Biomechanical Characteristics of Multidirectional Single-leg Landing

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Table of Contents

List of tables

List of figures

List of abbreviations

SLL	Single-leg landing	RCT	Randomised control trial
3D	Three-dimensional	ITB	Iliotibial band
2D	Two-dimensional	DVJ	Drop vertical jump
SEM	Standard error of measurement	IC	Initial contact
FPPA	Frontal plane projection angle	ITBS	Iliotibial band syndrome
HADD	Hip adduction	ASIS	Anterior superior iliac spine
ACL	Anterior cruciate ligament	VGRF	Vertical ground reaction force
SLS	Single-leg squat	ICC	Intraclass correlation coefficient
VDJ	Vertical drop jump	FPT	Functional performance test
PFPS	Patellofemoral pain syndrome	FSLL	Forward single-leg landing
GRF	Ground reaction force	FSLLP	Forward single-leg landing off platform
ROM	Range of motion	LSLL	Lateral single-leg landing
PCL	Posterior cruciate ligament	LSLLP	Lateral single-leg landing off platform
MCL	Medial collateral ligament	MSLL	Medial single-leg landing
LCL	Lateral collateral ligament	MSLLP	Medial single-leg landing off platform

Continued

Continued list of abbreviations

NAIRS	National Athletic Injury/ Illness Reporting System	CI	Confidence interval
OA	Osteoarthritis	CAST	The calibration anatomical system technique
ACLR	Anterior cruciate ligament reconstruction	CMC	Coefficient of multiple correlation
VMO	Vastus medialis oblique	SLTH	Single-leg triple hop
IN	Intercondylar notch	SLHH	Single-leg hurdles hop
NCAA	National Collegiate Athletics Association	LSI	Limb symmetry index
MRI	Magnetic resonance imaging	PSIS	Posterior superior iliac spines
DLL	Double-leg landing	DoF	Degrees of freedom
PFJ	Patellofemoral joint		
Nm/kg	Newton-meter per kilogram		

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Abstract

Single-leg landing (SLL) is a functional task that has been linked to injury. It is the test most used in both research and clinical practice to evaluate the dynamic stability of the lower extremities, particularly the knee joint. It is also an important screening tool that can be used to identify those who are at risk of lower-extremity injury and to evaluate the progress of rehabilitation regimes for individuals with lower-limb injuries. However, SLL occurs in multiple directions and from different heights during sport activity. Limited literature explores the biomechanical characteristics of SLL tasks and the association between different directions of SLL. A better understanding of SLL biomechanical characteristics and the relationship between different types of SLL may provide important information to help understand how individual joint biomechanics behave under different types of SLL to meet the demands of sport.

Four themed studies are included in this thesis. The first study is a systematic review that aims to review the available literature that has investigated the biomechanics of the lower extremities during multidirectional SLL. The results indicate that only SLL in a forward direction is tested in the majority of the literature using three-dimensional (3D) motion analysis, indicating the importance of examining other directions that seen in sports or used clinically.

The second study aims to examine within-day and between-days reliability and establish standard errors of measurement (SEM) for lower-extremity biomechanical variables using both two-dimensional (2D) and 3D motion analysis during multidirectional SLL. The majority of 2D and 3D variables show good to excellent reliability with relatively small SEM. However, knee valgus moment and hip adduction moment are less reliable among all the tasks assessed using 3D motion analysis.

The third study investigates the correlation between 2D and 3D motion-analysis techniques when measuring the lower-extremity frontal plane of movement during multidirectional SLL. The results indicate that the 2D frontal plane projection angle (FPPA), at best, moderately correlates with 3D knee valgus angle, while the 2D hip adduction (HADD) angle shows strong significant correlation with 3D HADD angle, ranging between r = 0.70 to r = 0.90 across all tasks, apart from the right leg during medial single-leg landing off-platform, which had only a small association (r = 0.27), suggesting that 2D is a good alternative to 3D when measuring hip angles, though it should be used with caution when measuring knee angles.

The final study examines the relationship between biomechanical variables during multidirectional SLL using both 2D and 3D motion-analysis techniques. The vast majority of 2D and 3D variables reported significant moderate to very strong correlations across all examined tasks. These findings suggest that a single task can be used to represent the biomechanical variables found across other tasks, so that when measuring lower-limb biomechanics, a clinician may not need to conduct all these tests.

What this thesis adds to the current body of knowledge is that multidirectional SLL can be done in a reliable manner to measure lower-extremity biomechanical variables using either 2D or 3D motion analysis. 2D motion analysis can be used as a valid alternative to 3D, particularly for hip angle assessment, and single tasks can be used in isolation to represent lower-limb biomechanics across a multitude of tasks.

1. Introduction

In the last few decades, participation in sports and physical activities has increased significantly (Niemuth, Johnson, Myers, & Thieman, 2005). Such activities include different technical skills and dynamic manoeuvres and involve activities of different intensities, such as landing (Bangsbo & Michalsik, 2002). This may have led to an escalation in the numbers of injuries, with most of these injuries affecting both genders and being predominantly in the lower limbs (Emery, Meeuwisse, & McAllister, 2006). Lower-limb injury rates can be as high as 8.0, 21.5 and 2.10 injuries per 1,000 hours of playing football (Ekstrand, Hagglund, & Walden, 2011), tennis or volleyball (Sattler, 2011), respectively. Such injuries impose a significant economic burden' for example, about 200,000 anterior cruciate ligament (ACL) injuries are reported in the USA every year (Maffulli, & Osti, 2013), with a total cost of around $2 billion (McCullough et al., 2012). These injuries cost about A$75 million per year in Australia (Von Porat, Roos, & Roos, 2004) and approximately NZ$222 million in New Zealand (Gianotti & Hume, 2007). There is, however, a lack of studies in other countries on the economic and social impact.

The knee joint is one of those most commonly injured (Heintjes et al., 2009). Over the last twenty years, epidemiological studies have noted a significant increase in acute and overuse injuries to the lower extremities (Heinert, Kernozek, Greany, & Fater, 2008; Snyder, Earl, O'Connor, & Ebersole, 2009) and in knee pain which affects over 40 per cent of athletes, across all sports, during their careers (Stakes et al., 2006). Most knee injuries are non-contact in nature, meaning that injury may occur because of the movement of the person, not because of external force being applied by another person or object (Hewett et al., 2005; Olsen, Myklebust, Engebretsen, & Bahr, 2004).

One of the most severe and damaging knee injuries is non-contact ACL rupture, and the incidence of this across both genders has risen by almost 50 per cent during the last decade (Donnelly et al., 2012). Such injuries are usually seen in sports that include rapid deceleration or change-of-direction manoeuvres, such as cutting and landing (Borotikar, Newcomer, Koppes, & McLean, 2008; Quatman, Quatman-Yates, & Hewett, 2010).

Non-contact ACL injury has been associated with many factors. One of the main ones is abnormal biomechanics of the lower extremities on landing or when changing direction (Souza & Powers, 2009). Dynamic knee valgus, which can be explained as alteration to knee, hip and ankle kinematics, is suggested as being a significant biomechanical motion related to knee injury (Hewett et al., 2005; Souza & Powers, 2009) as it puts a large force on the knee joint, specifically the ACL. Decreased active neuromuscular control of the lower extremities due to neuromuscular control deficits, particularly of the lower extremities, are also suggested as contributing to knee-ligament injury, as this can lead to increased abduction load and strain on the knee (Myer et al., 2009). Such injury occurs as part of a complex multifaceted process that needs to be fully defined, along with its epidemiology, aetiology, risk factors and the exact mechanism (Bittencourt et al., 2016).

Injury risk factors are generally divided into intrinsic and extrinsic factors (Murphy, Connolly, & Beynnon, 2003). Intrinsic factors have received more attention, and the majority of research, as they are potentially changeable (Halabchi, Mazaheri, & Seif-Barghi, 2013) and have been suggested as being more closely related to injury prediction than are extrinsic factors (Orchard, Seward, McGivern, & Hood, 2001).

To assess intrinsic risk factors, different methods have been used. Baseline or pre-participation screening is the most common and has been used to identify the characteristics of musculoskeletal systems of those who not fully recovered from injury or who are prone to injury (Dennis, Finch, McIntosh, & Elliott, 2008). Additionally, it is commonly used to enhance performance strategies (Mottram & Comerford, 2008). Only a limited number of high quality studies have examined injury-risk factors. Therefore, the validity of current protocols is not fully established (Mottram & Comerford, 2008).

Lower-limb biomechanics during functional tasks has been examined in various studies (Table 1.1). Some of these tests are bilateral tests which prevent a comparison between the sound and the affected legs. Such a comparison is possible with tests that require only one leg to be examined, as the sound leg can be used as a control while quantifying the function of the affected leg. Differences in function between the injured and non-injured legs were found in a study that assessed only one leg (Goerger et al., 2014). As landing in sport mostly occurs unilaterally, and unilateral tasks make up about 70 per cent of non-contact ACL injuries (Olsen et al., 2004; Boden, Dean, Feagin, & Garrett, 2000; Kirkendall & Garrett, 2000), it seems imperative to improve the biomechanics knowledge of multi-directional single-leg landing. Studies that have examined single-leg tasks have mainly focused only on the sagittal plane. Yet, the frontal plane of movement is also important because excessive movement within it, particularly knee valgus and HADD, are associated with non-contact ACL injury (Shin et al., 2011; Paterno et al., 2010). Thus, examining frontal plane biomechanics during a unilateral task is important to understand how individual joint biomechanics respond to meet sport demands.

Table 1.1 : Functional tasks that have been used to examine lower-limb biomechanics

Functional task	Authors
Single-leg squat (SLS)	Zeller, McCrory, Kibler, & Uhl, 2003; DiMattia, Livengood, Uhl, Mattacola, & Malone, 2005; Pantano, White, Gilchrist, & Leddy, 2005; Claiborne, Armstrong, Gandhi, & Pincivero, 2006; Zwerver, Bredeweg, & Hof, 2007; Whatman, Hing, & Hume, 2011; Munro et al., 2012a; Mauntel, Frank, Begalle, Blackburn, & Padua, 2014; Stickler, Finley, & Gulgin, (2015).
Landing	Ford, Myer, & Hewett, 2003; Decker, Torry, Wyland, Sterett, & Steadman, 2003; Hewett, Myer, & Ford, 2004; Kernozek, Torry, Van Hoof, Cowley, & Tanner, 2005; Hass et al., 2005; Herrington, 2014; Atkin, Herrington, Alenezi, Jones, & Jones, (2014).
Vertical drop-jump (VDJ)	Munro et al., (2012a); Mok, Kristianslund, & Krosshaug, (2015).
Side-step and side-jump	McLean et al., (2005).
Cutting	Besier, Lloyd, Cochrane, & Ackland, 2001; Malinzak, Colby, Kirkendall, Yu, & Garrett, 2001; Houck & Yack, 2003; McLean, Lipfert, & van den Bogert, 2004a; Pollard, Davis, & Hamill, 2004; Hamill, Heiderscheit, & Pollard, 2005; McLean et al., 2005; Dempsey, Lloyd, Elliott, Steele, & Munro, 2009; Mok et al., (2015).
Running	Rand & Ohtsuki, 2000; Besier et al., 2001; Malinzak et al., 2001; Ferber, Davis, & Williams Iii, 2003; Vanrenterghem, Venables, Pataky, & Robinson, 2012; Atkin et al., (2014).
Single-limb step-down	Hollman et al., (2009).
SLL	Malinzak et al., 2001; Lephart, Ferris, Riemann, Myers, & Fu, 2002; Fagenbaum & Darling. 2003; Decker et al., 2003; Ford et al., 2003; Zeller et al., 2003; McLean et al., 2004a; Hass et al., 2005; Yu, Lin, & Garrett, 2006; Pappas, Hagins, Sheikhzadeh, Nordin, & Rose, 2007; Yeow, Lee, & Goh, 2010; Munro et al., (2012a).
Jumping	Willson & Davis, (2009).
Lunge-hop and step-down	Whatman et al., (2011).
Drop-jump	Noyes, Barber-Westin, Fleckenstein, Walsh, & West, 2005; Ford, Myer, & Hewett, (2007).
Single-leg hop and/or one-leg hop for distance	Noyes, Barber, & Mangine, 1991; Hurd, Axe, & Snyder-Mackler, 2008; Orishimo, Kremenic, Mullaney, McHugh, & Nicholas, 2010; Grindem et al., 2011; Logerstedt et al., 2012; Roos, Button, Sparkes, & van Deursen, (2014).
Triple-jump	Ostenberg, Roos, Ekdahl, & Roos, 1998; Holm et al., (2004).
Cross-over hop for distance	Wilk, Romaniello, Soscia, Arrigo, & Andrews, 1994; Eastlack 1999; Bjorklund, Andersson, & Dalén, 2009; Myer et al., (2011).
Side-hop	Elmlinger, Nyland, & Tillett, 2006; Gustavsson et al., (2006).

