
Knee Flex Moment	Correlation	-0.267	-0.035	-0.316	-0.267	-0.287	-.599*	-0.079	-0.023	-0.038	-0.398
	Sig. (2-tailed)	0.300	0.894	0.217	0.300	0.263	0.011	0.764	0.930	0.886	0.113
GRFV	Correlation	-0.438	-0.098	-0.239	-0.418	-0.380	-.623**	-0.297	-.497*	-0.367	-.629**
	Sig. (2-tailed)	0.078	0.710	0.356	0.095	0.133	0.008	0.246	0.042	0.147	0.007

*. Correlation is significant at the 0.05 level (2-tailed).

**. Correlation is significant at the 0.01 level (2-tailed).
Highlighted Variables = Non-parametric data

Male Lt. FL											
		Gmax EMG	Gmed EMG	Ext CON.	Ext. Ecc.	Abd. Con.	AbdEcc	Ext CON. / BW	Ext. Ecc. / BW	Abd. Con. / BW	AbdEcc. / BW
Hip Flex Angle	Correlation	0.378	0.163	0.113	0.144	0.184	0.150	0.350	0.353	0.277	0.291
	Sig. (2-tailed)	0.135	0.531	0.666	0.582	0.481	0.565	0.169	0.165	0.282	0.257
Hip Flex Moment	Correlation	-0.478	0.016	0.165	0.327	-0.028	0.230	0.002	0.223	-0.119	0.096
	Sig. (2-tailed)	0.052	0.952	0.528	0.200	0.915	0.374	0.994	0.390	0.648	0.714
Hip IntRot Angle	Correlation	0.113	-0.117	0.295	0.317	-0.331	-0.116	0.440	0.421	-0.307	-0.108
	Sig. (2-tailed)	0.665	0.654	0.250	0.215	0.194	0.657	0.077	0.092	0.231	0.681
Hip IntRot Moment	Correlation	-0.214	0.367	-0.065	0.079	0.099	0.174	-0.136	0.055	0.116	0.190
	Sig. (2-tailed)	0.410	0.147	0.805	0.763	0.707	0.503	0.603	0.835	0.658	0.466
Hip Add Angle	Correlation	0.455	0.122	-0.397	-0.247	-0.193	-0.341	-0.189	0.008	-0.052	-0.137
	Sig. (2-tailed)	0.066	0.640	0.114	0.339	0.458	0.181	0.466	0.975	0.842	0.599
Hip Add Moment	Correlation	-0.463	0.008	0.399	0.398	0.277	.550*	0.430	0.406	0.233	.489*
	Sig. (2-tailed)	0.061	0.977	0.113	0.114	0.281	0.022	0.085	0.106	0.368	0.046
Knee Valgus Angle	Correlation	-0.347	-0.115	0.133	0.191	-0.179	0.225	0.179	0.240	-0.189	0.183
	Sig. (2-tailed)	0.172	0.659	0.611	0.464	0.492	0.385	0.492	0.354	0.466	0.481
Knee Valgus Moment	Correlation	-0.068	0.315	0.138	0.142	0.197	0.169	-0.006	0.019	0.112	0.070
	Sig. (2-tailed)	0.796	0.217	0.597	0.587	0.448	0.516	0.982	0.943	0.668	0.789
Knee Flex Angle	Correlation	0.363	0.078	-0.126	-0.013	0.111	0.028	0.171	0.286	0.303	0.296
	Sig. (2-tailed)	0.152	0.767	0.629	0.962	0.671	0.915	0.512	0.267	0.237	0.248
Knee Flex Moment	Correlation	-0.002	-0.154	-.600*	-0.329	-0.423	-0.409	-.515*	-0.133	-0.246	-0.181
	Sig. (2-tailed)	0.994	0.556	0.011	0.198	0.091	0.103	0.035	0.611	0.341	0.487
GRFV	Correlation	0.042	-0.250	-0.318	-0.252	-.724**	-.780**	-0.469	-0.352	-.724**	-.805**
	Sig. (2-tailed)	0.872	0.333	0.214	0.329	0.001	0.000	0.058	0.166	0.001	0.000

*. Correlation is significant at the 0.05 level (2-tailed).

**. Correlation is significant at the 0.01 level (2-tailed).
Highlighted Variables = Non-parametric data

Male Rt. SML											
		Gmax EMG	Gmed EMG	Ext CON.	Ext. Ecc.	Abd. Con.	AbdEcc	Ext CON. / BW	Ext. Ecc. / BW	Abd. Con. / BW	AbdEcc. / BW
Hip Flex Angle	Correlation	0.512	0.260	0.257	0.335	0.159	0.290	0.399	0.474	0.185	0.318
	Sig. (2-tailed)	0.032	0.314	0.319	0.189	0.542	0.258	0.112	0.054	0.476	0.213
Hip Flex Moment	Correlation	0.262	0.362	-0.044	0.057	-0.188	-.643**	-0.136	0.006	-0.239	-.718**
	Sig. (2-tailed)	0.309	0.153	0.868	0.828	0.469	0.005	0.603	0.982	0.355	0.001
Hip IntRot Angle	Correlation	0.120	-0.047	0.065	0.048	-0.095	0.216	0.277	0.247	0.017	0.328
	Sig. (2-tailed)	0.647	0.857	0.803	0.855	0.717	0.405	0.281	0.340	0.948	0.198
Hip IntRot Moment	Correlation	0.491	-0.133	-0.468	-.549*	-0.386	-.590*	-0.366	-0.453	-0.180	-0.416
	Sig. (2-tailed)	0.041	0.611	0.058	0.022	0.125	0.013	0.149	0.068	0.489	0.097
Hip Add Angle	Correlation	0.202	.556*	.568*	.645**	0.201	0.118	.671**	.742**	0.142	0.057
	Sig. (2-tailed)	0.436	0.021	0.017	0.005	0.440	0.652	0.003	0.001	0.588	0.828
Hip Add Moment	Correlation	0.148	0.257	-0.129	-0.197	-0.056	-0.260	-0.061	-0.169	0.038	-0.189
	Sig. (2-tailed)	0.571	0.319	0.623	0.448	0.831	0.314	0.815	0.517	0.885	0.468
Knee Valgus Angle	Correlation	0.284	-0.253	-0.103	-0.155	-0.181	0.002	0.033	-0.031	-0.075	0.093
	Sig. (2-tailed)	0.269	0.328	0.694	0.553	0.486	0.993	0.901	0.906	0.775	0.723
Knee Valgus Moment	Correlation	-0.136	0.209	-0.269	-0.197	-0.071	-0.194	-0.322	-0.221	-0.045	-0.162
	Sig. (2-tailed)	0.603	0.421	0.296	0.448	0.787	0.456	0.208	0.394	0.863	0.535
Knee Flex Angle	Correlation	.656*	0.267	0.013	0.060	-0.162	-0.123	0.248	0.288	-0.008	0.007
	Sig. (2-tailed)	0.012	0.299	0.961	0.819	0.534	0.639	0.337	0.263	0.976	0.978
Knee Flex Moment	Correlation	-0.002	0.252	-0.257	-0.165	-0.450	-.699**	-0.106	0.017	-0.301	-.590*
	Sig. (2-tailed)	0.995	0.328	0.320	0.527	0.070	0.002	0.686	0.949	0.240	0.013
GRFV	Correlation	-.388	-.548*	-0.526	-0.316	-0.083	-0.145	-0.177	-0.344	-0.015	-0.081
	Sig. (2-tailed)	0.147	0.024	0.049	0.216	0.753	0.579	0.497	0.176	0.956	0.757

*. Correlation is significant at the 0.05 level (2-tailed).

**. Correlation is significant at the 0.01 level (2-tailed).
Highlighted Variables = Non-parametric data

Male Lt. SML											
		Gmax EMG	Gmed EMG	Ext CON.	Ext. Ecc.	Abd. Con.	AbdEcc	Ext CON. / BW	Ext. Ecc. / BW	Abd. Con. / BW	AbdEcc. / BW
Hip Flex Angle	Correlation	.763**	0.290	0.087	0.192	0.001	-0.052	0.213	0.334	0.022	-0.003
	Sig. (2-tailed)	0.000	0.259	0.741	0.461	0.997	0.842	0.412	0.191	0.934	0.989
Hip Flex Moment	Correlation	0.130	-0.047	-.667**	-.507*	-0.322	-.537*	-.591*	-0.364	-0.110	-0.266
	Sig. (2-tailed)	0.619	0.859	0.003	0.038	0.208	0.026	0.012	0.151	0.673	0.303
Hip IntRot Angle	Correlation	0.422	0.245	0.152	0.170	-0.353	-0.286	0.249	0.246	-0.347	-0.289
	Sig. (2-tailed)	0.091	0.343	0.561	0.515	0.164	0.266	0.334	0.342	0.173	0.260
Hip IntRot Moment	Correlation	0.395	0.084	-0.019	0.210	-0.001	0.114	0.011	0.290	0.047	0.166
	Sig. (2-tailed)	0.117	0.748	0.942	0.418	0.995	0.664	0.968	0.259	0.859	0.523
Hip Add Angle	Correlation	.139	.611**	-0.229	-0.048	-0.343	-0.474	-0.153	0.079	-0.270	-0.374
	Sig. (2-tailed)	0.539	0.009	0.376	0.856	0.177	0.055	0.558	0.764	0.295	0.139
Hip Add Moment	Correlation	-0.019	0.045	0.185	0.267	0.332	0.324	0.111	0.233	0.221	0.206
	Sig. (2-tailed)	0.943	0.865	0.476	0.300	0.193	0.205	0.670	0.367	0.394	0.427
Knee Valgus Angle	Correlation	0.203	0.189	0.276	0.362	-0.125	0.243	0.209	0.314	-0.213	0.107
	Sig. (2-tailed)	0.435	0.467	0.283	0.153	0.632	0.347	0.422	0.220	0.412	0.683
Knee Valgus Moment	Correlation	0.073	-0.180	-0.334	-0.293	-0.354	-.570*	-0.145	-0.091	-0.200	-0.357
	Sig. (2-tailed)	0.782	0.488	0.190	0.254	0.164	0.017	0.578	0.729	0.441	0.159
Knee Flex Angle	Correlation	.841**	0.258	-0.079	0.085	0.008	-0.136	0.126	0.317	0.114	0.027
	Sig. (2-tailed)	0.000	0.317	0.762	0.747	0.975	0.603	0.629	0.215	0.664	0.917
Knee Flex Moment	Correlation	-0.314	-.645**	-0.414	-0.315	-0.287	-0.394	-0.325	-0.170	-0.140	-0.193
	Sig. (2-tailed)	0.219	0.005	0.098	0.218	0.263	0.118	0.203	0.513	0.591	0.458
GRFV	Correlation	-.676**	-0.210	0.155	-0.056	-0.103	0.116	0.058	-0.220	-0.140	0.031
	Sig. (2-tailed)	0.003	0.418	0.552	0.831	0.695	0.657	0.824	0.396	0.592	0.905

*. Correlation is significant at the 0.05 level (2-tailed).

**. Correlation is significant at the 0.01 level (2-tailed).
Highlighted Variables = Non-parametric data

Male Rt. SLL											
		Gmax EMG	Gmed EMG	Ext CON.	Ext. Ecc.	Abd. Con.	AbdEcc	Ext CON. / BW	Ext. Ecc. / BW	Abd. Con. / BW	AbdEcc. / BW
Hip Flex Angle	Correlation	.508*	0.064	0.256	0.309	0.314	0.439	0.354	0.390	0.319	0.454
	Sig. (2-tailed)	0.037	0.808	0.321	0.227	0.220	0.078	0.163	0.122	0.212	0.067
Hip Flex Moment	Correlation	-0.100	0.244	-0.408	-0.281	-0.139	-.499*	-0.278	-0.121	0.063	-0.330
	Sig. (2-tailed)	0.704	0.345	0.104	0.274	0.596	0.041	0.281	0.643	0.810	0.195
Hip IntRot Angle	Correlation	-0.060	-0.246	0.118	0.070	-0.067	0.086	0.374	0.299	0.054	0.190
	Sig. (2-tailed)	0.818	0.341	0.651	0.791	0.800	0.744	0.140	0.243	0.838	0.466
Hip IntRot Moment	Correlation	0.289	0.205	-0.427	-0.336	-0.453	-.581*	-0.424	-0.274	-0.373	-.515*
	Sig. (2-tailed)	0.260	0.430	0.087	0.187	0.068	0.015	0.090	0.288	0.141	0.034
Hip Add Angle	Correlation	0.448	0.151	.550*	.643**	0.060	0.196	.655**	.743**	-0.011	0.124
	Sig. (2-tailed)	0.071	0.562	0.022	0.005	0.820	0.451	0.004	0.001	0.967	0.635
Hip Add Moment	Correlation	0.116	0.329	0.476	.540*	-0.128	-0.273	0.342	0.433	-0.388	-.532*
	Sig. (2-tailed)	0.659	0.198	0.053	0.025	0.625	0.288	0.179	0.082	0.123	0.028
Knee Valgus Angle	Correlation	0.298	-0.242	-0.037	-0.050	-0.097	0.097	0.034	0.020	-0.062	0.127
	Sig. (2-tailed)	0.246	0.350	0.888	0.849	0.710	0.710	0.898	0.940	0.814	0.628
Knee Valgus Moment	Correlation	-0.487	0.095	-0.293	-0.235	-0.080	-0.064	-0.376	-0.284	-0.080	-0.050
	Sig. (2-tailed)	0.042	0.717	0.253	0.364	0.761	0.809	0.137	0.269	0.761	0.848
Knee Flex Angle	Correlation	.521*	-0.001	0.183	0.135	0.096	0.190	0.356	0.271	0.175	0.264
	Sig. (2-tailed)	0.032	0.998	0.482	0.606	0.714	0.464	0.160	0.292	0.501	0.307
Knee Flex Moment	Correlation	-0.339	-0.305	-.539*	-.575*	-0.098	-0.089	-0.468	-.501*	0.106	0.110
	Sig. (2-tailed)	0.183	0.234	0.026	0.016	0.709	0.734	0.058	0.041	0.684	0.674
GRFV	Correlation	-.659**	-0.309	-0.189	-0.329	-0.443	-.540*	-0.236	-0.388	-0.443	-.556*
	Sig. (2-tailed)	0.042	0.227	0.468	0.198	0.075	0.025	0.361	0.124	0.075	0.020

*. Correlation is significant at the 0.05 level (2-tailed).

**. Correlation is significant at the 0.01 level (2-tailed).
Highlighted Variables = Non-parametric data

Male Lt. SLL											
		Gmax EMG	Gmed EMG	Ext CON.	Ext. Ecc.	Abd. Con.	AbdEcc	Ext CON. / BW	Ext. Ecc. / BW	Abd. Con. / BW	AbdEcc. / BW
Hip Flex Angle	Correlation	0.288	0.393	0.125	0.112	0.315	0.274	0.167	0.177	0.288	0.393
	Sig. (2-tailed)	0.262	0.119	0.633	0.670	0.218	0.287	0.523	0.497	0.262	0.119
Hip Flex Moment	Correlation	-0.178	0.270	-0.159	-.533*	-.495*	-0.312	-0.123	-0.469	-0.178	0.270
	Sig. (2-tailed)	0.495	0.294	0.543	0.028	0.043	0.223	0.639	0.057	0.495	0.294
Hip IntRot Angle	Correlation	.558*	0.105	-0.304	-0.180	0.181	0.167	-0.210	-0.062	.558*	0.105
	Sig. (2-tailed)	0.020	0.689	0.235	0.489	0.488	0.521	0.418	0.812	0.020	0.689
Hip IntRot Moment	Correlation	-0.023	0.194	-0.244	-0.289	-0.415	-0.169	-0.130	-0.149	-0.023	0.194
	Sig. (2-tailed)	0.929	0.455	0.345	0.260	0.097	0.518	0.620	0.568	0.929	0.455
Hip Add Angle	Correlation	-0.064	-0.023	-0.379	-.490*	-0.027	0.003	-0.338	-0.442	-0.064	-0.023
	Sig. (2-tailed)	0.806	0.929	0.133	0.046	0.918	0.992	0.185	0.075	0.806	0.929
Hip Add Moment	Correlation	0.045	-0.233	0.401	.523*	0.287	-0.090	0.426	.563*	0.045	-0.233
	Sig. (2-tailed)	0.863	0.367	0.111	0.031	0.265	0.730	0.088	0.019	0.863	0.367
Knee Valgus Angle	Correlation	0.176	0.171	-0.190	0.168	0.043	0.142	-0.122	0.242	0.176	0.171
	Sig. (2-tailed)	0.498	0.511	0.464	0.518	0.870	0.586	0.641	0.350	0.498	0.511
Knee Valgus Moment	Correlation	-0.048	-0.299	0.431	0.168	0.164	-0.134	0.392	0.130	-0.048	-0.299
	Sig. (2-tailed)	0.856	0.243	0.084	0.520	0.529	0.607	0.120	0.618	0.856	0.243
Knee Flex Angle	Correlation	0.357	0.253	0.081	-0.008	0.245	0.393	0.202	0.161	0.357	0.253
	Sig. (2-tailed)	0.160	0.326	0.758	0.977	0.343	0.119	0.437	0.537	0.160	0.326
Knee Flex Moment	Correlation	-0.116	-0.186	-0.260	-0.352	-0.113	0.257	-0.203	-0.283	-0.116	-0.186
	Sig. (2-tailed)	0.657	0.475	0.313	0.166	0.665	0.319	0.434	0.271	0.657	0.475
GRFV	Correlation	-.508*	-.489*	-0.476	-0.335	-0.231	-0.284	-0.347	-0.205	-.508*	-.489*
	Sig. (2-tailed)	0.037	0.046	0.054	0.188	0.373	0.269	0.173	0.430	0.037	0.046

*. Correlation is significant at the 0.05 level (2-tailed).

**. Correlation is significant at the 0.01 level (2-tailed).
Highlighted Variables = Non-parametric data

Appendices 6

Single Leg Squat (Intraclass Correlations (ICC), Confidence Intervals (CI), Mean, and SEM):

Variables	ICC (95% CI)	Mean	SEM	CV%
Right				
Joint angles(°)				
Hip Flex	0.87 (0.550 - 0.96)	69.91	3.84	5.94
Hip Int. Rotation	0.97 (0.89 – 0.99)	12.00	1.20	10
Hip Adduction	0.98 (0.917 - .095)	17.69	1.12	6.33
Knee abduction	0.97 (0.875 – 0.99)	-0.94	0.14	14.83
Knee Flexion	0.85 (0.495 – 0.95)	82.92	2.83	3.41
Moments (Nm/Kg)				
Hip Flex	0.79 (0.366 – 0.94)	-1.06	0.17	16.03
Hip Int. Rotation	0.81 (0.415 – 0.95)	-0.43	0.06	13.95
Hip Adduction	0.81 (0.406 – 0.94)	-1.08	0.11	10.18
Knee Abduction	0.77 (0.401 – 0.93)	-0.10	0.04	40
Knee flexion	0.95 (0.813 – 0.97)	1.67	0.07	4.19
VGRF (*bw)	0.90 (0.750 – 0.97)	1.12	0.02	1.17
Left				
Joint angles(°)				
Hip Flexion Angle	0.93 (0.756 – 0.98)	70.02	2.31	3.29
Hip Int. Rotation	0.99 (0.964 – 0.99)	13.18	1.67	5.08
Hip Adduction	0.93 (0.743 – 0.98)	14.98	1.86	12.41
Knee Abduction	0.98 (0.906 – 0.99)	-1.77	0.38	21.46
Knee Flex	0.95 (0.826 – 0.98)	83.78	1.26	1.50
Moments (Nm/Kg)				
Hip Flexion	0.99 (0.940 – 0.99)	-1.09	0.05	4.58
Hip Int. Rotation	0.65 (0.106 – 0.82)	-0.53	0.06	11.32
Hip Adduction	0.63 (0.366 – 0.89)	-1.11	0.08	7.20
Knee Abduction	0.61 (0.331 – 0.88)	-0.15	0.06	40
Knee flexion	0.97 (0.891 – 0.99)	1.60	0.05	2.95
VGRF (*bw)	0.95 (0.813 – 0.98)	1.11	0.01	0.90

Forward Land (Intraclass Correlations (ICC), Confidence Intervals (CI), Mean, and SEM):

Variables	ICC (95% CI)	Mean	SEM	CV%
Right				
Joint Angles(°)				
Hip Flex	0.78 (0.33 – 0.84)	61.95	3.81	6.15
Hip Int. Rotation	0.99 (0.95 – 0.99)	11.51	2.89	7.73
Hip Adduction	0.79 (0.45 – 0.84)	11.58	2.83	24.4
Knee Abduction	0.91 (0.69 – 0.97)	-1.40	0.73	52.14
Knee Flexion	0.90 (0.74 – 0.97)	70.47	2.79	3.95
Moments (Nm/Kg)				
Hip Flex	0.59 (0.36 – 0.71)	-2.05	0.32	15.60
Hip Int. Rotation	0.52 (0.21 – 0.75)	-0.44	0.08	3.52
Hip Adduction	0.71 (0.48 - .081)	-1.63	0.13	7.97
Knee Abduction	0.81 (0.50 – 0.94)	0.09	0.06	41.23
Knee flexion	0.84 (0.68 – 0.95)	2.86	0.20	6.99
VGRF (*bw)	0.91 (0.76 – 0.97)	3.04	0.13	4.27
Left				
Joint Angles(°)				
Hip Flexion	0.96 (0.86 – 0.99)	56.52	1.86	3.30
Hip Int. Rotation	0.96 (0.85 – 0.99)	14.19	2.38	9.72
Hip Adduction	0.91 (0.69 – 0.97)	8.76	1.22	13.92
Knee Abduction	0.98 (0.93 – 0.99)	-0.79	0.06	7.59
Knee Flex	0.96 (0.84 – 0.99)	64.09	2.59	4.04
Moments (Nm/Kg)				
Hip Flexion	0.55 (0.27 – 0.76)	-1.94	0.48	24.74
Hip Int. Rotation	0.41 (0.23 – 0.67)	-1.18	0.17	14.40
Hip Adduction	0.49 (0.28 – 0.60)	-1.81	0.26	14.36
Knee Abduction	0.61 (0.34 – 0.88)	0.18	0.07	38.33
Knee Flexion	0.84(0.52 – 0.90)	2.69	0.17	6.31
VGRF (*bw)	0.78 (0.43 – 0.94)	3.09	0.19	6.14