The use of functional tests has become the most popular mechanism to assess athletes' functional disability and readiness to return to participation, because it provides quantitative and qualitative information about specific movement. Although it has some limitations, such as the need for equipment that is often not available to coaches or sports medical teams and the presence of a practice effect, its limitations are still less than other tests (Reiman & Manske, 2009; Reiman & Manske, 2011; Narducci, Waltz, Gorski, Leppla, & Donaldson, 2011). An SLL test is a functional performance test and is most commonly used in both research and clinical practice to evaluate the dynamic stability of the lower extremities, particularly the knee joint (Dos Reis et al., 2015;

Bjorklund et al., 2009; Bjorklu et al., 2006). It is also an important screening tool that can be used to identify those who are at risk of lower-extremity injury and to evaluate the progress of rehabilitation for individuals with ACL injury or patellofemoral pain syndrome (PFPS) (Fukuda et al., 2012; Magalhaes et al., 2010; Grindem et al., 2011). This task has been suggested as being a good indicator of athletes' readiness to return to sport and has shown good reliability and validity in measuring different components of movement, such as strength, stability, joint mobility, neuromuscular control, balance and agility (Ardern, Webster, Taylor, & Feller, 2011; Reiman & Manske, 2009; Clark, 2001b). It is multi-segmental movement that requires coordination (Orishimo et al., 2010) and can place high demands on the lower limbs to absorb ground reaction force (GRF) (Decker et al., 2003; Paterno, Ford, Myer, Heyl, & Hewett, 2007). There are different types of SLL described in the literature. Because of the differences in terms that have been used in previous studies and the importance of the landing phase in any task, for the purposes of this project, the term will be standardised and ″SLL″ will be used when referring to a task involving a landing on one leg. But when discussing a specific study, the same terms used by authors will be employed.

Many studies have investigated the biomechanics of SLL (Pappas et al., 2007; Yeow et al., 2010; Munro et al., 2012a; Yu et al., 2006; Hass et al., 2005; Malinzak et al., 2001; Lephart et al., 2002; Fagenbaum & Darling, 2003; Decker et al., 2003; Ford et al., 2003; Zeller et al., 2003; McLean et al., 2004a). These studies have, however, mainly examined SLL in one direction only (forward) while sports demands require multidirectional landings; consequently, multidirectional SLL needs to be examined.

Functional tests are usually evaluated by quantity (e.g. distance) and quality (e.g. kinematics and kinetics) information about specific movements. The quality of movement can be determined during landing (Ekegren, Miller, Celebrini, Eng, & Macintyre, 2009). Most of the studies that have examined quality of movement only focused on one component (kinematics or kinetics). Examining both components would appear to be crucial in rehabilitation and avoiding injury and re-injury (Renstrom et al., 2008; Paterno et al., 2010; Thomeé, & Werner, 2011). Therefore, frontal plane kinematics and kinetics during multidirectional SLL should be examined.

Different methods have been used to evaluate lower-body mechanics during athletic tasks. However, 3D and 2D motion analysis are the methods most commonly used. 3D motion analysis is frequently used to quantify knee- and hip-joint biomechanics in the literature (Ford et al., 2003; Hewett et al., 2005; Souza & Powers, 2009; Pappas et al., 2007; Orishimo, Kremenic, Pappas, Hagins, & Liederbach, 2009; Ortiz, Olson, Trudelle-Jackson, Rosario, & Venegas, 2011). It is considered to be the gold standard, as it allows investigating biomechanics in multiple directions (frontal, sagittal and transverse). The above-mentioned studies conclude that altered loading on the knee joint is associated with injury. This load is a result of proximal and distal segment movement. But most of these studies focused on the sagittal plane or landing in one direction only (forwards). Moreover, 3D outcomes should be reproducible in order to see, for instance, changes in performance over time. Consequently, the reliability of 3D variables during multidirectional SLL should be established to a greater extent than is currently the case.

While 3D motion analysis can collect valuable information about joint biomechanics, its extension into clinical settings or large populations is limited. Therefore, 2D video-motion analysis could prove to be a good alternative. This method is more suitable for use in the field or clinical settings and may provide similar results to a 3D system (Mclean et al. 2005). Many studies have used 2D methods to evaluate lower-limb extremities (Munro et al., 2012a; Mclean et al., 2005; Norris & Olson, 2011), they show excellent intra-rater reliability for knee valgus and HADD angles (Hollman et al., 2009). Moderate to high reliability exists for knee valgus angle (Miller & Callister, 2009) during different performance tests (Norris and Olson, 2011). Also, the validity of 2D vis-à-vis 3D has been established during different tasks, such as side-jump (Mclean et al., 2005), mechanical lifting (Norris and Olson, 2011), single-leg step (Olson, Chebny, Willson, Kernozek, & Straker, 2011), SLS and single-leg step-down (Willson & Davis, 2008; Olson et al., 2011). However, no studies have considered the reliability and validity of 2D to examine multidirectional SLL and this needs to be addressed.

Comparisons of biomechanics across athletic tasks can explain the characteristics of these tasks and help to identify those which pose a risk of injury. This, in turn, can help in the prevention and treatment of injury. A few studies have compared tasks in terms of biomechanical characteristics, particularly frontal-plane kinematic and kinetics (Whatman et al., 2011; Kristianslund & Krosshaug, 2013; Whatman, Hume, & Hing, 2013; Jones, Herrington, Munro, & Graham-Smith, 2014; Donohue et al., 2015; McLean et al., 2005; Earl, Monteiro, & Snyder, 2007; Pappas et al., 2007; Imwalle, Myer, Ford, & Hewett, 2009; Harty, DuPont, Chmielewski, & Mizner, 2011).

Most of these studies only examined a small sample of females and did not calculate a Coefficient of determination (r^2), which is important when conducting a correlation study as it explains how the proportion of one variable can be explained from the other variables. Moreover, most of the studies examined the correlation between double- and single-leg tasks. The possibility of a relationship between a high-demand task (a single-leg landing is suggested to place a high load on the ACL) and a lower-demand task (double leg) might be limited due to the differences in the nature of the tasks. There are many tasks that are often seen in the sporting environment and used as a screening tool or rehabilitation exercise. Many of these tasks are not covered in the literature, particularly single-leg tasks, which have been linked to non-contact injury. Therefore, examining the correlation between different types of single-leg tasks should be done in order to facilitate understanding the causes and contributing factors that may lead to lower-extremity injury, and to understand how individuals use joint biomechanics to meet the demands of these sport tasks.

2. Literature review

2.1 Knee stability

Joint stability plays an essential part in movement of the body, especially for athletes (Williams, Chmielewski, Rudolph, Buchanan, & Snyder-Mackler, 2001). The knee joint lies between the two longest lever arms in the body and is surrounded by the most powerful muscles (quadriceps and hamstring). It is subjected to large forces and moments during activity (Williams et al. 2001) Therefore, keeping the knee joint stable may help to reduce injury. Riemann and Lephart (2002) define joint stability as a joint's ability to hold its normal position, remain steady or control movement under a different surrounding force in either static or dynamic position.

2.1.1 Static Knee Stability

The femur, tibia, fibula and patella articulate in different combinations to form the knee joint. While there are different joints that form the knee-joint complex, the proximal tibiofibular joint, which is an articulation between the head of the fibula and the tibia, has no significant function in relation to the knee joint but is involved in all ankle activities. However, its hypomobility may result in knee pain (VanDijk & Hermens, 2006). The patellofemoral joint (PFJ) mainly functions to improve the efficiency of quadriceps contraction and thus, indirectly, the tibiofemoral joint's movement, particularly in the last 30º of extension. It functions as a guide for the quadriceps tendon and as a bony shield for the femoral condyle cartilage. Moreover, it helps to control knee capsular tension (VanDijk & Hermens, 2006). The actual knee joint is the tibiofemoral joint. It is the biggest joint in the body and functions as a modified hinged synovial joint. Although flexion and extension are its primary motions, tibial rotation also occurs within this joint (Arnheim & Prentice, 2000). The screw home mechanism, which is rotation of the tibia on the femur during knee extension and flexion, offers the majority of knee stability in full extension (static stability) (Arnheim & Prentice, 2000).

Furthermore, the medial meniscus is C-shaped. Its posterior portion is thicker than its anterior portion, while it is roughly equal in the O-shaped lateral meniscus. These menisci, located on the tibial plateau, contribute to static knee stability by functioning as a cushion to reduce stress on the knee (Arnheim & Prentice, 2000). However, it seems that static stability is not offered only by these components. In addition to the congruency of the femur and tibia, which provide the majority of stability at the knee joint, ACL may offer a large amount of knee static stabilization as its function is to protect the femur from posterior translation during weight-bearing activity and extreme internal rotation of the tibia (Michelle, 2007).

2.1.2 Dynamic Knee Stability

Williams et al. (2001) define dynamic joint stability as the ability of a joint to balance external load during activities in order to remain or return to the correct position. Riemann and Lephart (2002) explain the components that may play an important role in joint stabilization. These components are: joint structure, muscles and soft tissues, which hold bones together and act as joint guidance through the appropriate range of motion (ROM). However, there are other components that may contribute to knee stability, such as the shape, orientation and functional properties of the meniscus and condyles, which may all improve joint harmony. This harmony may give extra stability to joints (Kakarlapudi and Bickerstaff, 2000). Further, proprioception may contribute to dynamic joint stability. Improvement in proprioception increases the ability of muscles around the joint to respond appropriately to applied force. Poor proprioception limits functional ability. Its interaction with muscle strength relates to functional ability (Van der Esch et al., 2007) and its improvement increases joint position sense in professional female handballers (Panics et al., 2008). However, there is a little evidence for this, as another study found no difference in functional ability between participants with good and poor joint position sense (Bannell et al., 2003).