Side Medial Land (Intraclass Correlations (ICC), Confidence Intervals (CI), Mean, and SEM):

Variables	ICC (95% CI)	Mean	SEM	CV%
Right				
Joint Angle *(°)*				
Hip Flex	0.74 (0.41 – 0.93)	61.84	3.75	6.06
Hip Int. Rotation	0.93 (0.74 – 0.98)	11.95	2.13	17.82
Hip Adduction	0.91 (0.68 – 0.97)	11.85	1.62	13.67
Knee Abduction	0.95 (0.80 – 0.98)	-1.57	0.61	38.21
Knee Flexion	0.73 (0.42 – 0.92)	69.54	4.24	6.09
Moments (Nm/Kg)				
Hip Flex	0.06 (-0.63 – 0.56)	-1.91	0.60	31.41
Hip Int. Rotation	0.47 (0.18 – 0.73)	-0.79	0.10	12.65
Hip Adduction	0.65 (0.37 – 0.89)	-1.52	0.18	11.84
Knee Abduction	0.64 (0.36 – 0.89)	0.34	0.15	44.11
Knee Flexion	0.89 (0.62 – 0.92)	2.88	0.24	8.33
VGRF (*bw)	0.74 (0.37 – 0.92)	3.05	0.17	5.57
Left				
Joint Angle *(°)*				
Hip Flexion	0.53 (0.20 – 0.85)	58.47	6.28	10.60
Hip Int. Rotation	0.93 (0.74 – 0.98)	16.08	3.79	11.13
Hip Adduction	0.97 (0.87 – 0.99)	10.36	0.93	8.97
Knee Abduction	0.62 (0.23 – 0.89)	-1.34	0.58	43.28
Knee Flexion	0.74 (0.44 – 0.92)	67.01	5.65	8.43
Moments (Nm/Kg)				
Hip Flexion	0.67 (0.21 – 0.90)	-2.18	0.40	18.34
Hip Int. Rotation	0.62 (0.43 – 0.89)	-1.17	0.23	19.6
Hip Adduction	0.48 (0.25 – 0.64)	-1.94	0.30	15.46
Knee Abduction	0.43 (0.17 – 0.75)	0.70	0.18	25.71
Knee Flexion	0.48 (0.16 – 0.74)	2.40	0.37	15.41
VGRF (*bw)	0.48 (0.16 – 0.74)	2.97	0.40	13.46

Side Lateral Land (Intraclass Correlations (ICC), Confidence Intervals (CI), Mean, and SEM):

Variables	ICC (95% CI)	Mean	SEM	CV%
Right				
Joint Angle (°)				
Hip Flex	0.81 (0.41 – 0.95)	58.86	3.88	6.59
Hip Int. Rotation	0.94 (0.77 – 0.98)	11.91	3.93	16.20
Hip Adduction	0.83 (0.46 – 0.95)	9.28	2.57	27.69
Knee Abduction	0.88 (0.59 – 0.97)	-2.38	0.87	36.55
Knee Flexion	0.79 (0.36 – 0.94)	64.04	4.16	6.49
Moments (Nm/Kg)				
Hip Flex	0.20 (0.05 – 0.61)	-2.24	0.42	18.75
Hip Int. Rotation	0.94 (0.76 – 0.98)	-0.85	0.05	5.88
Hip Adduction	0.89 (0.61 – 0.97)	-1.40	0.09	6.42
Knee Abduction	0.87 (0.57 – 0.96)	0.36	0.06	16.66
Knee flexion	0.86 (0.54 – 0.96)	2.65	0.18	6.79
VGRF (*bw)	0.64 (0.26 – 0.89)	3.16	0.24	7.59
Left				
Joint Angle (°)				
Hip Flexion	0.87 (0.57 – 0.96)	53.71	2.77	5.15
Hip Int. Rotation	0.98 (0.90 – 0.99)	14.63	3.18	8.06
Hip Adduction	0.91 (0.67 – 0.97)	6.29	1.59	25.27
Knee Abduction	0.91 (0.69 – 0.97)	-1.45	0.73	50.34
Knee Flex	0.97 (0.86 – 0.99)	63.75	1.69	2.65
Moments (Nm/Kg)				
Hip Flexion	0.15 (0.05 – 0.68)	-2.06	0.51	24.75
Hip Int. Rotation	0.82 (0.43 – 0.95)	-1.12	0.08	7.14
Hip Adduction	0.87 (0.56 – 0.96)	-1.96	0.13	6.63
Knee Abduction	0.50 (0.14 – 0.84)	0.09	0.06	66.66
Knee Flexion	0.97 (0.86 – 0.99)	2.44	0.10	4.09
VGRF (*bw)	0.75 (0.36 – 0.99)	3.13	0.16	5.11

Kinetics and kinematics differences between legs during SLS, FL, SML and SLL:

Variables	Single leg squat				Forward land				Side medial land				Side lateral land			
	Rt.	**Lt.**	MD	**P value**	**Rt.**	**Lt.**	MD	**P value**	**Rt.**	**Lt.**	**MD**	**P value**	**Rt.**	**Lt.**	**MD**	**P valu**
Joint Angle (°)																
Hip Flex	69.74	69.55	.18	.83	57.34	54.84	2.49	.07	56.81	54.07	2.73	.01*	54.30	53.37	.93	.45
Hip Int. Rotation	7.43	9.97	2.54	.03*	7.59	9.86	2.27	.05*	9.10	11.49	2.39	.08	8.50	9.45	.94	.43
Hip Adduction	13.59	13.43	.15	.89	7.70	7.53	.16	.88	9.62	9.78	.16	.89	6.20	5.65	.55	.64
Knee Abduction	-1.14	-.56	.57	.31	-1.60	-1.05	.54	.36	-1.48	-1.72	.23	.73	-2.49	-2.07	.42	.72
Knee Flexion	83.33	82.38	.94	.36	67.97	64.17	3.79	.005*	66.82	63.74	3.08	.01*	66.74	62.99	3.74	.005*
Moments (Nm/Kg)																
Hip Flex	-1.05	1.10	.04	.28	-1.77	-1.76	.01	.92	-1.75	-1.95	.20	.10	-1.90	-1.75	.15	.14
Hip Int. Rotation	-.40	-.50	.10	.005*	-.78	-1.02	.23	.005*	-.71	-1.02	.30	.005*	-.73	-.97	.23	.005*
Hip Adduction	-.98	-1.07	.08	.01*	-1.62	-1.85	.23	.005*	-1.50	-1.93	.43	.005*	-1.53	-1.93	.40	.005*
Knee Abduction	-.05	-.11	.06	.02*	.15	.13	.02	.74	.36	.14	.21	.005*	.31	.12	.19	.005*
Knee Flexion	1.65	1.56	.09	.005*	2.77	2.69	.06	.28	2.76	2.44	.32	.005*	2.66	2.45	.20	.05*
GVRF (*bw)	1.12	1.12	0.00	.44	3.22	3.25	.03	.57	3.25	3.24	.01	.67	3.31	3.27	.04	.66

- * The mean difference is significant at the .05 level.

Kinetics and kinematics differences of right leg between SLS, FL, SML and SLL tasks

Variables	SLS vs. FL	SLS vs. SML	SLS vs. SLL	FL vs. SML	FL vs. SLL	SML vs. SLL
Joints Angles (°)	**MD / P value**	**MD / P value**	**MD / P value**	**MD / P value**	**MD / P value**	**MD / P value**
Hip Flexion	12.40 / 0.005*	12.93/ 0.005*	15.43 / 0.005*	0.52 / 0.98	3.03 / 0.14	2.50 / 0.06
Hip Int. Rotation	0.16 / 0.63	1.67 / 0.01*	1.07 / 0.44	1.51 / 0.01*	0.91 / 0.25	0.59 / 0.80
Hip Adduction	5.89 / 0.005*	3.97 / 0.005*	7.38 / 0.005*	1.91 / 0.01*	1.49/ 0.02*	3.41 / 0.005*
Knee Abduction	0.46 / 0.14	0.34 / 0.60	1.35 / 0.13	0.11 / 0.48	0.88 / 0.02*	1.00 / 0.01*
Knee Flexion	15.35 / 0.005*	16.50 / 0.005*	16.58 / 0.005*	1.14 / 0.66	1.23 / 0.24	0.08 / 0.41
Moments (Nm/kg)						
Hip Flexion	0.72 / 0.005*	0.69 / 0.005*	0.85 / 0.005*	0.02 / 1.00	0.12 / 1.00	0.15 / 1.00
Hip Int. Rotation	0.38 / 0.005*	0.31 / 0.005*	0.33 / 0.005*	0.06 / 0.23	0.04 / 0.77	0.01 / 1.00
Hip Adduction	0.63 / 0.005*	0.51 / 0.005*	0.54 / 0.005*	0.11 / 0.25	0.09 / 1.00	0.02 / 1.00
Knee Abduction	0.21 / 0.005*	0.41 / 0.005*	0.36 / 0.005*	0.20 / 0.005*	0.15 / 0.02*	0.04 / 0.93
Knee Flexion	1.12 / 0.005*	1.10 / 0.005*	1.00 / 0.005*	0.01 / 1.00	0.11 / 0.93	0.10 / 0.70
VGRF (*bw)	2.10 / 0.005*	2.13 / 0.005*	2.19 / 0.005*	0.03 / 0.65	0.09 / 0.12	0.06 / 0.36

- * The mean difference (MD) is significant at the .05 level.
- Significant adjustment for multiple comparisons: Bonferroni.

Kinetics and kinematics differences of left leg between SLS, FL, SML and SLL tasks

Variables	SLS vs. FL	SLS vs. SML	SLS vs. SLL	FL vs. SML	FL vs. SLL	SML vs. SLL
Joints Angles (°)	**MD / P value**	**MD / P value**	**MD / P value**	**MD / P value**	**MD / P value**	**MD / P value**
Hip Flexion	14.70 / 0.005*	15.48 / 0.005*	16.18 / 0.005*	0.77 / 1.00	1.47 / 1.00	0.69 / 1.00
Hip Int. Rotation	0.11 / 1.00	1.51 / 0.43	0.52 / 1.00	1.63 / 0.01*	0.41 / 1.00	2.04 / 0.02*
Hip Adduction	5.89 / 0.005*	3.65 / 0.005*	7.78 / 0.005*	2.24 / 0.001*	1.88 / 0.03*	4.12 / 0.005*
Knee Abduction	0.49 / 1.00	1.15 / 0.19	1.50 / 0.01*	0.66 / 0.66	1.01 / 0.14	0.34 / 1.00
Knee Flexion	18.21 / 0.005*	18.64 / 0.005*	19.38 / 0.005*	0.43 / 0.89	1.17 / 1.00	0.74 / 1.00
Moments (Nm/kg)						
Hip Flexion	0.66 / 0.005*	0.85 / 0.005*	0.65 / 0.005*	0.19 / 1.00	0.008 / 1.00	0.20/ 1.00
Hip Int. Rotation	0.51 / 0.005*	0.51 / 0.005*	0.46 / 0.005*	0.05 / 0.87	0.01 / 1.00	0.05 / 0.30
Hip Adduction	0.77 / 0.005*	0.86 / 0.005*	0.86 / 0.005*	0.08 / 0.93	0.08 / 0.96	0.002 / 1.00
Knee Abduction	0.24 / 0.005*	0.26 / 0.005*	0.23 / 0.005*	0.01 / 1.00	0.01 / 1.00	0.02 / 0.93
Knee Flexion	1.13 / 0.005*	0.88 / 0.005*	0.89 / 0.005*	0.25 / 0.005*	0.23 / 0.01*	0.01 / 1.00
VGRF (*bw)	2.13 / 0.005*	2.12 / 0.005*	2.15 / 0.005*	0.01 / 0,72	0.02 / 0.75	0.03 / 0.47

- * The mean difference is significant at the .05 level.
- Significant adjustment for multiple comparisons: Bonferroni.

Gender differences during SLS task

Variables	Right					Left				
	Females		Males		P value	Females		Males		P value
	Mean	SD	Mean	SD		Mean	SD	Mean	SD	
Joints Angles (°)										
Hip Flexion	70.76	9.99	68.72	10.35	.56	68.51	10.72	70.59	8.83	.54
Hip Int. Rotation	6.61	7.36	8.25	6.48	.49	9.26	7.51	10.69	8.85	.61
Hip Adduction	16.86	8.34	10.31	5.04	.01*	15.44	6.53	11.42	4.05	.03*
Knee Abduction	-3.39	4.54	1.11	4.14	.005*	-2.23	4.27	1.10	2.34	.01*
Knee Flexion	81.46	6.99	85.20	6.72	.21	80.87	6.99	83.89	8.47	.26
Moments (Nm/kg)										
Hip Flexion	-1.13	.42	-.98	.33	.27	-1.06	.37	-1.14	.37	.51
Hip Int. Rotation	-.43	.16	-.38	.13	.34	-.51	.14	-.49	.15	.71
Hip Adduction	-1.03	.25	-.93	,19	.21	-1.07	.23	-1.06	,13	.90
Knee Abduction	.03	.18	-.14	.16	.005*	-.05	.18	-.17	.10	.02*
Knee Flexion	1.55	.27	1.76	.19	.01*	1.50	.27	1.62	.26	.19
VGRF (*bw)	1.12	.10	1.11	.07	.93	1.11	.10	1.12	.06	.78

- The mean difference is significant at the .05 level.

Gender differences during FL task

Variables	Right					Left				
	Females		Males		P value	Females		Males		P value
	Mean	SD	Mean	SD		Mean	SD	Mean	SD	
Joints Angles (°)										
Hip Flexion	56.32	13.79	58.35	11.73	.64	52.58	13.03	57.10	13.50	.32
Hip Int. Rotation	7.22	7.75	7.96	6.76	.77	9.16	7.88	10.56	8.25	.61
Hip Adduction	11.00	6.71	4.40	5.11	.005*	9.37	3.99	5.70	3.35	.005*
Knee Abduction	-4.03	4.92	.82	4.12	.005*	-2.93	3.77	.81	3.97	.005*
Knee Flexion	65.83	11.86	70.12	13.01	.32	62.03	11.72	66.31	16.04	.38
Moments (Nm/kg)										
Hip Flexion	-1.83	.54	-1.72	.66	.59	-1.59	.56	-1.93	.80	.15
Hip Int. Rotation	-.79	.27	-.77	.19	.84	-1.00	.26	-1.04	.32	.64
Hip Adduction	-1.68	.25	-1.55	,32	.20	-1.93	.25	-1.77	,40	.18
Knee Abduction	.28	.25	.03	.17	.005*	.20	.22	.05	.13	.02*
Knee Flexion	2.66	.50	2.89	.41	.15	2.67	.42	2.72	.42	.70
VGRF (*bw)	3.32	.34	3.11	.49	.16	3.36	.43	3.14	.45	.16

- The mean difference is significant at the .05 level.

Gender differences during SML task

Variables	Right					Left				
	Females		Males		P value	Females		Males		P value
	Mean	SD	Mean	SD		Mean	SD	Mean	SD	
Joints Angles (°)										
Hip Flexion	54.83	17.03	58.78	12.15	.44	52.84	13.65	55.29	13.05	.59
Hip Int. Rotation	8.35	8.01	9.85	7.08	.56	10.01	7.36	12.89	9.76	.35
Hip Adduction	12.97	5.49	6.26	4.75	.005*	12.21	4.04	4.04	7.34	.005*
Knee Abduction	-3.68	4.64	.70	4.53	.005*	-3.34	3.96	.93	5.40	.04*
Knee Flexion	63.67	11.84	69.97	10.85	.11	62.70	10.09	64.78	11.58	.58
Moments (Nm/kg)										
Hip Flexion	-1.72	.63	-1.78	.62	.81	-2.06	.67	-1.85	.51	.32
Hip Int. Rotation	-.73	.22	-.70	.18	.64	-1.01	.31	-1.04	.28	.77
Hip Adduction	-1.55	.32	-1.44	.25	.28	-1.99	.38	-1.87	.30	.32
Knee Abduction	.46	.35	.25	.28	.06	.16	.29	.12	.24	.68
Knee Flexion	2.61	.53	2.92	.50	.09	2.33	.58	2.55	.50	.23
VGRF (*bw)	3.49	.49	3.01	.43	.005*	3.47	.58	3.01	.30	.005*

*The mean difference is significant at the .05 level.

Gender differences during SLL task

Variables	Right					Left				
	Females		Males		P value	Females		Males		P value
	Mean	SD	Mean	SD		Mean	SD	Mean	SD	
Joints Angles (°)										
Hip Flexion	52.06	14.44	56.54	12.91	.34	53.20	16.09	53.54	14.31	.94
Hip Int. Rotation	7.66	7.34	9.34	7.56	.51	9.12	7.73	9.78	10.70	.84
Hip Adduction	9.65	5.97	2.76	4.28	.005*	7.75	4.08	3.55	4.94	.01*
Knee Abduction	-4.55	4.21	-.43	5.16	.01*	-3.99	4.84	-.41	4.57	.02*
Knee Flexion	63.53	11.07	69.95	11.82	.11	62.04	11.31	63.94	11.42	.63
Moments (Nm/kg)										
Hip Flexion	-1.76	.59	-2.05	.52	.14	-1.79	.58	-1.72	.47	.69
Hip Int. Rotation	-.67	.19	-.79	.21	.09	-.97	.28	-.96	.20	.91
Hip Adduction	-1.52	.33	-1.53	.38	.99	-2.01	.39	-1.85	.29	.18
Knee Abduction	.37	.31	.25	.23	.22	.16	.20	.07	.12	.11
Knee Flexion	2.40	.47	2.91	.50	.005*	2.45	.46	2.46	.62	.93
VGRF (*bw)	3.48	.48	3.14	.39	.02*	3.45	.39	3.08	.36	.005*

*The mean difference is significant at the .05 level.

Kinetics and kinematics differences between legs during SLS, FL, SML and SLL in female group:

Variables	Single leg squat				Forward land				Side medial land				Side lateral land			
	Rt.	**Lt.**	MD	**P value**	**Rt.**	**Lt.**	MD	**P value**	**Rt.**	**Lt.**	**MD**	**P value**	**Rt.**	**Lt.**	**MD**	**P value**
Joint Angle (°)																
Hip Flex	70.76	68.51	2.24	.04*	56.32	52.58	3.73	.08	54.83	52.84	1.98	.23	52.06	53.20	1.13	.52
Hip Int. Rotation	6.61	9.26	2.64	.16	7.22	9.16	1.94	.30	8.35	10.10	1.74	.39	7.66	9.12	1.46	.46
Hip Adduction	16.86	15.44	1.42	.48	11.00	9.37	1.62	.34	12.97	12.21	.75	.68	9.65	7.75	1.89	.31
Knee Abduction	-3.39	-2.23	1.15	.10	-4.03	-2.93	1.10	.26	-3.68	-3.34	.33	.69	-4.55	-3.99	.56	.57
Knee Flexion	81.46	80.87	.58	.61	65.83	62.03	3.79	.03*	63.67	62.70	.97	.57	63.53	62.04	1.49	.20
Moments (Nm/Kg)																
Hip Flex	-1.13	1.06	.06	.19	-1.83	-1.59	.24	.12	-1.72	-2.06	.33	.84	-1.76	-1.79	.02	.85
Hip Int. Rotation	-.43	-.51	.08	.005*	-.79	-1.00	.20	.005*	-.73	-1.01	.27	.005*	-.67	-.97	.29	.005*
Hip Adduction	-1.03	-1.07	.04	.42	-1.68	-1.93	.24	.005*	-1.55	-1.99	.43	.005*	-1.52	-2.01	.48	.005*
Knee Abduction	-.03	-.05	.09	.03*	.28	.20	.07	.36	.46	.16	.30	.005*	.37	.16	.20	.05*
Knee Flexion	1.55	1.50	.05	.19	2.66	2.67	.01	.92	2.61	2.33	.27	.01*	2.40	2.45	.05	.63
GVRF (*bw)	1.12	1.11	0.01	.84	3.32	3.36	.04	.57	3.49	3.47	.02	.77	3.48	3.45	.03	**.74**

- * The mean difference is significant at the .05 level.