Most knee injuries are non-contact injuries and frequently occur during cutting, landing and squatting manoeuvres (Renstrom et al., 2008; Kakarlapudi and Bickerstaff 2000), as the joint is exposed to a large amount of force, so a decline in knee-stability component work may occur and result in the joint becoming unstable.

Claiborne et al. (2006) hold that knee motion control can be achieved by the association of three stabilizing mechanisms. These mechanisms are: tibio-femoral contact, static and dynamic stabilities (passive and active restraint systems). The active restraint system can be explained as muscles that work to control and/or produce motion and proprioception, while passive restraint refers to capsules and ligaments. Hewett et al. (2005) suggest that a combination of active muscle force and passive ligament restraint can give dynamic stability to the joint as a result of load-sharing between them, because the knee is stabilized by synchronized work done by soft tissues, dynamic muscle force and outside load (Schipplein & Andriacchi, 2005).

Kakarlapudi and Bickerstaff (2000) report that the stability of the knee joint is mainly achieved by four ligaments: which are the anterior cruciate ligament (ACL), posterior cruciate ligament (PCL), lateral collateral ligament (LCL) and medial collateral ligament (MCL). The ACL crosses the knee joint just laterally to the inner surface of the lateral condyle and attaches in front of the tibia (Arnheim & Prentice, 2000). It consists of anteromedial, intermediate and posterolateral fibrous bands. The anteromedial bands become taut during flexion while the posterolateral band becomes taut during extension (Arnheim & Prentice, 2000). The primary function of the ACL is to protect the femur from posterior translation and extreme internal rotation of the tibia during weight-bearing activity. However, Kakarlapudi and Bickerstaff (2000) suggest that anterior tibial displacement is mostly restrained by the ACL. It also functions as a secondary restraint for varus and valgus stress (Arnheim & Prentice, 2000).

The PCL, which is stronger, originates from the posterior surface of the tibia, travels upwards and attaches to the anterior medial condyle of the femur (Arnheim & Prentice, 2000). During ROM, PCL fibres are taut to protect the knee and femur from hyperextension. This is supported by Kakarlapudi and Bickerstaff (2000), who suggest that the primary restraint for tibial posterior draw is the PCL.

The LCL supports the knee laterally against varus forces by relaxing in flexion and being taut in extension (Arnheim and Prentice 2000), thus restraining knee abduction (Kakarlapudi and Bickerstaff 2000). The MCL provides medial knee-joint stabilization. Its fibres support the knee against valgus and external rotating forces, as different portions of them are taut at different points of the ROM (Arnheim & Prentice, 2000). Similarly, Amis et al. (2003) suggest that knee valgus is restrained by the MCL. Nonetheless, it seems that it is not only these ligaments that contribute to knee stability as the meniscofemoral ligaments make a significant contribution to resisting tibial posterior draw, and thus to posterior knee stability (Amis et al., 2006). Moreover, when the knee joint is flexed during weight-bearing activities, the surrounding muscles, including muscles that cross the knee and hip joints, are activated to produce more knee stability (Ross, 1997). However, Walsh, Boling, McGrath, Blackburn, & Pauda. (2012) recently found a correlation between muscle activation and slight knee flexion only during a jump-landing task.

Clark (2001a) suggests that overactivity or tightness of one muscle may weaken its antagonist muscle, which may result in an imbalance between agonist and antagonist. Many movement dysfunctions, which place the joint in a high injury-risk position, may be caused by muscle weakness (Clark, 2001a). One of the lower extremity dysfunctions is knee valgus. Weakness or inadequate strength of the hip muscles is widely thought to be one of the main causes of excessive knee valgus, particularly during dynamic movement, such as a forward lunge or squatting (Zeller et al., 2003; Fredericson et al., 2000; Ireland, Willson, Ballantyne, & Davis, 2003; Homan, Norcross, Goerger, Prentice, & Blackburn, 2013; Ireland, Durbin, & Bolgla, 2012). In athletes, it is reasonable to expect that they have enough lower-limb strength; however, Clark (2001a) and Barry (2012) suggest that muscle strength with poor or abnormal neuromuscular control does not help to prevent dysfunctional movements. This abnormal neuromuscular control has been suggested as a main reason for ACL injuries in female athletes (Hewett et al., 2002; McLean et al., 2004a).

Recently, Petrigliano et al. (2012) found that a large posterior tibial slope may improve dynamic knee stability, particularly in the sagittal plane, suggesting that posterior tibial slope should be considered when treating ACL and PCL injuries. Moreover, LaPrade et al. (2010) call the popliteus tendon the ‘fifth ligament’ of the knee and state that it plays an important role in both static and dynamic knee stability. However, Thaunat et al. (2014). criticised the dissection procedure of this study as it dissociates the popliteus muscle-tendon unit from other structures of the posterolateral corner, which are not included. Some of these structures play important roles in static and dynamic knee stability (Thaunat et al., 2014).

Neuromuscular control can be defined as the unconscious ability to respond to joint movement and loading to maintain a functional joint (Lephart et al., 2002). It is the ability to coordinate muscle activity to produce controlled movement (Williams et al., 2001). Some sport skills may put joints, particularly the knee, under a high load. This load may exceed static stabilisers’ ability to maintain joint stability, which then requires an extra stabilizing mechanism to keep ligament strain within a safe limit. The stability produced by muscle contraction may not be enough and agonist-antagonist muscle contraction may increase joint compression and thus enhance joint stability. The high load on the joints during functional tasks forces the lower extremities to rely on a dynamic restraining mechanism (Wikstrom, Tillman, Chmielewski, Borsa, 2006).

2.2 Definition of sport injury

Many definitions of sport injury are found in the literature. It is, therefore, difficult to make comparisons between studies. These differences can be attributed to differences in the methodological approaches for data collection and analysis. Therefore, it is essential to settle on standardised approach and means of analysis.

National Athletic Injury/Illness Reporting Systems (NAIRS) define a sport injury as any injury that occurs during sport participation and leads to absence from participation for at least one day (Junge et al., 2004). However, loss of participation may not be an accurate standard as there are some injuries that may not lead to such loss. Therefore, Fuller et al. (2006) report a consensus statement, regarding the definition of injury in soccer, as any physical complaint resulting from participation in a soccer match or training session without considering the need for medical care or loss of participation. Similar definitions are reported for tennis and other sports (Pluim et al., 2009; Timpka et al., 2014). Although Fuller et al. (2007) present such a definition for a rugby injury, Brooks, Fuller, Kemp, & Reddin (2005) state that a rugby injury should result in more than 24 hours' time loss. However, the definition of a sport injury according to the International Ski Federation (FIS) specifies that an injury should require medical care (Flørenes, Nordsletten, Heir, & Bahr, 2011).

In a 2006 consensus statement, an overuse injury was defined as an injury caused by repeated microtrauma, not a single identifiable event (Fuller et al., 2006). However, the application of this definition to technical sports (e.g. tennis and weightlifting), which include repeated movements, is questionable (Bahr, 2009). A recent study defined an overuse injury as any injury that occurs due to repetitive submaximal loading and that does not allow adequate recovery and structural adaptation. It can affect muscles, tendons, bones, bursas and the neurovascular system (DiFiori et al., 2014).

On the other hand, a re-injury has been defined as an injury that occurs in the same part of the body, of the same type as the previous one, after an athlete fully returns to sport participation (Fuller et al., 2006; Fuller et al., 2007; Junge et al., 2008; Pluim et al., 2009). Moreover, the incidence of injuries is a key factor in sport-injury research, particularly the "sequence of prevention" (Frisch, Croisier, Urhausen, Seil, & Theisen, 2009), which is defined by Rothman (2012) as the number of athletes who have a particular injury divided by the length of exposure. The most popular method to express the incidence of an injury is the total number of injuries per 1,000 hours of exposure (Fuller et al., 2006; Fuller et al., 2007)

The severity of an injury is linked to the definition of injury. Six criteria have been established to describe the severity of an injury. These are: nature of the injury, nature and duration of treatment, time lost, permanent damage and cost. However, most sports-medicine research only considers the sporting time lost. For example, the NAIRS system divides injuries into minor (1–7 days), moderate (8–21 days) and serious (over 21 days) (Fuller et al., 2006; Fuller et al., 2007). However, recent research for team and individual sports divides it into minor (1–7 days), moderate (8–28 days), severe (29 days – 6 months) and long-term or career injuries (more than 6 months) (Pluim et al., 2009; Flørenes et al., 2011; Timpka et al., 2014; Junge et al., 2008).

2.3 Knee injury in sports

Knee injury is one of the most common injuries in most sports, particularly sports that involve landing, jumping and changes in direction (Swenson et al., 2013; Hootman, Dick, & Agel, 2007). Compared to other injuries to other body regions, the knee is the second most commonly injured joint during both practice and competition (Shea, Grimm, Ewing, & Aoki, 2011). Knee injuries can affect articular cartilage homeostasis and lead to early osteoarthritis (OA) (Murrell, Maddali, Horovitz, Oakley, & Warren, 2001) and time loss from sport (Starkey, 2000). Patellofemoral pain syndrome can reduce the ability to perform daily activity and sport tasks (Weiss & Whatman, 2015). ACL injury may also lead to a failure return to the same level of competition (Starkey, 2000). A cohort study by Ardern et al. (2011) found that only 33% of 503 patients were able to return to the same level of competition at 12 months after anterior cruciate ligament reconstruction (ACLR).

There are different factors associated with ACL rupture. They can be categorized into intrinsic and extrinsic factors. Examples of intrinsic factors are joint laxity, gender, femoral notch size, limb alignment, hormonal changes and knee flexion angle during landing (Ramesh, Von Arx, Azzopardi, & Schranz, 2005; Vaishya & Hasija, 2013). Examples of extrinsic factors are type of shoes, playing surface, weather, level of play, training method and rules of the game (Oestergaard Nielsen, Buist, Srensen, Lind, & Rasmussen, 2012; Aoki et al., 2010; Smith et al., 2012). So far, no evidence for the superiority of one of these factors over others has been reported. It is accepted that multiple factors are likely to interact to cause an ACL injury, though a clear illustration of multifactorial injury is still to be created (Shultz et al., 2012; Arendt, Bershadsky, & Agel, 2002). A recent conceptual study by Bittencourt et al. (2016) criticises a reductionist view of injury aetiology, i.e. simplifying the complex system of injury into units (isolated risk factors) and dealing with them separately, and individual factors' ability to determine the risk of injury. In their paper, a complex model was proposed for sport injury. The underlying theory of this model is that its interacting factors lead to a web of determinants that interact with each other in unpredictable ways, which in turn form a pattern of injury or adaptation. They then suggest abandoning isolated risk factors and turning to risk pattern recognition as this may improve established injury prediction and prevention. But much has still to be understood about how individual factors generate risk, prior to understanding their interaction.