Kinetics and kinematics differences of right leg between SLS, FL, SML and SLL tasks in female

Variables	SLS vs. FL	SLS vs. SML	SLS vs. SLL	FL vs. SML	FL vs. SLL	SML vs. SLL
Joints Angles (°)	**MD / P value**	**MD / P value**	**MD / P value**	**MD / P value**	**MD / P value**	**MD / P value**
Hip Flexion	14.43 / 0.001*	15.92 / 0.001*	18.69 / 0.005*	1.49 / 1.00	4.26 / 0.11	2.76 / 0.32
Hip Int. Rotation	0.60 / 1.00	1.73 / 0.05	1.04 / 0.41	1.12 / 0.38	0.43 / 1.00	0.68 / 0.90
Hip Adduction	5.86 / 0.005*	3.89 / 0.11	7.21 / 0.005*	1.97 / 0.20	1.34 / 0.43	3.32 / 0.001*
Knee Abduction	0.63 / 1.00	0.28 / 1.00	1.15 / 0.61	0.35 / 1.00	0.52 / 0.81	0.87 / 0.27
Knee Flexion	15.63 / 0.005*	17.78 / 0.005*	17.92 / 0.005*	2.15 / 1.00	2.29 / 1.00	0.14 / 1.00
Moments (Nm/kg)						
Hip Flexion	0.70 / 0.005*	0.59 / 0.03*	0.63 / .02*	0.10 / 1.00	0.07 / 1.00	0.03 / 1.00
Hip Int. Rotation	0.36 / 0.005*	0.30 / 0.005*	0.24 / 0.005*	0.05 / 0.88	0.11 / 0.04*	0.05 / 0.34
Hip Adduction	0.65 / 0.005*	0.52 / 0.005*	0.49 / 0.005*	0.12 / 0.19	0.15 / 0.19	0.02 / 1.00
Knee Abduction	0.25 / 0.005*	0.43 / 0.005*	0.33 / 0.005*	0.18 / 0.12	0.08 / 1.00	0.09 / 1.00
Knee Flexion	1.15 / .005*	1.05 / 0.005*	0.84 / 0.005*	0.05 / 1.00	0.26 / .04*	0.21 / .03*
VGRF (*bw)	2.20 / 0.005*	2.37 / 0.005*	2.36 / 0.005*	0.17 / 0.47	0.16 / 0.43	0.01 / 1.00

- * The mean difference is significant at the .05 level.
- Significant adjustment for multiple comparisons: Bonferroni.

Kinetics and kinematics differences of left leg between SLS, FL, SML and SLL tasks in female

Variables	SLS vs. FL	SLS vs. SML	SLS vs. SLL	FL vs. SML	FL vs. SLL	SML vs. SLL
Joints Angles (°)	**MD / P value**	**MD / P value**	**MD / P value**	**MD / P value**	**MD / P value**	**MD / P value**
Hip Flexion	15.92 / 0.005*	15.66 / 0.005*	15.31 / 0.005*	0.25 / 1.00	0.61 / 1.00	0.35 / 1.00
Hip Int. Rotation	0.09 / 1.00	0.83 / 0.76	0.13 / 1.00	0.93 / .26	0.04 / 1.00	0.97 / 0.69
Hip Adduction	6.06 / 0.005*	3.22 / 0.07	7.68 / 0.005*	2.84 / 0.006*	1.61 / 0.52	4.46 / 0.001*
Knee Abduction	0.69 / 1.00	1.11 / 0.33	1.75 / 0.12	0.41 / 1.00	1.06 / 1.00	0.64 / 1.00
Knee Flexion	18.84 / 0.005*	18.17 / 0.005*	18.83 / 0.005*	0.67 / 0.63	0.14 / 1.00	0.66 / 1.00
Moments (Nm/kg)						
Hip Flexion	0.53 / 0.003*	1.00 / 0.005*	0.73 / 0.002*	0.46 / .01*	0.20 / 0.83	0.26 / 0.75
Hip Int. Rotation	0.48 / 0.005*	0.49 / 0.005*	0.45 / 0.005*	0.01 / 1.00	0.02 / 1.00	0.03 / 1.00
Hip Adduction	0.85 / 0.005*	0.91 / 0.005*	0.93 / .005*	0.06 / 1.00	0.08 / 1.00	0.01 / 1.00
Knee Abduction	0.26 / 0.007*	0.22 / .01*	0.22 / .008*	0.04 / 1.00	0.03 / 1.00	0.02 / 0.93
Knee Flexion	1.16 / 0.005*	0.83 / 0.005*	0.94 / 0.005*	0.33 / .01*	0.21 / 0.07	0.11 / 0.90
VGRF (*bw)	2.25 / 0.005*	2.36 / 0.005*	2.34 / 0.005*	0.11 / 0.74	0.09 / 0.89	0.02 / 1.00

- * The mean difference is significant at the .05 level.
- Significant adjustment for multiple comparisons: Bonferroni.

Kinetics and kinematics differences between legs during SLS, FL, SML and SLL in male group

Variables	Single leg squat				Forward land				Side medial land				Side lateral land			
	Rt.	**Lt.**	MD	**P value**	**Rt.**	**Lt.**	MD	**P value**	**Rt.**	**Lt.**	**MD**	**P value**	**Rt.**	**Lt.**	**MD**	**P value**
Joint Angle (°)																
Hip Flex	68.72	70.59	1.23	.18	58.35	57.10	3.73	.50	58.78	55.29	2.73	.02*	56.54	53.54	.93	.09
Hip Int. Rotation	8.25	10.69	2.64	.12	7.96	10.56	2.04	.06	9.85	12.89	2.39	.10	9.34	9.78	.94	.77
Hip Adduction	10.31	11.42	1.09	.43	4.40	5.70	1.62	.35	6.26	7.34	.16	.45	2.76	3.55	.55	.61
Knee Abduction	1.11	1.10	1.15	.99	.82	.81	1.10	.99	.70	-.93	.23	.46	-.43	-.14	.42	.74
Knee Flexion	85.20	83.89	.58	.46	70.12	66.31	3.79	.08	69.97	64.78	5.08	.005*	69.95	63.94	3.74	.01*
Moments (Nm/Kg)																
Hip Flex	-.98	-1.14	.16	.01*	-1.72	-1.93	.01	.39	-1.78	-1.85	.20	.63	-2.05	-1.72	.28	.01*
Hip Int. Rotation	-.38	-.49	.11	.005*	-.77	-1.04	.27	.005*	-.70	-1.04	.34	.005*	-.79	-.96	.17	.005*
Hip Adduction	-.93	-1.06	.13	.01*	-1.55	-1.77	.23	.06	-1.44	-1.87	.43	.005*	-1.53	-1.85	.32	.02*
Knee Abduction	-.14	-.17	.03	.22	.15	.13	.02	.63	.25	.12	.21	.16	.25	.17	.08	.005*
Knee Flexion	1.76	1.62	.14	.005*	2.89	2.72	.06	.14	2.93	2.55	.38	.005*	2.91	2.46	.45	.005*
GVRF (*bw)	1.11	1.12	0.01	.18	3.11	3.14	.03	.78	3.01	3.01	.01	.93	3.14	3.08	.04	.33

- * The mean difference is significant at the .05 level.

Kinetics and kinematics differences of right leg between SLS, FL, SML and SLL tasks in male group

Variables	SLS vs. FL	SLS vs. SML	SLS vs. SLL	FL vs. SML	FL vs. SLL	SML vs. SLL
Joints Angles (°)	**MD / P value**	**MD / P value**	**MD / P value**	**MD / P value**	**MD / P value**	**MD / P value**
Hip Flexion	10.37 / 0.09	9.93 / 0.16	12.17 / 0.02	0.43 / 1.00	1.80 / 1.00	2.24 / 0.65
Hip Int. Rotation	0.28 / 1.00	1.60 / 0.53	1.09 / 1.00	1.89 / 0.09	1.38 / 0.28	0.51 / 1.00
Hip Adduction	5.91 / 0.005*	4.05 / 0.005*	7.55 / 0.005*	1.86 / 0.27	1.64 / 0.20	3.50 / 0.005*
Knee abduction	0.29 / 1.00	0.40 / 1.00	1.54 / 0.67	0.11 / 1.00	1.25 / 0.08	1.14 / 0.11
Knee Flexion	15.08 / 0.005*	15.23 / 0.005*	15.25 / 0.005*	0.14 / 1.00	0.16 / 1.00	0.02 / 1.00
Moments (Nm/kg)						
Hip Flexion	0.74 / 0.005*	0.79 / 0.005*	1.06 / 0.005*	0.05 / 0.90	0.32 / 0.69	0.27 / 0.13
Hip Int. Rotation	0.39 / 0.005*	0.32 / 0.005*	0.41 / 0.005*	0.07 / 1.00	0.01 / 1.00	0.09 / 1.00
Hip Adduction	0.62 / 0.005*	0.51 / 0.005*	0.59 / 0.005*	0.10 / 0.25	0.02 / 1.00	0.03 / 1.00
Knee abduction	0.17 / 0.005*	0.39 / 0.005*	0.39 / 0.005*	0.22 / 0.01*	0.22 / 0.008*	0.004 / 1.00
Knee Flexion	1.13 / 0.005*	1.15 / 0.005*	1.15 / 0.005*	0.02 / 1.00	0.02 / 1.00	0.001 / 1.00
VGRF (*bw)	2.00 / 0.005*	1.90 / .005*	2.03 / 0.005*	0.10 / 1.00	0.03 / 1.00	0.13 / 1.00

- * The mean difference is significant at the .05 level.
- Significant adjustment for multiple comparisons: Bonferroni.

Kinetics and kinematics differences of left leg between SLS, FL, SML and SLL tasks in male group

Variables	SLS vs. FL	SLS vs. SML	SLS vs. SLL	FL vs. SML	FL vs. SLL	SML vs. SLL
Joints Angles (°)	**P value**	**P value**	**P value**	**P value**	**P value**	**P value**
Hip Flexion	13.49 / 0.01*	15.30 / 0.005*	17.05 / 0.005*	1.81 / 1.00	3.56 / 0.34	1.74 / 1.00
Hip Int. Rotation	0.13 / 1.00	2.19 / 1.00	0.91 / 1.00	2.33 / 0.08	0.78 / 1.00	3.11 / 0.10
Hip Adduction	5.72 / 0.005*	4.08 / 0.005*	7.87 / 0.005*	1.64 / 0.18	2.15 / 0.19	3.79 / 0.004*
Knee abduction	0.29 / 1.00	1.19 / 1.00	1.25 / 0.24	0.90 / 1.00	0.96 / 0.18	0.05 / 0.11
Knee Flexion	17.57 / 0.005*	19.11 / 0.005*	19.94 / 0.005*	1.53 / 1.00	2.37 / 1.00	0.83 / 1.00
Moments (Nm/kg)						
Hip Flexion	0.79 / 0.03*	0.70 / 0.005*	0.57 / 0.01*	0.08 / 1.00	0.21 / 1.00	0.13 / 1.00
Hip Int. Rotation	0.55 / 0.005*	0.54 / 0.005*	0.46 / 0.005*	0.009 / 1.00	0.08 / 1.00	0.07 / 1.00
Hip Adduction	0.70 / 0.005*	0.80 / 0.005*	0.78 / 0.005*	0.10 / 1.00	0.07 / 1.00	0.02 / 1.00
Knee abduction	0.23 / 0.005*	0.30 / 0.005*	0.24 / 0.005*	0.07 / 1.00	0.01 / 1.00	0.05 / 1.00
Knee Flexion	1.10 / 0.005*	0.93 / 0.005*	0.84 / 0.005*	016 / 0.57	0.24 / 0.34	0.09 / 1.00
VGRF (*bw)	2.02 / 0.005*	1.89 / 0.005*	1.96 / 0.005*	0.13 / 1.00	0.06 / 1.00	0.07 / 1.00

- * The mean difference is significant at the .05 level.
- Significant adjustment for multiple comparisons: Bonferroni.