The terms patellofemoral pain syndrome (PFPS) and anterior knee pain are often used interchangeably. The development of PFPS has been linked to many factors, such as decreased knee-flexion angle, increased vertical ground reaction force (VGRF), decreased quadriceps and hamstring strength and increased hip-external rotator strength compared to uninjured individuals (Boling et al., 2009; Emami, Ghahramani, Abdinejad, & Namazi, 2007; Powers, 2010). However, there is no agreement between studies regarding the relative importance or significance of these factors. For example, Emami et al. (2007) found a greater Q-angle for women who were diagnosed with PFPS compared to a control group, while Park and Stefanyshyn (2011) suggest that a greater Q angle many not be a risk factor for PFPS as they found a negative correlation between it and peak knee abduction moment. Incompatible findings were also obtained for onset timing of the VMO muscle (Chester et al., 2008). Furthermore, there are some other factors that are suggested as being related to PFPS, such as pelvic tilt and femoral anteversion, which have been found to influence the magnitude of the Q-angle (Nguyen et al., 2010). However, many of the aforementioned studies are case-controlled and do not provide a comprehensive overview of all the factors. This has led to a lack of classification of factors related to PFPS, and thus potentially poor outcomes in rehabilitation of PFPS (Lankhorst, Bierma-Zeinstra, & van Middelkoop, 2012a).

2.4 Incidence of and gender differences in ACL injury and PFPS in sports

Increasing numbers of participants in sports results in increased numbers of injuries, particularly in females, as the number of female participants increased more than nine-fold after implementation of Part IX of the Educational Assistance Act, while male participation increased by only around 3 per cent (Hewett et al., 2005). Many studies have been conducted to examine the incidence of injuries in different types of sports. Studies that compared injury rates in males and females participating in similar sports show conflicting results. Some studies found that the overall rate of severe injury was similar for both genders (Hagguland, Walden, & Ekstrand, 2009; Sallis, Jones, Sunshine, Smith, & Simon, 2001). Other studies reported higher injury rates in men than in women (Layde, Laud, Guse, & Hargarten, 2005; Conn, Annest, & Gilchrist, 2003; Dempsey et al., 2005). However, the majority of previous studies found that females had higher rates of injury than males (Boling et al., 2010; Robinson & Nee, 2007; Waldén, Hägglund, Werner, & Ekstrand, 2011a; Myer et al., 2010c; Fuller et al., 2007; Mihata et al., 2006) and the mean age for females who sustained injury was lower than for males.

One idea was that the lack of experience of female soccer players might be the reason for them having a higher rate of injury. However, there is no strong evidence for this and this idea has been disproved since the increase in female participation in sports (Muffy et al., 2015). Therefore, other suggestions are that females tend to have higher ligamentous laxity, greater Q-angle, larger patellar tendon tibia angle, smaller ligament size and narrower intercondylar notch (IN) than males (Jaiyesimi & Jegede, 2009; Ebeye, Abade, & Okwoka, 2014; Shultz, Sander, Kirk, & Perrin, 2005). Some of these factors are discussed in depth in section 2.5.2.

Hootman et al. (2007) analysed 16 years of data from the National Collegiate Athletic Association's (NCAA) injury surveillance. By combining data, it was found that football players sustain the highest numbers of ACL injuries with more than 50 per cent of the total, followed by basketball players (10%), volleyball players (3%) and wrestlers (3%), while the lowest numbers of ACL injuries were reported for ice hockey and baseball (1.16%). However, female gymnastics and soccer reported the highest rates of ACL injuries (0.33 per 1,000 hours of athlete exposure). The average annual rate of ACL injury also increased by 1.3% over the study period. However, the definition of "injury" in this study was not determined. In contrast, a meta-analysis found that ACL tear incidence was higher in basketballers, followed by soccer players and lacrosse players

(Prodromos, Han, Rogowski, Joyce, & Shi, 2007). Injury in this study was determined as a “tear” of the ACL. However, it was not clear if this study included partial tears or not.

Researchers noted that females participating in sports are at high risk of ACL injury. In a study that evaluated 12 years of injury data, Agel et al. (2005) used the NCAA Injury Surveillance System to evaluate ACL injuries in collegiate soccer and basketball players (both genders). They found that female soccer and basketball players had more ACL injuries compared to males (Mountcastle et al., 2007).

Many studies have also reported that females have more non-contact ACL injuries than males participating in the same sport, often by a factor of 2–8 times (Waldén et al. 2011a; Prodromos et al., 2007; Gianotti, Marshall, Hume, & Bunt, 2009). Mountcastle et al. (2007) examined 10,419 (86.6% male) students who graduated between 1994 and 2003 and found that 34.8% of ACL injuries occurred in football, 51.2% in rugby and 8.3% in basketball, while 17.6%, 13.7% and 9.8% of female participants had ACL injuries from basketball, gymnastics and soccer, respectively. However, unequal numbers of male and female subjects may have affected the results. Although many researchers agree that females participating in sports that include high-risk movement are more prone to non-contact ACL injury than males, examination of gender differences in ACL injury in different sports and with varying levels of participation, while considering other factors such as age, experience and level of participation, is needed. Finally, a higher ACL injury rate was reported during match play than training in most studies (Waldén et al. 2011b; Le Gall, Carling., Reilly, 2008).

With regard to PFPS, it mainly affects both adolescents and young adults who participate in cutting, jumping and pivoting sports (Heintjes et al., 2009; Willson & Davis, 2008). It may affect about 22/1000 persons per year, and it is present in more females than males, particularly active young people and adolescent and adult females (Boling et al., 2010; Robinson & Nee, 2007). Myer et al. (2010c) examined the incidence of PFPS in middle- and high-school female athletes (n = 240) during the competitive season. The incidence of unilateral PFPS was 9.66% at the beginning of the season. However, many factors that may contribute to developing PFPS, such as hormonal and anatomical factors, were not controlled in this study. Devereaux and Lachman (1984), in an epidemiological study, reported that PFPS accounted for 25% of all knee injuries. Of PFPS patients who attended a sports injury clinic, 30% related their pain to running (short and long distance).

This could be because of the high load applied to the PFJ. Twenty per cent of PFPS patients were footballers and rugby players. This means that the duration of exposure may play an important role. Netball and weightlifting had the lowest rates at less than 5% of all PFPS patients. PFPS was defined in this study as anterior knee pain, while the symptoms the patients had varied, such as knee-locking, giving way, clicking knee and other symptoms. This suggests that the accuracy of diagnosis is questionable and, thus, the validity of the findings.

The incidence of PFPS has been examined in different groups. Among all of them, women showed a higher incidence rate than men (Boling et al., 2010; Calmbach & Hutchens, 2003; McGuine, 2006). The ages of 70% of PFPS patients were between 16 and 25. Additionally, Boling et al. (2010) concluded that females are 2.23 times more likely to develop PFPS than males. Such a conclusion was drawn from a large sample size study (n = 1520), which included follow up for 2.5 years. However, all participants were from one academy, which may not represent the general population.

PFPS is reported to be 15% higher in females than in males (Boling et al., 2010). A recent study examined adolescent basketballers and confirmed the findings of the aforementioned studies. They found that 26% of females compared to 18% of males developed PFPS (Foss, Myer, Magnussen, & Hewett, 2014). However, these studies allowed for overestimation, as they investigated data that included past and current cases. Other researchers who examined the incidence of PFPS using the most common measure of incidence (incidence proportion) in a military population which has greater demands did not examine gender differences (Wills, Ramasamy, Ewins, & Etherington, 2004; Jordaan & Schwellnus. 1994). Only Boling et al. (2010) reported that military females were 25% more likely to develop PFPS. However, this was statistically insignificant. Although it is common for females to have a high incidence of PFPS, and current studies support this discrepancy between the genders, there is a shortage of epidemiological data on this condition (Boling et al., 2010), leaving the incidence of PFPS in the general population unknown (Rothermich, Glaviano, Li, & Hart, 2015). With regard to prevalence, most studies were conducted a long time ago (Devereaux & Lachman 1984; DeHaven & Lintner 1986), apart from Myer et al. (2010c) who found that the prevalence of PFPS in female athletes was 16.3% at the beginning of the competitive season. Therefore, updated data are needed to support or refute these findings.

2.5 Risk factors for noncontact ACL injury

Several potential risk factors for ACL injury have been identified. Researchers have divided these into two categories: extrinsic and intrinsic factors (Murphy et al., 2003). Although there is another scheme that divide these factors into environmental, hormonal, anatomical and neuromuscular (Griffin et al., 2006), they still fall under the first scheme. Therefore, these factors will be discussed as extrinsic (external to personal) and intrinsic (related to personal) factors. However, greater focus will be on intrinsic factors as they are more closely related to the current project.

2.5.1 Extrinsic factors

Extrinsic risk factors include footwear, type and condition of playing surface, and weather. These factors mainly influence shoe-surface interaction, which increases the load on the knee joint and its musculature as a result of increased friction force (Aoki et al., 2010; Sterzing, Müller, & Milani, 2010). This is considered a relevant risk factor (Smith et al., 2012). However, other factors should be considered when examining this scenario, such as foot posture, impact force, technique and player position, and joint mobility.

The rules of sports are another modifiable extrinsic risk factor of injury and thus supervision or changing them based on suitable injury surveillance data is important to avoid or reduce the risk of injury (Dick et al., 2007). An example of a rule change that provides evidence of how to reduce injury is the implementation of a rule change (of 10 metre outer centre circle in which ruck-men must be positioned at centre bounces) that resulted in a significant reduction in PCL injury in Australian football. Also, the implementation of substitute rules in 2011 led to reduced hamstring and groin injuries in the Australian football league (Orchard, McCrory, Makdissi, Seward, & Finch, 2014). However, a rule change may unwittingly increase other injuries or remove an important part of the game. Also, it depends on appropriate implementation of a rule change and the compliance of coaches, players and referees. Moreover, because rule changes are based on personal experience, future changes should be based on sport-specific studies and their impact needs to be assessed accurately (Tucker, Raftery, & Verhagen, 2016; Mtshali, Mbambo-Kekana, Stewart, & Musenge, 2010; Finch, 2006).

Training load is another extrinsic risk factor that can be modified and controlled by players or coaches (Oestergaard et al., 2012). Rugby training injuries happen more frequently when training intensity and duration are high (early stages of the season) (Gabbett, 2003), and the highest incidence of injuries in semi-professional rugby players is recorded at the end of pre-season, when the training load is at the maximum (Gabbett, 2004), while a less than adequate training load may lead to failure to reach the required level of physiological development (Gabbett & Domrow, 2007). Effective pre-season training will result in athletes' peak level of physical readiness as the season starts (Buchheit et al., 2013). However, what constitutes an effective training load or load-change rate remains unclear.

Injury odds and illness were reduced in a pre-season high training load group compared to a low training load group (Veugelers, Young, Fahrner, & Harvey, 2016). This suggests that training load is a crucial component of injury prevention. However, Harrison and Johnston (2017) examined the relationship between training load and injury in sixty sub-elite Australian rules footballers. The highest injury rates (0.52–0.63) were found in players with a preseason training load of less than 1250AU/week (AU = arbitrary unit which in this case is rate of perceived exertion multiplied by time). Therefore, they suggest that a high training load is not responsible for injury but is required to increase the level of fitness. Also, they found that more than a 4000AU training load in a 2-week period significantly increased the risk of injury in the following week. Therefore, they suggested that a training load of more than 2000AU over several weeks may increase the risk of injury.