Appendices 7

Correlation between kinematics and kinetics with strength and EMG data during right SLS task

Right SLS										
Variables	G Max	G Med	Ext. Con.	Ext. Ecc.	Ext. Con./BW	Ext. Ecc./BW	Abd. Con.	Abd. Ecc	Abd. Con./BW	Abd. Ecc./BW
Joint Angle (°)										
Hip Flex	r= -.36 p= .03 R^2=0.12	ρ= .06 p= .97	ρ= -.03 p= .83	r= .02 p= .87	r= -.07 p= .68	r= .06 p= .73	r= .16 p= .35	r= .19 p= .27	ρ = .22 p= .21	r= .24 p= .16
Hip Int. Rotation	r= .01 p= .95	ρ= -.14 p= .40	ρ= .29 p= .09	r= .06 p= .72	r= .29 p= .09	r= .06 p= .72	r= .26 p= .13	r= .25 p= .15	ρ = .30 p = .07	r= .26 p= .12
Hip Adduction	r= -14 p = .40	ρ= -.03 p= .83	ρ= .09 p= .58	r= -.04 p= .81	r= .12 p= .47	r= .02 p= .89	r= .09 p= .57	r= .11 p= .52	ρ = .17 p = .31	r= .20 p= .25
Knee Abduction	r=-.31 p = .07	ρ= -.41 p= .01	ρ= .15 p= .39	r= .04 p= .79	r= .15 p= .39	r= .05 p= .75	r= 32 p= .058	r = .29 p= .09	ρ = .36 p = .06	r= .30 p= .08
Knee Flexion	r= -.29 p = .90	ρ= .04 p= .82	ρ= -.15 p= .37	r= .03 p= .85	r= .07 p= .68	r= .21 p= .21	r= -.11 p= .52	r= .08 p= .61	ρ = .01 p= .91	r= -.23 p= .17
Moments (Nm/Kg)										
Hip Flex	r= .20 p= .08	ρ= -.11 p= .51	ρ= .26 p= .88	r= .01 p = .93	r= .10 p= .55	r= .08 p= .64	r= -.04 p= .79	r= -.11 p= .52	ρ = .01 p= .91	r= -.08 p= .64
Hip Int. Rotation	r= .02 p = .95	ρ= -.18 p= .29	ρ= -.12 p= .47	r= -.17 p= .45	r= -.05 p= .78	r= -.11 p= .51	r= -.21 p= .23	r= -.08 p= .64	ρ = .16 p = .36	r= -.02 p= .90
Hip Adduction	r= .22 p= .19	ρ= -.02 p= .90	ρ= -.18 p= .29	r= -.07 p= .66	r= -.18 p= .29	r= -.05 p= .77	r= -.27 p= .11	r= -.27 p= .11	ρ = .28 p= .10	r= -.28 p = .10
Knee Abduction	r= .17 p = .32	ρ= .29 p= .09	ρ= -.48 p= .004	r= -.26 p= .19	r= -.43 p= .01 R^2=0.18	r= -.23 p= .18	r= -.50 p=.003 R^2=0.25	r= -.44 p= .008 R^2=0.19	ρ= -.46 p= .005	r= -.37 p= .02 R^2=0.13
Knee Flexion	r= .06 p = .70	ρ= .06 p= .70	ρ= .03 p= .85	r= .18 p= .28	r= .05 p= .74	r= .24 p= .16	r= -.14 p= .42	r= .01 p= .92	ρ = -.22 p = .20	r= -.04 p = .80
GVRF (*bw)	ρ= .25 p= .14	ρ = .38 p= .02	ρ= -.41 p= .01	ρ = .25 p= .14	ρ = .06 p= .73	ρ = .03 p= .86	ρ = .25 p= .14	ρ = .25 p= .14	ρ = .01 p = .91	ρ = -.13 p = .44

(ρ) Spearman and (r) Pearson correlation coefficients, (R^2) Coefficient of determination; correlation is significant at the level .05 (2-tailed) which has been highlighted.

Correlation between kinematics and kinetics with strength and EMG data during left SLS task

Left SLS										
Variables	G Max	G Med	Ext. Con.	Ext. Ecc.	Ext. Con./BW	Ext. Ecc./BW	Abd. Con.	Abd. Ecc	Abd. Con./BW	Abd. Ecc./BW
Joint Angle (°)										
Hip Flex	ϱ= -.19 p= .27	r= .20 p= .25	ϱ= .20 p= .25	r= .21 p= .21	r= .14 p= .41	r= .27 p= .11	r= .27 p= .11	r= .22 p= .19	ϱ= .24 p= .16	r= .20 p= .24
Hip Int. Rotation	ϱ= -.16 p= .35	r= -.34 p= .04 R^2= 0.12	ϱ= .09 p= .85	r= .08 p = .64	r= .05 p = .74	r= .06 p = .73	r= -.25 p = .14	r= -.03 p = .84	ϱ= .28 p= .10	r= .06 p = .73
Hip Adduction	ϱ= -.16 p= .35	r= .12 p= .45	ϱ= -.31 p= .07	r= -.27 p= .11	r= -.26 p= .12	r= -.17 p= .30	r= -.26 p= .14	r= -.17 p= .32	ϱ= .15 p= .37	r= .02 p= .90
Knee Abduction	ϱ= -.05 p= .72	r= -.24 p= .17	ϱ= .09 p= .60	r= .07 p = .65	r= .14 p = .40	r= .14 p = .42	r= .09 p = .60	r= .26 p = .12	ϱ= .13 p= .45	r= .30 p = .07
Knee Flexion	ϱ= -.33 p= .052	r= .08 p= .34	ϱ= .02 p= .78	r= .09 p = .60	r= .14 p = .40	r= .33 p = .05	r= .13 p = .43	r= .03 p = .86	ϱ= .29 p= .02	r= .23 p = .17
Moments (Nm/Kg)										
Hip Flex	ϱ= .01 p= .95	r= .18 p= .30	ϱ= -.16 p= .34	r= -.35 p= .04 R^2=0.12	r= -.19 p= .26	r= -.17 p= .33	r= .17 p= .31	r= .13 p= .43	ϱ= .01 p= .95	r= .10 p= .54
Hip Int. Rotation	ϱ= -.15 p= .39	r= .03 p = .87	ϱ= .29 p= .08	r= -.29 p = .09	r= -.17 p = .31	r= .18 p = .28	r= -.31 p = .07	r= .15 p = .37	ϱ= -.14 p= .45	r= .03 p = .83
Hip Adduction	ϱ= .07 p= .68	r= .06 p= .70	ϱ= .10 p= .55	r= .02 p= .89	r= .10 p= .55	r= .03 p= .84	r= .02 p= 90	r= .01 p= .91	ϱ= .01 p= .85	r= .02 p= .87
Knee Abduction	ϱ= .07 p= .67	r= .24 p = .15	ϱ= -.13 p= .43	r= .05 p = .77	r= .06 p = .70	r= .08 p = .61	r= .07 p = .69	r= .12 p = .49	ϱ= .08 p= .61	r= .12 p = .47
Knee Flexion	ϱ= .02 p= .87	r= .16 p = .34	ϱ= .12 p= .48	r= .16 p = .35	r= .15 p = .39	r= .25 p = .14	r= .25 p = .24	r= .01 p = .95	ϱ= .23 p= .18	r= .04 p = .81
GVRF (*bw)	ϱ= .29 p= .08	r= .18 p = .29	ϱ= .10 p= .55	r= .26 p = .11	r= .05 p = .73	r= .11 p = .53	r= .15 p = .38	r= .04 p = .79	ϱ= .01 p= .94	r= -.17 p = .32

(ρ) Spearman and (r) Pearson correlation coefficients, (R^2) Coefficient of determination; correlation is significant at the level .05 (2-tailed) which has been highlighted.

Correlation between kinematics and kinetics with strength and EMG data during right FL task

Right FL										
Variables	G Max	G Med	Ext. Con.	Ext. Ecc.	Ext. Con./BW	Ext. Ecc./BW	Abd. Con.	Abd. Ecc	Abd. Con./BW	Abd. Ecc./BW
Joint Angle (°)										
Hip Flex	r= .28 p= .10	r= .08 p= .62	ρ= .28 p= .10	r= .39 p= .02 R^2=0.15	r= .36 p= .03 R^2=0.13	r= .46 p= .006 R^2=0.21	r= .15 p= .38	r= .20 p= .23	ρ= .04 p= .37	r= .20 p= .21
Hip Int. Rotation	r= .04 p = .81	r= -.11 p = .51	ρ= .26 p= .12	r= .02 p= .88	r= .16 p= .33	r= .08 p= .99	r= .20 p = .24	r= .15 p = .39	ρ= .07 p= .62	r= .14 p = .40
Hip Adduction	r= .13 p= .46	r= .30 p= .07	ρ= .16 p= .36	r= .10 p = .56	r= .32 p = .06	r= .21 p = .21	r= -.02 p= .88	r= -.16 p= .36	ρ= .03 p= .83	r= -.09 p= .60
Knee Abduction	r= .09 p = .58	r= -.27 p = .11	ρ= .23 p= .17	r= .06 p= .73	r= -.22 p= .12	r= .06 p= .65	r= .33 p = .055	r= .32 p = .06	ρ= .12 p= .47	r= .32 p = .058
Knee Flexion	r= .16 p = .36	r= .11 p = .52	ρ= .07 p= .86	r= .15 p = .38	r= .28 p = .10	r= .31 p = .07	r= .09 p = .97	r= .18 p = .29	ρ= .00 p= .99	r= .28 p = .10
Moments (Nm/Kg)										
Hip Flex	r= .08 p= .63	r= -.06 p = .70	ρ= .03 p= .83	r= -.07 p = .68	r= .18 p = .31	r= .21 p = .22	r= .13 p= .44	r= -.01 p= .91	ρ= .23 p= .67	r= .07 p = .65
Hip Int. Rotation	r= .06 p = .71	r= .01 p = .95	ρ= .12 p= .47	r= -.16 p= .35	r= -.15 p= .38	r= -.10 p= .56	r= .09 p = .58	r= .01 p = .97	ρ= .07 p= .43	r= .05 p= .76
Hip Adduction	r= .16 p = .36	r= -.04 p = .80	ρ= .00 p= .99	r= .11 p = .51	r= .11 p = .50	r= .07 p = .65	r= .11 p = .51	r= .21 p = .22	ρ= .18 p= .67	r= .11 p = .53
Knee Abduction	ρ= .07 p= .67	ρ= .28 p= .10	ρ= -.48 p= .004	ρ= -.22 p= .19	ρ= .07 p= .67	ρ= .07 p= .67	ρ= -.47 p= .005	ρ= -.65 p= .001	ρ= .07 p= .67	ρ= .07 p= .67
Knee Flexion	r= .16 p = .36	r= -.11 p = .50	ρ= .18 p= .67	r= -.01 p= .11	r= .03 p= .11	r= .07 p= .76	r= -.20 p = .25	r= .11 p = .52	ρ= .07 p= .67	r= .22 p = .90
GVRF (*bw)	r= -.39 p= .02 R^2=0.15	r= .01 p = .95	ρ= -.42 p= .01	r= -.40 p= .01 R^2= 0.16	r= -.36 p= .03 R^2=0.12	r= -.34 p= .04 R^2= 0.12	r= -.47 p= .005 R^2=0.22	r= -.56 p= .001 R^2=0.31	ρ= -.41 p= .001	r= -.49 p= .01 R^2=0.24

(ρ) Spearman and (r) Pearson correlation coefficients, (R^2) Coefficient of determination; correlation is significant at the level .05 (2-tailed) which has been highlighted.

Correlation between kinematics and kinetics with strength and EMG data during left FL task

Left FW										
Variables	G Max	G Med	Ext. Con.	Ext. Ecc.	Ext. Con./BW	Ext. Ecc./BW	Abd. Con.	Abd. Ecc	Abd. Con./BW	Abd. Ecc./BW
Joint Angle (°)										
Hip Flex	r= .07 p = .65	r= .07 p = .65	ρ= .34 p= .04	r= .33 p = .054	r= .42 p= .01 R^2=0.17	r= .43 p= .01 R^2=0.18	r= .24 p = .16	r= .19 p = .28	ρ= .24 p= .16	r= .20 p = .25
Hip Int. Rotation	r= .07 p = .65	r= .07 p = .65	ρ= .11 p= .51	r= .01 p = .92	r= .09 p = .60	r= .05 p = .67	r= .20 p = .25	r= .01 p = .98	ρ= .17 p= .31	r= .05 p = .77
Hip Adduction	r= .07 p = .65	r= .40 p= .01 R^2= 0.16	ρ= -.36 p= .03	r= -.31 p = .07	r= .26 p = .12	r= .23 p = .17	r= .23 p = .18	r= -.38 p= .02 R^2= 0.14	ρ= .14 p= .40	r= -.25 p = .14
Knee Abduction	r= -.47 p= .004 R^2=0.22	r= -.38 p= .01 R^2= 0.14	ρ= .13 p= .45	r= .17 p = .31	r= .22 p = .19	r= .25 p = .14	r= .06 p = .71	r= .33 p = .051	ρ= .08 p= .64	r= .35 p= .03 R^2=0.12
Knee Flexion	ρ= .08 p= .62	ρ= -.19 p= .54	ρ= .22 p= .19	ρ= .18 p= .28	ρ= .32 p= .06	ρ= .42 p= .06	ρ= .15 p= .39	ρ= .09 p= .54	ρ= .29 p= .09	ρ= .27 p= .11
Moments(Nm/Kg)										
Hip Flex	r= .07 p = .65	r= .07 p = .65	ρ= .05 p= .47	r= .04 p = .78	r= .15 p = .38	r= -.02 p = .90	r= .11 p = .51	r= .07 p = .67	ρ= .16 p= .35	r= .02 p = .90
Hip Int. Rotation	ρ= .09 p= .61	ρ= .02 p= .89	ρ= .18 p= .30	ρ= .09 p= .58	ρ= .17 p= .31	ρ= .03 p= .85	ρ= .05 p= .67	ρ= .05 p= .74	ρ= .04 p= .79	ρ= .18 p= .29
Hip Adduction	r= .07 p = .65	r= .07 p = .65	ρ= .24 p= .16	r= .33 p = .052	r= .28 p = .10	r= .33 p = .054	r= .22 p = .19	r= -.48 p= .004 R^2= 0.23	ρ= .17 p= .32	r= .42 p= .01 R^2=0.17
Knee Abduction	ρ= .07 p= .67	ρ= .07 p= .67	ρ= .11 p= .52	ρ= .05 p= .74	ρ= .20 p= .24	ρ= .09 p= .57	ρ= .04 p= .67	ρ= .13 p= .35	ρ= .07 p= .67	ρ= .16 p= .32
Knee Flexion	r= .07 p = .65	r= .07 p = .65	ρ= .14 p= .42	r= .08 p = .62	r= .09 p = .58	r= .07 p = .66	r= .17 p = .31	r= .20 p = .22	ρ= .04 p= .78	r= .06 p = .71
GVRF (*bw)	r= .07 p = .65	r= .07 p = .65	ρ= -.55 p= .01	r= -.43 p= .01	r= -.53 p= .001	r= .47 p= .005	r= -.51 p= .002	r= -.64 p= .001	ρ= -.48 p= .004	r= -.62 p= .001

(ρ) Spearman and (r) Pearson correlation coefficients, (R^2) Coefficient of determination; correlation is significant at the level .05 (2-tailed) which has been highlighted.

Correlation between kinematics and kinetics with strength and EMG data during right SML task

Right SML										
Variables	G Max	G Med	Ext. Con.	Ext. Ecc.	Ext. Con./BW	Ext. Ecc./BW	Abd. Con.	Abd. Ecc	Abd. Con./BW	Abd. Ecc./BW
Joint Angle(°)										
Hip Flex	r= .37 p= .02 R^2=0.11	r= .12 p = .48	ρ= .24 p= .16	r= .26 p = .13	r= .33 p = .054	r= .24 p = .16	r= .21 p = .23	r= .24 p = .17	ρ= .24 p= .16	r= .10 p = .57
Hip Int. Rotation	r= .26 p = .13	r= .07 p = .65	ρ= .25 p= .14	r= .11 p = .57	r= .01 p = .92	r= .20 p = .25	r= .22 p = .19	r= .21 p = .24	ρ= .17 p= .31	r= .25 p = .15
Hip Adduction	r= .19 p = .26	r= .47 p= .005 R^2=0.22	ρ= .03 p= .86	r= .05 p = .74	r= -.31 p = .07	r= .23 p = .18	r= .19 p = .26	r= .19 p = .26	ρ= .14 p= .40	r= .02 p = .87
Knee Abduction	r= .18 p = .28	r= -.20 p = .25	ρ= .23 p= .18	r= .17 p = .33	r= .17 p = .31	r= .06 p = .71	r= .33 p= .04 R^2=0.11	r= .38 p= .02 R^2=0.16	ρ= .39 p= .02	r= .41 p= .01 R^2=0.16
Knee Flexion	r= .25 p = .14	r= .04 p = .14	ρ= .13 p= .44	r= .13 p = .45	r= .18 p= .28	r= .03 p = .85	r= .02 p = .95	r= .09 p = .59	ρ= .24 p= .16	r= .09 p = .59
Moments(Nm/Kg)										
Hip Flex	r= .01 p = .91	r= .22 p = .19	ρ= .13 p= .44	r= .10 p = .57	r= .16 p = .34	r= .04 p = .57	r= -.12 p = .47	r= .34 p = .06	ρ= .24 p= .16	r= .38 p = .07
Hip Int. Rotation	r= .02 p = .90	r= .03 p = .68	ρ= .22 p= .19	r= .25 p = .15	r= .17 p = .15	r= .12 p = .45	r= .25 p = .15	r= .18 p = .30	ρ= .17 p= .31	r= .01 p = .15
Hip Adduction	r= .05 p = .75	r= .04 p = .75	ρ= .03 p= .85	r= .02 p = .87	r= .02 p = .87	r= .07 p = .37	r= .13 p = .45	r= .03 p = .85	ρ= .14 p= .40	r= .02 p = .87
Knee Abduction	ρ= .08 p= .62	ρ= .17 p= .31	ρ= .23 p= .17	ρ= .18 p= .29	ρ= .31 p= .06	ρ= .22 p= .19	ρ= .15 p= .39	ρ= -.47 p= .01	ρ= .15 p= .39	ρ= .39 p= .03
Knee Flexion	ρ= -.09 p= .58	ρ= -.22 p= .19	ρ= .03 p= .87	ρ= .05 p= .77	ρ= .06 p= .46	ρ= .11 p= .13	ρ= .15 p= .39	ρ= .09 p= .54	ρ= .29 p= .09	ρ= .27 p= .11
GVRF (*bw)	r= -.39 p= .02 R^2=0.15	r= .22 p = .19	ρ= -.37 p= .01	r= -.34 p= .02 R^2=0.11	r= -.28 p = .09	r= -.24 p = .15	r= -.45 p= .007 R^2=0.2	r= -.45 p= .004 R^2=0.2	ρ= -.37 p= .02	r= -.35 p= .04 R^2=0.12

(ρ) Spearman and (r) Pearson correlation coefficients, (R^2) Coefficient of determination; correlation is significant at the level .05 (2-tailed) which has been highlighted.