2.5.2 Intrinsic factors

2.5.2.1 Anatomical factors

2.5.2.1.1 Q-angle

Q-angle was coined by Brattström (1964) as the angle between the pull line of the quadriceps and the line of the patellar tendon, and it is used to measure the tendency of the patella to move laterally when the quadriceps contracts. This tendency is in proportion to the angle (Fredericson & Yoon, 2006). However, the quadriceps appear to have a more lateral alignment than that of the patella. This may make the relationship between Q-angle and knee injury unclear. Q-angle has been suggested as contributing to non-contact ACL injury as it alters the kinematics of the lower limbs (Mizuno et al., 2001). There is a consensus in the literature that larger Q-angles are observed in athletes who have sustained ACL injury compared to those who have not. Therefore, Q-angle has been suggested as being a risk factor for non-contact ACL. However, there is little evidence to support this (Nguyen et al., 2010). Shambaugh, Klein, & Herbert, (1991) examined the relationship between lower extremities alignment and knee injury in 45 basketballers. The 14 who Fourteen of them sustained ACL injury had higher Q-angles than the rest of the cohort. However, it is not clear whether Q-angle has a direct effect on injury or if it is indirect via integration with other risk factors. Buchanan (2004) examined the possibility of age, gender and level of experience predicting Q-angle in fifteen healthy male and female basketballers. Prepubescent players (both genders) showed increased dynamic knee valgus during landing compared to peripubescent and postpubescent. With postpubescent subjects, females have increased dynamic knee valgus while males show more knee varus. Buchanan (2004) suggests that Q-angle may predict 32 per cent of knee valgus-varus position. Pantano et al. (2005) examined 20 subjects to see whether a large Q-angle group would exhibit large knee valgus during SLS compared to a small Q-angle group. There was no difference between the groups. However, differences in methods to measure Q-angle may produce different results. Therefore, a standardized, reliable and valid method needs to be established. Mohamed, Useh, & Mtshali, (2012) examined 24 female soccer players who were divided into injury and control groups. Q-angle, pelvic width and intercondylar notch width were measured and showed no differences between the groups. The authors suggest that these variables cannot be used to predict ACL injury. However, Q-angle was measured using a goniometer, which does not seem to be the best way to measure it. Conflicting results suggest that these factors do not independently influence injury, rather the effect is through being linked to other ones. However, a correlation between these factors is not established in the literature. Some studies have examined gender differences of Q-angle. All of them report larger Q-angles for females compared to males (Jaiyesimi, & Jegede, 2009; Ebeye et al. 2014), which may explain the gender differences in ACL injury and PFPS.

2.5.2.1.2 Knee flexion angle

It has been suggested that the high loading that results from abnormal knee movement in the sagittal plane may damage the knee structure (Quatman et al., 2010). Non-contact ACL injury appears to be influenced by knee flexion angle and studies have reported a reduced ACL load when the knee flexion angle increases (Dai, Mao, Garrett, & Tu, 2014). Many observational studies state that the knee flexion angle ranges from 0–30 degrees, observed when ACL injuries occur (Cochrane, Lloyd, Buttfield, Seward, & McGivern, 2007; Krosshaug et al., 2007; Olsen et al., 2004). Quadriceps muscle contraction causes an anterior shear force to the proximal tibia via the patellar tendon (DeMorat, Weinhold, Blackburn, Chudik, & Garrett, 2004) and may reach a level that could be enough to cause ACL microtrauma when knee flexion is between 10–30 degrees (Griffin et al., 2000). Also, the quadriceps muscle acts eccentrically to control knee flexion during dynamic tasks, hence researchers suggested a relationship between reduced knee flexion angle and a weak quadriceps (Lephart et al., 2002).

Landing with a smaller knee flexion angle increases the risk of injury by increasing GRF and reducing energy absorption (Schmitz, Kulas, Perrin, Riemann, & Shultz, 2007). However, the actual underlying mechanism during landing is unclear. Recently, Fisher et al. (2016) examined 18 female and 18 male recreational athletes and found no relationship between force production during isometric squats and different knee flexion angles (40°, 60°, 80°, 100°) during landing. However, such a correlation was found when female participants were analysed separately, which contribute to sex differences in ACL injury rates.

Knee flexion angle affects the patella tendon-tibia shaft angle, ACL elevation angle and ACL loading. When knee flexion decreases, the patella tendon-tibia shaft angle and anterior draw force applied to the proximal tibia increase, which increases ACL loading and the risk of injury (Lin et al., 2012). Nonetheless, some researchers deny the theory of a single-plane injury mechanism (Quatman et al., 2010; DeMorat et al., 2004) and content that knee flexion angle may not predict ACL injury (Hewett et al., 2005). Moreover, isolated sagittal-plane force was found not to be enough to damage the ACL (McLean, Huang, & van den Bogert, 2008). It seems that knee-flexion angle does not directly, or on its own, cause ACL injury; rather, it adds secondary additional stresses to other risk factors. Recently, Favre, Clancy, Dowling, and Andriacchi, (2016) conducted a study to examine the effect of modifying knee flexion angle on other risk factors during jump landing. Thirty-nine recreational athletes were examined in this study and the findings show that increasing knee flexion angle reduces GRF and knee flexion moment. This agrees with the study of Nagano, Ida, Akai, and Fukubyashi (2011), who reported that an increase in knee flexion angle following participation in a jump-and-balance exercise reduces ACL strain and, in turn, the risk of injury. This may support the fact that females report more ACL injuries and reduced knee flexion angle during sporting tasks (Sigward et al., 2012). The findings of Pollard, Sigward, and Powers (2010) may explain this, as they found that decreased knee flexion during landing may increase frontal-plane angles and moments. Although GRF and knee-flexion moment are a risk factor for

ACL injury, other critical risk factors, such as knee-valgus angle and moment, are not affected by changing the knee-flexion angle. Moreover, Nagano et al.'s (2011) study only included eight female subjects. This suggests that such findings need to be confirmed on a larger scale, and both genders and knee-flexion angle as a risk factor for knee injury need more investigation.

2.5.2.1.3 Frontal plane movement

The majority of current studies focus on knee frontal-plane movement and its association with ACL injury. Overall, studies are divided into supporters of frontal plane as a single-risk factor of ACL injury, and supporters of frontal plane in combination with other movements increasing the injury risk. Excessive knee frontal-plane movement has been suggested as a risk factor of knee injury. Knee-valgus collapse with slight knee extension (0–30 degrees) and a tibia externally rotated while the foot is on the ground have been identified as an injury position during dynamic movement (Krosshaug et al., 2007; Boden et al., 2000; Olsen et al., 2004). Greater knee-valgus angles and moments were demonstrated by women compared to men during landing activities (Kernozek et al., 2005), suggesting an increased risk of excessive frontal-plane motion and ACL injury. Moreover, Shultz et al. (2007) found that knee-abduction/-adduction load causes frontal-plane knee rotation. This has been found to increase ACL tension (Miyasaka, Matsumoto, Suda, Otani, & Toyama, 2002). Similarly, external knee-abduction moment is reported to apply a large force to the ACL (Hewett et al., 2005). However, Yu and Garrett (2007) suggest that ACL is not the main structure that receives the highest load during valgus load. Therefore, they suggest that knee-valgus collapse may not be associated with an isolated ACL injury.

Using magnetic resonance imaging (MRI), post-ACL injury patients show bone bruising in the lateral compartment of the distal femur (Nishimori et al. 2008; Nakamae, Engebretsen, Bahr, Krosshaug, & Ochi, 2006). The location of these bruises may suggest the presence of high impact forces on the proximal tibia and distal femur. This supports the hypothesis that frontal-plane movement, particularly knee-valgus loading and tibial internal rotation, may correlate with ACL injury (Shin, Chaudhari & Andriacchi, 2011). However, at least 1,200N (for females) or 1,500–2,000N (for males) of force is needed to cause damage to the ACL (Chandrashekar, Mansouri, Slauterbeck, & Hashemi, 2006). The greatest strain on the ACL results from anterior tibial shear but this cannot reach the level of causing ACL rupture (McLean et al., 2004a). Therefore, it seems probable that more than one excessive movement is required to generate enough force to tear the

ACL. For example, an anterior tibial shear force combined with knee-valgus and/or rotational moments increase the strain on the ACL significantly, which raises the possibility of ACL injury. This applies more at angles nearer to knee extension and supports the mechanism suggested for ACL injury (McLean et al., 2004a). Although researchers have shown some evidence for a relationship between knee frontal-plane motion and ACL injury, the exact biomechanics within this plane of motion that lead to ACL injury are not clear.

2.5.2.1.4 Intercondylar notch

The ACL passes through the intercondylar notch (IN) of the femur. Palmer (1983) first suggested a relationship between IN and ACL injury. This was later supported by Souryal, Moore, and Evans, (1988), using radiography to show a significant correlation between bilateral ACL tear and IN width. However, there was no difference in IN width between a normal group and an ACL tear group. Researchers have suggested that ACL might be impinged in some knee positions due to either a narrow IN or a large ACL size (Griffin et al., 2006). Other studies have also reported a relationship between IN width and risk of ACL injury (Fernandez-Jaen et al. 2015; Domzalski, Grzelak, & Gabos, 2010; Sonnery-Cottet et al. 2011; Hoteya et al. 2011). Recent studies have examined the relationship between IN and the risk of ACL injury by comparing three groups (unilateral ACL injury, bilateral ACL injury, healthy control). Using MRI imaging, they found that IN width was narrowed in both ACL injury groups compared with the control group. Also, there were differences between injured and uninjured legs in the unilateral ACL group. This suggests a relationship between IN width and risk of ACL injury (Görmeli et al., 2015). This relationship may be due to the amplified impact force between the medial wall of the femoral condyle, which leads to ACL abrasion (Fernandez-Jaen et al., 2015; Geng et al., 2016). However, the aforementioned studies did not consider the size of the ACL itself. Narrowing of the IN does not mean that injury is as a result of impingement as the size of the ligament might be appropriate for the size of the IN. Smith et al. (2012) suggest that both the size of the IN and the volume of the ACL should be considered. In contrast, others deny such findings. Lombardo, Sethi, & Starkey (2005) conducted a case-controlled study based on radiography of the knee. They prospectively followed 305 professional basketballers for a period of 11 years and concluded that there is no association between IN width and ACL injury. Similarly, studies have compared ACL athletes injured and healthy controls, and unilateral and bilateral ACL injuries with healthy controls. Using direct radiography and MRI, no significant relationship between ACL and IN width was reported (Herzog, Silliman, Hutton, Rodkey, and Steadman, 1994; Schickendantz and Weiker 1993). Also, it is suggested that IN may not be a reliable measure to predict ACL injury. Alizadeh and Kiavash (2008) compared the mean of IN width and found no difference between an ACL group and a healthy group. Discrepancies in the findings may be due to using different methods. Most studies use direct radiography. However, it is well known that soft tissues cannot be measured using simple radiography. Hoteya et al. (2011) suggest that MRI is more reliable and accurate. Using IN width in some studies and IN width index in others may be another reason for findings conflicts. Also, the experience of the radiographer and the position of the subject may play an important role (Görmeli et al., 2015). This is not discussed in these studies.