Correlation between kinematics and kinetics with strength and EMG data during left SML task

Left SML										
Variables	G Max	G Med	Ext. Con.	Ext. Ecc.	Ext. Con./BW	Ext. Ecc./BW	Abd. Con.	Abd. Ecc	Abd. Con./BW	Abd. Ecc./BW
Joint Angle(°)										
Hip Flex	r= .41 p= .01 R^2=0.127	r= .06 p= .74	ϱ= .24 p= .16	r= .29 p = .09	r= .28 p = .10	r= 36 p= .03 R^2=0.13	r= .11 p = .51	r= .09 p = .59	ϱ = .07 p = .68	r= .06 p = .71
Hip Int. Rotation	r= .19 p= .27	r= .09 p= .58	ϱ= .04 p= .80	r= .03 p = .86	r= .06 p = .67	r= .05 p = .77	r= .18 p = .28	r= .05 p = .77	ϱ = .19 p = .27	r= -.04 p = .80
Hip Adduction	r= .22 p= .20	r= .39 p= .02 R^2=0.15	ϱ= .31 p= .07	r= .21 p = .22	r= -.31 p = .06	r= -.20 p = .24	r= -.42 p= .01 R^2=0.18	r= -.51 p= .002 R^2=0.26	ϱ= .39 p= .02	r= -.47 p= .004 R^2=0.22
Knee Abduction	r= .06 p= .73	r= .07 p= .67	ϱ= .18 p= .300	r= .18 p = .28	r= .15 p = .36	r= .18 p = .30	r= .07 p = .68	r= .26 p = .13	ϱ = .02 p = .29	r= .18 p = .35
Knee Flexion	ϱ= .38 p= .02	ϱ = .09 p= .61	ϱ= .13 p= .44	ϱ= .18 p= .28	ϱ= .32 p= .06	r= .40 p= .01 R^2=0.16	ϱ= .15 p= .39	ϱ= .03 p= .87	ϱ= .18 p= .28	ϱ= .32 p= .06
Moments(Nm/Kg)										
Hip Flex	r= .10 p = .57	r= -.05 p= .56	ϱ= -.07 p= .66	r= .04 p = .81	r= -.06 p = .86	r= .03 p = .68	r= .03 p = .90	r= .04 p = .82	ϱ = .07 p = .58	r= .08 p = .61
Hip Int. Rotation	ϱ= -.12 p= .47	ϱ = .01 p= .20	ϱ= .25 p= .14	ϱ= .18 p= .28	ϱ= .32 p= .06	ϱ= .42 p= .06	ϱ= .15 p= .39	ϱ= .32 p= .06	ϱ= .42 p= .06	ϱ= .15 p= .39
Hip Adduction	r= .13 p = .48	r= .02 p= .88	ϱ= .24 p= .15	r= .30 p = .08	r= .18 p = .29	r= .30 p = .08	r= .25 p = .14	r= .35 p= .04 R^2=0.12	ϱ = .16 p = .35	r= -.26 p = .13
Knee Abduction	ϱ= -.03 p= .58	ϱ= .22 p= .20	ϱ= .23 p= .18	ϱ= .18 p= .28	ϱ= .32 p= .06	ϱ= .42 p= .06	ϱ= .15 p= .39	ϱ= .03 p= .87	ϱ= .18 p= .28	ϱ= .32 p= .06
Knee Flexion	r= -.22 p = .08	r= .07 p= .54	ϱ= .13 p= .44	r= .13 p = .45	r= .08 p = .65	r= .13 p = .29	r= .07 p = .69	r= .11 p = .52	ϱ = .01 p = .95	r= -.05 p = .35
GVRF (*bw)	ϱ= -.16 p = .36	ϱ= -.04 p= .19	ϱ= -.22 p= .11	ϱ= -.20 p= .11	ϱ= -.16 p= .23	ϱ= -.14 p= .14	ϱ= -.17 p= .51	ϱ= -.26 p= .17	ϱ= -.21 p= .21	ϱ= -.29 p= 28

(ρ) Spearman and (r) Pearson correlation coefficients, (R^2) Coefficient of determination; correlation is significant at the level .05 (2-tailed) which has been highlighted.

Correlation between kinematics and kinetics with strength and EMG data during right SLL task

Right SLL										
Variables	G Max	G Med	Ext. Con.	Ext. Ecc.	Ext. Con./BW	Ext. Ecc./BW	Abd. Con.	Abd. Ecc	Abd. Con./BW	Abd. Ecc./BW
Joint Angle (°)										
Hip Flex	r= .52 p= .001 R^2=0.27	r= .09 p = .62	ϱ= .39 p= .02	r= .36 p= .02 R^2=0.13	r= .37 p= .03 R^2=0.13	r= .35 p= .03 R^2=0.12	r= .33 p = .054	r= .37 p = .07	ϱ= .28 p= .10	r= .31 p = .06
Hip Int. Rotation	r= .06 p = .73	r= .15 p = .38	ϱ= .19 p= .26	r= .02 p = .88	r= .21 p= .28	r= .07 p = .68	r= .16 p= .28	r= .15 p = .85	ϱ= .23 p= .28	r= .19 p = .27
Hip Adduction	r= .13 p = .45	r= .34 p= .04 R^2=0.12	ϱ= .07 p= .67	r= .04 p = .79	r= .20 p= .28	r= .19 p = .85	r= .18 p= .28	r= .19 p = .34	ϱ= .10 p= .28	r= .09 p = .85
Knee Abduction	r= .37 p= .07	r= .20 p = .23	ϱ= .32 p= .06	r= .11 p = .50	r= .25 p= .18	r= .09 p = .34	r= .34 p= .04 R^2=0.12	r= .37 p= .03 R^2=0.14	ϱ= .33 p= .05	r= .34 p= .04 R^2=0.12
Knee Flexion	r= .44 p= .01 R^2=0.17	r= .04 p = .14	ϱ= .34 p= .04	r= .13 p = .45	r= .37 p= .02	r= .33 p = .05	r= .19 p = .26	r= .29 p = .09	ϱ= .24 p= .16	r= .28 p = .59
Moments(Nm/Kg)										
Hip Flex	r= .25 p = .14	r= .04 p = .14	ϱ= .16 p= .35	r= .13 p = .45	r= .13 p= .28	r= .03 p = .85	r= .12 p = .45	r= -.41 p = .06	ϱ= .24 p= .16	r= .09 p = .59
Hip Int. Rotation	r= .03 p = .91	r= .14 p = .40	ϱ= -.36 p= .03	r= -.37 p= .03 R^2=0.14	r= -.32 p= .06	r= -.20 p = .25	r= -.46 p= .006 R^2=0.21	r= -.47 p= .005 R^2=0.22	ϱ= -.33 p= .055	r= .32 p = .06
Hip Adduction	ϱ= -.07 p= .69	ϱ= -.07 p= .66	ϱ= .01 p= .96	ϱ= .05 p= .77	ϱ= .06 p= .46	ϱ= .11 p= .13	ϱ= .15 p= .39	ϱ= .09 p= .54	ϱ= .29 p= .09	ϱ= .27 p= .11
Knee Abduction	ϱ= -.22 p= .19	ϱ= .11 p= .51	ϱ= -.41 p= .01	ϱ= .05 p= .77	ϱ= .06 p= .46	ϱ= .11 p= .13	ϱ= .15 p= .39	ϱ= .09 p= .54	ϱ= .29 p= .09	ϱ= .27 p= .11
Knee Flexion	r= -.30 p = .07	r= .04 p = .14	ϱ= .02 p= .44	r= .13 p = .45	r= .18 p= .28	r= .03 p = .85	r= .01 p = .92	r= .15 p = .24	ϱ= .03 p= .16	r= .16 p = .59
GVRF (*bw)	r= -.60 p= .001 R^2= 0.36	r= -.03 p = .86	ϱ= -.48 p= .004	r= -.40 p= .01 R^2=0.16	r= -.37 p= .03 R^2=0.13	r= -.31 p= .07	r= .15 p = .85	r= -.57 p= .001 R^2=0.32	ϱ= -.47 p= .005	r= -.48 p= .004 R^2=0.23

(ρ) Spearman and (r) Pearson correlation coefficients, (R^2) Coefficient of determination ; correlation is significant at the level .05 (2-tailed) which has been highlighted.

Correlation between kinematics and kinetics with strength and EMG data during left SLL task

Left SLL										
Variables	G Max	G Med	Ext. Con.	Ext. Ecc.	Ext. Con./BW	Ext. Ecc./BW	Abd. Con.	Abd. Ecc	Abd. Con./BW	Abd. Ecc./BW
Joint Angle (°)										
Hip Flex	r= .06 p= .54	r= .07 p= .75	ϱ = .12 p = .49	r= .22 p= .19	r= .22 p= .19	r= .22 p= .19	r= .13 p= .44	r= .14 p= .40	ϱ = .12 p = .49	r= .11 p= .30
Hip Int. Rotation	r= .16 p = .34	r= .03 p = .84	ϱ = .12 p = .49	r= -.03 p = .84	r= -.03 p = .84	r= -.03 p = .84	r= .11 p= .57	r= .07 p= .86	ϱ = .12 p = .49	r= .07 p= .69
Hip Adduction	r= -.15 p= .37	r= .37 p= .03 R^2=0.13	ϱ = .12 p = .49	r= -.16 p= .34	r= -.16 p= .34	r= -.17 p= .32	r= -.38 p= .02 R^2=0.14	r= -.51 p= .002 R^2=0.26	ϱ= -.37 p= .03	r= -.49 p= .003 R^2=0.24
Knee Abduction	ϱ = .07 p = .62	ϱ = .06 p = .72	ϱ = .12 p = .49	ϱ = .03 p = .49	ϱ = .12 p = .49	ϱ = .12 p = .49	ϱ = .12 p = .49	ϱ = .12 p = .49	ϱ = .12 p = .49	ϱ = .36 p = .03
Knee Flexion	r= .06 p = .73	r= .11 p = .52	ϱ = .12 p = .49	r= .26 p = .86	r= .26 p = .86	r= .03 p = .86	r= .02 p= .61	r= .09 p= .59	ϱ = .12 p = .49	r= .20 p= .25
Moments(Nm/Kg)										
Hip Flex	r= .14 p= .43	r= .02 p = .87	ϱ = .12 p = .49	r= .04 p= .19	r= .22 p= .19	r= .22 p= .19	r= .02 p= .19	r= .14 p= .41	r= .22 p= .19	r= .18 p= .29
Hip Int. Rotation	r= -.28 p = .37	r= .19 p = .73	ϱ = .12 p = .49	r= -.11 p = .51	r= -.03 p = .84	r= -.03 p = .84	r= -.13 p = .84	r= .17 p= .33	r= .22 p= .19	r= .05 p= .76
Hip Adduction	r= .01 p= .91	r= -.39 p= .02 R^2=0.15	ϱ = .12 p = .49	r= -.11 p= .51	r= -.17 p= .32	r= -.17 p= .32	r= -.22 p= .32	r= .44 p= .008 R^2=0.2	ϱ = .12 p = .49	r= .45 p= .007 R^2=0.2
Knee Abduction	ϱ = .12 p = .49	ϱ = .12 p = .49	ϱ = .12 p = .49	ϱ = .07 p = .49	ϱ = .12 p = .49	ϱ = .12 p = .49	ϱ = .12 p = .49	ϱ = .12 p = .49	ϱ = .12 p = .49	ϱ = .12 p = .49
Knee Flexion	r= .01 p = .95	r= .19 p = .28	ϱ = .12 p = .49	r= .14 p = .42	r= .03 p = .86	r= .03 p = .86	r= .13 p = .86	r= .31 p= .07	r= .22 p= .19	r= .28 p= .10
GVRF (*bw)	r= .04 p = .79	r= .01 p = .93	ϱ = -.45 p= .006	r= -.47 p= .005 R^2=0.22	r= -.49 p= .004 R^2=0.24	r= -.48 p= .003 R^2=0.23	r= -.48 p= .003 R^2=0.23	r= -.48 p= .004 R^2=0.23	r= -.39 p= .02	r= -.40 p= .02 R^2=0.16

(ρ) Spearman and (r) Pearson correlation coefficients, (R^2) Coefficient of determination ; correlation is significant at the level .05 (2-tailed) which has been highlighted.