A recent study showed evidence of the importance of the intercondylar notch in ACL ruptures. This study recruited 308 ACL patients and 222 healthy controls and compared them using MRI. Although there was no difference in the groups' ages, the ACL group showed smaller IN. Females also showed smaller IN, which may explain the more numerous ACL injuries among women (Fernandez-Jaen et al.,2015). On a positive note, this study utilised coronal and axial-plane MRI images to measure IN width, which is considered one of the best procedures to measure IN. The most dangerous scenario for ACL injury was described by Murshed et al. (2007). In addition to impingement of the ACL in IN, a weaker and smaller ACL accompanied by a small IN might be a contributor to ACL rupture risk.

Interestingly, a smaller IN width was reported for female subjects (Chandrashekar et al., 2006; Chandrashekar, Slauterbeck, & Hashemi, 2005; Anderson, Dome, Gautam, Awh, & Rennirt, 2001; Charlton, John, Ciccotti, Harrison, & Schweitzer, 2002) and a difference in IN width between African Americans and white Americans was reported for both genders with a larger IN width for Africans (Shelbourne, Gray, & Benner, 2007). Therefore, sex and ethnic background should also be controlled. Lastly, those who reported a relationship did not explain if it is related to ligament size, impingement or integration with other risk factors.

2.5.2.1.5 Knee laxity

Knee joint laxity has been reported as a risk factor but the evidence supporting this is inconsistent (Griffin et al., 2000). Uhorchak et al. (2003) found that generalised joint laxity is a risk factor for

ACL injury in military cadets. Similarly, Ramesh et al. (2005) found more ACL injuries in those with greater general joint laxity and particularly those with greater knee-joint laxity. Recently, Myer, Ford, Paterno, Nick, and Hewett, (2008a) prospectively examined the possibility of joint laxity to predict ACL injury and concluded that increased knee laxity may contribute to increased risk of ACL. However, this study only included female soccer and basketball athletes. So, the generalisability of the result to male subjects or other sport athletes is questionable. In contrast, one study contradicts this and reported that there was no strong relationship between joint laxity and ligament injury (Huston, Greenfield, & Wojtys, 2000). One hundred ACL tears were analysed and a strong relationship between hamstring flexibility and ACL tears was reported (Boden et al., 2000). These conflicting findings might be due to the different methods used to assess laxity or the level of activity of the target population. Nevertheless, most studies have reported higher joint hypermobility with ACL patients. A summary of these studies is presented in table 2.1.

Table 2.1 : Summary of studies reporting higher joint hypermobility with ACL patients

Author	**Study design**	**Participants**	**Number of subjects**	**Number of injured subjects**	**Significant finding(s)**
Uhorchak et al. (2003)	Prospective cohort	cadets	895	24	Generalized joint laxity for injured subjects ($P = 0.01$)
Ramesh et al. (2005)	Prospective cohort	N/A	169 ACL 65 controls	169	Significant joint laxity and hyperextension for those who both underwent and booked for ACLR compared to controls ($P = 0.01$).
Myer et al. (2008a)	Case-control study	Soccer and basketball players	95 4 match controls	19	Side-to-side difference in knee laxity (P = 0.02)
Vaishya and Hasija (2013)	Case-control study	N/A	210 ACL 55 match controls	210	ACL patients were more likely to have joint hypermobility (odd ratio = 4.46) P value – N/A
Vauhnik et al. (2008)	Prospective cohort	Female basketball, volleyball, and handball players	540	11	Knee hyperextension is a risk factor
Loudon et al. (1996)	Match-case control	Female athletes	20 ACL injured 20 controls	20	Increased recurvatum and navicular-drop subtalar joint pronation are risk factors
Woodford-Rogers et al. (1994)	Match-case control	Footballers and gymnasts	95	19	Anterior knee laxity is a risk factor for ACL

ACL = Anterior cruciate ligament, N/A = Not available, ACLR = Anterior cruciate ligament reconstruction.

Vaishya and Hasija (2013) contend that the control group was not matched for occupation and sport activity and the examiners were not blinded. Also, the hours of exposure were significantly different between injured and uninjured players in the study by Vauhnik et al. (2008). This may affect the results. Only a static position was examined in the study by Loudon, Jenkins, and Loudon, (1996). As injury occurs during movement, risk factors should be examined during a dynamic position. Other risk factors were not controlled in the study of Woodford-Rogers, Cyphert, and Denegar (1994). Moreover, it seems difficult to understand the implications of joint laxity for ACL injury in cases where females show different baseline values from males. Additionally, a recent systematic review suggests that ACL becoming more lax during menstrual cycle particularly pre-ovulatory phase ligamentous laxity is only related to the menstrual cycle (Belanger, Burt, Callaghan, Clifton, & Gleberzon, 2013). Despite these studies' limitations, there are consistent findings that knee laxity is a risk factor for ACL injury.

2.5.2.2 Hormonal risk factors

Female hormones change during the menstrual cycle and vary from one cycle to another (Smith et al., 2012). Researchers suggest that female hormones influence the metabolic process and thus the biomechanical properties of ACL, which makes women more prone to ACL injury (Barber-Westin, Noyes, Smith, & Campbell, 2009; Warren, Panossian, Hatch, Liu, & Finerman, 2001). However, it is not clear how this influence occurs. Moreover, there is no study on humans that proves the presence of either oestrogen and progesterone receptors in the ACL, or the effects of hormone concentrations on ACL properties. All examinations were performed on animal models (Smith et al., 2012). Noteworthy is that women footballers are found to be prone to injury during the premenstrual and menstrual stages more than the rest of the menstrual cycle (Ruedl et al., 2009). Additionally, women aged 15–19 years show about a 20 per cent reduced risk of ACL injury while on an oral contraceptive than a matched control (Gray, Gugala, Baillargeon, 2016). This study used notational insurance company data and had a large sample size. However, it depended on the presence of prescriptions for oral contraceptives, which may not be accurate as some women use them without prescriptions. In contrast, some studies show that the use of contraceptives has no protective effect against ACL injury amongst recreational skiers (Lefevre et al., 2013; Ruedl et al., 2009). Moreover, Samuelson, Balk, Sevetson, and Fleming, (2017) conducted a systematic review and reported that the evidence on oral contraceptives and ACL injuries is limited due to methodological concerns and small sample sizes.

Several studies have examined the relationship between differences in sex hormones and risk of ACL injury (Ruedl et al., 2009; Beynnon et al., 2006; Myklebust et al., 1998; Slauterbeck, Fuzie, Smith, & Clark, 2002; Myklebust et al., 2003). Different methods to identify the phases of the menstrual cycle were used in the aforementioned studies. Therefore, it seems difficult to compare their results and draw a firm conclusion about the effects of sex hormones on ACL injury. Table 2.2 below gives a brief description of these studies.

Table 2.2 : Summary of studies that have examined the relationship between sex hormones and risk of ACL injury

Author	**Design of the study**	**Participants**	**Number of female subjects**	**Number of ACL injuries**	**Time when most ACL injuries occurred**
Myklebust et al. (1998)	Prospective cohort	Handball players	23	17	Day 27 of the cycle
Slauterbeck et al. (2002)	Case-control	20 school, 15 high school, 1 middle school, 2 recreational	38	37	Day 1 and 2 of the cycle
Myklebust et al. (2003)	Prospective cohort	Handball players	69	46	Menstrual phase of the cycle.
Beynnon et al. (2006)	Case control	Alpine skiers			Preovulatory phase of the cycle
Ruedl et al. (2009)	Case control	Recreational skiers	186	93	Preovulatory phase of the cycle

ACL = Anterior cruciate ligament

These studies also used different approaches to identifying ACL injury. Myklebust et al. (1998) gathered injury data from coaches, physiotherapists, physicians, insurance companies and team officials. This may lead to bias. Slauterbeck et al. (2002) confirmed ACL injury by MRI or surgery. Myklebust et al. (2003) considered any knee injury that causes one week or more of missed participation. This may lead to overestimation, as not all of them were ACL injuries. However, their findings are yet to be challenged in the literature. Griffin et al. (2006) reviewed the Hunt

Valley II Meeting, January 2005, and found evidence for the highest rates of ACL injuries occurring during the early and late phases of the menstrual cycle. Most of the studies used a serum, salivary hormone concentration or urine alone, in combination with cycle history, as outcome measures to predict the cycle phase when injury occur. Although it might be the best approach, Smith et al. (2012) suggest that a serum sample should be taken every day from all samples rather for the full cycle before and after injury, which was not implemented in the aforementioned studies.

2.5.2.3 Psychological factors

The psychological aspect is also a risk factor in sport injuries; it may increase incidents through stress levels, or a player's personality or anxiety (Junge, 2000). Stress may increase injury incidence due to a declining focus on sport techniques, increased muscle tension and reduced movement coordination (Alizadeh, Pashabadi, Hosseini, & Shahbazi, 2012). Stress increases the release of adrenaline and the blood flow, which has many effects on muscle contraction and muscles' slow-twitch phase (Nielsen, Savard, Richter, Hargreaves, & Saltin, 1990). Increased muscle tension may make muscles and tendons tighter, thus increasing the injury risk (Mainwaring 1999).

Moreover, life's stresses, anxiety and poor survival rates in young football players increase injuries by 23 per cent (Johnson & Ivarsson, 2011). Steffen, Pensgaard, and Bahr (2009) found that increased life stress leads to a greater possibility of injury in young female football players. Anxiety, e.g. from time pressure, fans and coaches, may increase muscle tension, breathing, urine production and sweating, possibly resulting in decreased body water, dehydration and muscle fatigue. Moreover, anxiety may force players to make more effort, either to correct mistakes or put in better performance, which requires more practice and harder work, more physical and mental effort, and perhaps more risk behaviours. Consequently, more practice and effort may cause fatigue. Fatigue has been shown to cause a decline in physical performance and technical skills, which may increase the injury risk (Rampinini et al., 2009). However, it is difficult to generalise such findings as results differ from one player to another. Also, in addition to different levels of motivation, experienced players may be able to deal with such situations by prioritising and making good decisions and thus decrease the risk of injury (Kucera, Marshall, Kirkendall, Marchak, & Garrett, 2005). Additionally, athletes' awareness and consciousness of situations might lead to boosting their predisposition to non-contact ACL injury (Swanik, Covassin, Stearne, & Schatz, 2007). Such factors might impact on the balance between the three main factors, visual, vestibular and somatosensory connections, which form an integrated complex within the neuromuscular system (Di Fabio & Badke, 1991 as cited in Swanik et al. 2007). Football requires the right focus throughout the match, which might be affected by tunnel vision, excessive respiration or muscle tension resulting from stress. Thus, a player focusing only on the ball might suddenly try to withdraw during a counterattack, and succumb to injury (Cox, 2006). Coaches and parents may unwittingly increase the incidence of injury as some groups see injury as a weakness, thus forcing players to hide injury and continue playing (Bathgate et al., 2002). Pressure from coaches/ parents may put players under great psychological and physical pressure, resulting in reduced quality of play, loss of focus and poor technique, which may increase the injury risk (Timpka, Lindqvist, Ekstrand, & Karlsson, 2005). Yet psychological factors alone may not cause injury. Rather, they increase the risk when other physical reasons, such as muscle imbalance, exist or athletes are placed in injury-threatening situations (Gould, Petlichkoff, Prentice, & Tedeschi, 2000).

2.6 Mechanism of ACL injury

It has been suggested that almost 70–80 per cent of ACL injuries are non-contact in nature, i.e. there is no physical contact with another body (Renstrom et al., 2008). Qualitative analysis of real time videos taken during sport events suggests that the most common mechanism of ACL injury is injury that occurs during an SLL or deceleration movement before changing direction, with the foot firmly in contact with the ground and the injured knee appearing to be flexed by 30° at the time of injury (Krosshaug et al., 2007; Cochrane et al., 2007; Boden et al., 2000; Olsen et al., 2004). Moreover, knee-valgus collapse was reported for female athletes (Krosshaug et al. 2007; Boden et al. 2000; Olsen et al. 2004). Olsen et al. (2004) observed a combination of knee-valgus collapse, with the knee close to full extension and tibial rotation in female handball players, and suggested that such position is the most common position of injury (Figure. 2.1). Such a position increases the load on the ACL beyond it is capacity and leads to injury (Hashemi et al., 2005).

Figure 2.1: Examples of ACL injury positions during different directions of landing on one leg with knee valgus collapse

ACL injury often occurs during SLL. As SLL occurs from different vertical heights and horizontal distances, a recent study examined the effects of these on ACL injury risk (Ali, Robertson, & Rouhi, 2014). Knee kinematic and kinetic data were examined in nine male recreational athletes while they performed SLL from different vertical heights (20, 40 and 60 cm) and horizontal distances (30, 50 and 70 cm). The results showed that increasing both the vertical height and horizontal distance of SLL led to significant increases in GRF, hip-flexion angle and knee power, suggesting a higher risk of ACL injury. Moreover, the change in GRF occurred rapidly and differed from the GRF seen during double-leg landing (DLL). This suggests that the biomechanical profile of SLL is unique, more demanding and is not necessarily comparable to DLL. Such findings were previously reported by Yeow et al. (2010). Although sufficient statistical power was observed in this study, a generalisation of the findings to the general male or female population cannot be made due to the sample size. Also, the effects of the vertical height and horizontal distance of SLL on the frontal plane need to be established. However, this study showed reasonable correlation which may be used as a base for future research and explain the relationship between ACL injury and SLL manoeuvres.

2.7 Risk factors for PFPS

Risk factors for PFPS have been divided into intrinsic and extrinsic factors (Witvrouw, Lysens, Bellemans, Cambier, & Vanderstraeten, 2000). Extrinsic factors include equipment use, the manner of sport practice and sport activity. Intrinsic factors are more related to the individual body. Intrinsic factors are suggested to be modifiable and can be beneficial in the treatment of this condition (Halabchi et al., 2013). Therefore, this part will focus only on the intrinsic factor most suggested.

2.7.1 Patellar malalignment

Patellar malalignment can be defined as abnormal movement of the patella in any plane (Grelsamer, 2005). Abnormal movement at any time during flexion extension cycle results in patellar maltracking. The association of patellar maltracking with PFPS is a debatable concern (Petersen et al., 2014). However, researchers suggest that patellar maltracking plays a crucial role in and might be the origin of PFPS (Fulkerson, 2002; Dixit, Difiori, Burton, & Mines, 2007). Using MRI, patients with PFPS perform SLS with a high lateral tilt of the patella (Draper et al., 2009). Hypermobile patella showed significant correlation with PFPS (Witvrouw et al., 2000) (Figure. 2.2). Using 3D motion capture, Wilson, Press, Koh, Hendrix, & Zhang (2009) investigated the kinematics of the patella in vivo. They compared PFPS patients and a healthy control during standing and SLS. At 90° of knee flexion, the patella spin laterally in PFPS patients and medially in the healthy group. Also, significant lateral translation of the patella was reported for PFPS patients. Moreover, 12 per cent greater lateral displacement was reported in patients with PFJ osteoarthritis compared to asymptomatic controls. However, there was no difference in the lateral patellar tilt angle between groups (Crossley et al., 2009). Other studies have reported larger lateralization of the patella in PFPS patients compared to healthy ones (Salsich & Perman, 2007; Witoński & Goraj, 1999). The differences in these studies were detected at different degrees of knee flexion (0°, 0–15° and 20°). Moreover, the association of this factor with the development of PFPS is not clear as it also has been found in subjects with no knee symptoms (Nissen, Cullen, Hewett, & Noyes, 1998; Johnson et al., 1998). When the patella moves laterally, uneven stress is created on the patella and infrapatellar structure, resulting in PFPS (Wilson, 2007; Elias & White 2004b). Nevertheless, Laprade and Culham (2003) used axial radiographs to evaluate the patellar tilt in PFPS patients and healthy individuals and there were no differences between the groups in loaded and unloaded conditions. These conflicting results might be due to differences in the definition of PFPS and/or due to a lack of clear criteria for PFPS patient classification.

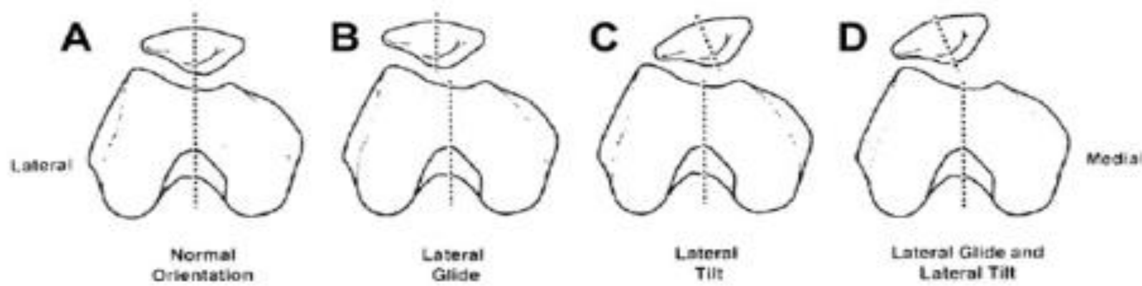


Figure 2.2: Example of patellar malalignment: A: normal position, B: lateral glide, C: lateral tilt, D: lateral glide and lateral tilt

From Collado and Fredericson, (2010)

2.7.2 Hamstring tightness

It has been theorised that hamstring tightness can either cause slight knee flexion during movement or require more quadriceps strength to resist passive strain on the hamstring. Hamstring tightness can increase PFJ reaction force (Piva, Goodnite, & Childs, 2005) and then result in PFPS. Hamstring tightness is an objective sign in PFPS patients and usually represents a target for treatment. However, it is not well supported by primary research (White, Dolphin, & Dixon, 2009).

Limited studies have examined hamstring tightness in PFPS (Patil, White, Jones, & Hui, 2010; White et al., 2009; Witvrouw et al., 2000; Piva et al., 2005). The findings regarding the association of hamstring tightness with developing PFPS are inconsistent. Some studies have reported a significant correlation between hamstring tightness and the development of PFPS (White et al., 2009; Patil et al., 2010). Both studies found significant hamstring tightness in PFPS patients compared to a control group. However, it was not clear whether hamstring tightness was a cause or effect of PFPS, as another study suggested that PFPS may result in hamstring tightness (Piva et al., 2005). Also, the examiners in White et al.'s (2009) study were not blinded to groups selection, which may have led to bias. In contrast, Witvrouw et al. (2000) failed to report any relationship. These inconsistent results might be due to differences in the sample populations and their characteristics and/or the length of period during which subjects developed PFPS. Consequently, the association between hamstring tightness and PFPS development needs more investigation.

2.7.3 Quadriceps tightness

It has been proposed that quadriceps tightness results in high stress on the PFJ. This in turn makes individuals more susceptible to developing PFPS (Post, 2005; Witvrouw et al., 2000). There is little evidence regarding the presence of quadriceps tightness in PFPS patients. Some studies have reported the presence of quadriceps tightness in PFPS subjects (Waryasz & McDermott, 2008; Piva et al., 2005; Fredericson & Yoon, 2006; Witvrouw et al., 2000). Therefore, they consider quadriceps tightness to be a risk factor of PFPS. However, Witvrouw et al. (2000) suggest that reduced quadriceps flexibility is not usually the result of PFPS as it existed prior to developing it. Moreover, Kibler (1987) found that 61 per cent of PFPS patients had tightness of the rectus femoris; nevertheless, no *P* value was reported. Although the study of Waryasz and McDermott (2008) was a high-quality review, they reported only 6 out of 27 PRISMA items and there was no meta-analysis.

Recently, two systematic reviews were conducted and obtained similar findings (Papadopoulos, Stasinopoulos, & Ganchev, 2015; Lankhorst, Bierma-Zeinstra, & Van Middelkoop, 2012b), showing no enough evidence of quadriceps tightness in PFPS patients. Both reviews were of high quality as the former was a meta- review and the latter was a review of randomised control trials (RCTs) and included meta-analysis. So, their findings should be taken seriously. However, the conflicting findings in the literature might be due to the lack of a gold-standard assessment for PFPS. Nowadays, examining the risk factors during functional tests such as SLL and squats is highly recommended and should be considered.

2.7.4 Iliotibial band (ITB) tightness

Through its anatomical correlations with the patella and lateral retinaculum, ITB increases the lateral force vector on the patella, particularly during flexion (Waryasz & McDermott, 2008). Tightness of this structure leads to ITB tightness and results in abnormal patellar tracking and increased stress on the PFJ (Fredericson & Yoon, 2006). ITB tightness and its relation to PFPS has been examined in some studies (Witvrouw et al., 2014; Halabchi et al., 2013; Hudson, & Darthuy, 2009). All of them reported ITB tightness in PFPS patients, which has been suggested alters knee-joint kinematics and increases the load on the PFJ. Hudson, and Darthuy (2009) examined 12 subjects with PFPS and 12 matched controls. They found higher ITB tightness in the PFPS group. Higher ITB tightness was also reported in the non-painful leg in PFPS patients. However, this study might be underpowered because no power calculation was conducted. In contrast, Piva et al. (2005) found no difference between PFPS subjects and age- and gender-matched controls. Though, the assessor was not blind to groups assignment.

It seems that ITB is not directly associated with PFPS, rather than interacting with other affecting factors which are therefore associated with PFPS. However, only one study reported a relationship between ITB tightness and patellar hypermobility (Puniello, 1993). Patellar hypermobility correlates with laxity of the medial ligaments of the patella and is commonly observed to associate with patellar subluxation (Conlan, Garth, & Lemons, 1993). These factors are reported as risk factors for PFPS.

2.8 Mechanism of PFPS

Regardless of the vast number of studies that have focused on PFPS and its root causes, the underlying mechanism is still not fully understood. It has been reported that the PFPS mechanism is multifactorial (Witvrouw, Lysens, Bellemans, Cambier, & Vanderstraeten, 2005). Recently, Song, Lin, Jan, and Lin (2011) conducted a systematic review to identify the potential mechanism of PFPS. The evidence suggests that tracking or lateral malalignment is not consistently associated with PFPS. Despite this, there is general agreement that patellar malalignment on an unstable femur is one of the most prevalent factors associated with PFPS (Petersen et al., 2014; Wilson, 2007; Elias & White, 2004b; Sanchis-Alfonso, Roselló-Sastre, & Revert, 2001). These studies' results suggest that patellar malalignment or maltracking results in high contact pressure on the patellofemoral joint which, over time, may cause PFPS. However, the position of the femur and tibia in relation to the patella which is considered in these studies could also affect patellofemoral joint contact force (Barton, Levinger, Crossley, Webster, & Menz, 2012). The PFJ contact area is reduced during tibial external rotation, hip adduction and internal rotation (Salsich & Perman, 2007; Lee, Morris, & Csintalan, 2003). This reduction in the loading surface leads to improper distribution of the forces around the PFJ and could damage it (Figure 2.3). This was reported in PFPS patients, particularly during dynamic tasks such as walking and squaring (Heino & Powers, 2002). The concentration of a high load over the PFJ could lead to a loss of peripatellar tissue (Dye, Stäubli, Biedert, & Vaupel, 1999). However, it is not clear whether this causes pain or not as it is not an innervated structure (Biedert & Sanchis-Alfonso, 2002). Also, Salsich and Perman (2007) suggested that high PF joint load may damage the articular cartilage but may not cause pain. Meanwhile, other studies have suggested that the source of anterior knee pain is the subchondral bone because of its rich innervation, its response to loads and its relationship with the overlying cartilage (Moisio et al., 2009; Biedert & Sanchis-Alfonso, 2002; Fulkerson, 2002). The results of Farrokhi, Colletti, and Powers' (2011) study support such a suggestion. They found that PFPS subjects exhibited reduced patellar cartilage thickness. The relationship between bone stress and pain was examined using metabolic activity measurement because bone stress cannot be directly measured in vivo. The painful knee, in this study, showed increased tracer uptake compared to the non-painful knee and correlated with pain intensity (Draper et al., 2012). Ho, Hu, Colletti, and Powers (2014) found that PFPS patients exhibited higher patella bone oedema compared to a control group. They suggested that such a finding is a sign of venous engorgement, which may in turn may lead to intraosseous pressure and pain. This may result in OA (Utting, Davies, & Newman, 2005). However, other studies have not reported any evidence for such a relationship (Kornaat et al., 2006; Kornaat et al., 2007). Lastly, Fulkerson (2002) suggested that a change in motion may damage the cartilage, reduce the activity level and increase overloading of the PFJ. Nevertheless, it seems that it is a coherent and integrative process; the proposed mechanisms may not be separate. All of this eventually leads to the main mechanism, which is seen to be overloading. Regardless of how the development of PFPS began, which varies according to the aforementioned studies, it is therefore as a result of a change in PFJ loading distribution. This change, in turn, may damage the cartilage, and then pain results. This might be one interpretation for the most common mechanism, which is that PFPS is a multifactorial mechanism.

Figure 2.3: Area affected by PFPS highlighted in red (patella and distal femur)

2.9 Knee and hip biomechanics and their association with ACL injury and PFPS

The risk of knee injury increases when body loading increases. Abnormal hip strength or neuromuscular control is associated with increased knee valgus. During sport manoeuvres, dynamic knee valgus can be seen during the deceleration phase of double-leg and single-leg landing. It has been suggested that knee valgus moment, which is directly associates with knee-valgus angle, is a predictor of ACL injury. This was seen in a prospective study of female athletes participating in soccer, basketball and volleyball. The joint angles and moments were measured for those subjects during a jump-land task. Those who sustained ACL injury (n = 9) showed greater knee valgus angle and moment (Hewett et al., 2005). Frontal plane knee movement along with hip moment (external) has been suggested as being a risk factor for ACL re-injury for young athletes who return to sport after ACLR and rehabilitation (Paterno et al., 2010). These findings were supported recently by Myer et al. (2015a). They found that increased knee valgus external moment was a predictor for ACL and PFPS in girls. However, such results may not be generalized to older athletes who are more prone to these injuries. Moreover, kinetic data were collected during a drop vertical jump (DVJ), so, this may not apply to those who participate in sports that include different tasks, such as forward or sideways SLL.

A small degree of knee flexion can increase ACL strain, particularly if it is combined with knee valgus or internal rotation loading, and cause injury. This is consistent with actual ACL injury, which was found to occur at small knee flexion following initial contact (IC) (Cochrane et al., 2007; Krosshaug et al., 2007; Boden et al., 2000; Olsen et al., 2004). With regard to the frontal plane, many studies have suggested that ACL injury occurs when the knee is abducted at IC or during the deceleration phase (Borotikar et al., 2008; Russell, Palmieri, Zinder, & Ingersoll, 2006). Knee valgus moment has been suggested as being one of the main risk factors for ACL during the deceleration phase of cutting and jumping tasks. This was observed in different biomechanical studies (Renstrom et al., 2008; Mclean et al., 2007; Besier et al., 2001). This agrees with cadaveric studies which found increased ACL strain due to increased abduction load (Shin, Chaudhari, & Andriacchi, 2009; Fukuda et al., 2003). Landing is a task commonly seen in sport, and ACL injury can occur during landing. This task was examined in a prospective study and the results showed that females who sustained ACL injury had more than double the average knee valgus moment (Hewett et al., 2005). Interestingly, about 38 Nm of difference between ACL injured and non-injured individuals was stated by Hewett and his colleagues. Nevertheless, this was in the absolute moment (Nm); therefore, such a difference may not exist when the moment is normalised to body weight as the injured group were heavier. However, most of the aforementioned studies examined a single task only. Given that measuring valgus moment needs laboratory testing, which is not usually available in sports clubs, as well as being time-consuming for both examiner and patient, it seems useful to find simpler methods to predict those who are at risk of ACL injury. Myer, Ford, Khoury, Succop, and Hewett (2010b) suggest that some biomechanical variables can predict 78 per cent of knee valgus moment during landing. However, this sensitivity may increase when examined in a battery of tests.

The aetiology of PFPS is suggested to be associated with PFJ kinetics, such as movement speed, step length and foot-strike pattern. The PFJ is exposed to a contact force ranging between 4 and 10 times body weight (Lenhart et al., 2014; Kernozek, Vannatta, & Van den Bogert, 2015; Willson, Ratcliff, Meardon, & Willy, 2015). The knee's repeated exposed to large forces, particularly at high loading, increases the pressure on the patella and subchondral bone metabolic activity. This is believed to be associated with PFPS (Barton, Menz, Levinger, Webster, & Crossley, 2011). Changes in lower-limb mechanics, knee-abduction angular impulse and adduction excursion have been examined in runners and suggested as being associated with PFPS (Willson & Davis, 2009; Stefanyshyn, Stergiou, Lun, Meeuwisse, & Worobets, 2006). Also, the kinematics of the lower extremities are thought to contribute to PFPS. Bazzett-Jones et al. (2013) reported increased knee- and hip-flexion angle among PFPS subjects, particularly at IC. However, the investigation was performed after exhausted running. Therefore, a fatigue effect may have been present. Increased knee valgus and altered tibiofemoral rotation are also reported as risk factors of PFPS, as they increase the compression on the lateral tibiofemoral joint and thus lead to lateral patellar tracking (Salisch & Perman 2007). However, most studies have examined PFPS runners, so their results cannot be generalised to other sport players. Q-angle is suggested as being related to PFPS by Souza & Powers (2009). They conducted a study to examine femoral inclination and anteversion between a PFPS group and a healthy control. The PFPS group showed greater femoral inclination angle, which may lead to biomechanical alteration, and thus increased patellofemoral load, and consequently alter the Q-angle. This creates lateral force on the patella and leads to PFPS. However, the association between Q-angle and PFPS is not agreed (Dixit et al., 2007; Herrington & Nester, 2004). Additionally, most studies have examined selected kinematic variables. Therefore, there is a lack of evidence about the relationship between lower-limb kinematics and PFPS. Table 2.3 is a summary of key studies that have examined the association of hip and knee mechanics with ACL injury and PFPS.

Table 2.3 : Summary of studies that have examined the association of hip and knee mechanics with ACL injury and PFPS

ACL injury		
Author	**Reported risk factor**	
	Hip	**Knee**
Boden et al. (2000)	Increased flexion	Increased abduction
Ebstrup, and Bojsen-Møller (2000).	N/A	Valgus with femur internal rotation, knee extension with valgus and femur internal rotation, varus with femur external rotation
Olsen et al. (2004)	N/A	Slight knee flexion. valgus and femur external rotation
Hewett et al. (2005)	N/A	Increase abduction angle and moment
Cochrane et al. (2007)	N/A	≥ 30° flexion, valgus, and femur internal or external rotation.
Krosshaug et al. 2007	Increased flexion	Increased valgus
Hewett et al. (2009).		Increased abduction angle
PFPS		
Willson et al. (2008)	Increased abduction and flexion. Decreased internal rotation.	N/A
Boling et al. (2009)	Increased internal rotation	Decreased knee flexion and increased VGRF
Souza and Power (2009)	Increased internal rotation and decreased hip torque	N/A
Willson and Davis (2009)	Decreased abduction and external rotation. Increased adduction excursion	N/A
Myer et al. (2010a)	N/A	Flexor moment. Increased abduction moment and load
Verrelst et al. (2014)	Increased transverse-plane movement	N/A
Dos Reis et al. (2015)	Increased hip adduction and internal rotation angle. Faster time to peak internal rotation angle. Slower time to peak adduction angle. Increased adductor moment. Less power absorption in frontal plane.	Decreased knee-flexion angle. Faster time to peak knee adduction and flexion angle. Increased adductor moment. Decreased extensor moment. Less power absorption in sagittal plane.

ACL = Anterior cruciate ligament, PFPS = Patellofemoral pain syndrome, N/A = Not available, VGRF = Vertical ground reaction force

2.10 Knee valgus as a risk factor

During dynamic activities, movement dysfunction as well as biomechanical abnormalities may lead to high joint-reaction force, increased knee-joint load, increased knee valgus motion and increased knee valgus angle. Variations in these factors may accelerate joint-disease progression and increase the risk of injury. Excessive knee-valgus malalignment has been linked to knee injury. However, the point of knee valgus at which injury may occur is debatable and varying points have been suggested by several researchers. For instance, Hewett, Ford, Myer, Wanstrath, and Scheper (2006) suggest that 8° or more of knee-valgus collapse during pre-season screening increases the risk of ACL injury during the season. Herrington and Munro (2010) provide normative data for 2D FPPA a physically active population during DLL and SLL. During DLL, knee-valgus angle ranged from 3° to 8° for males and from 7° to 13° for females, while it was 1–9° for males and 5–12° for females during SLL. The differences in the findings might be attributed to the different methods used and the populations examined. Knee-valgus angle may vary from one functional task to another and from one population to another. Therefore, such findings cannot be generalised, and it is useful to investigate the values of these angles in other common functional tasks.

Knee-valgus malalignment contributes to many knee conditions, including OA, iliotibial band syndrome (ITBS), ACL injury and PFPS (Kimura et al., 2012; Powers, 2010). This section will focus on ACL and PFPS as the two most common sports-related injuries.

2.10.1 Knee valgus in relation to ACL injury

Yu, Kirkendall, and Garrett, (2002) describe ACL injury occurrence as a result of non-contact injury. Such an injury typically occurs through a combination of knee valgus, minimal flexion and external tibial rotation with the foot firmly fixed on the ground (Olsen et al., 2004). The possibility of injury may rise due to an increased dynamic knee-valgus angle.

There is a potential relationship between excessive dynamic knee valgus and ACL injury risk. Athletes who participate in sports which include jumping, cutting and landing manoeuvres may increase their risk of ACL injury by six times, particularly when these tasks are performed with increased knee valgus (Griffin et al., 2000). Hewett et al. (2005) hypothesize that such activities increase the abduction load on the ligaments, resulting in an increased risk of ACL injury. A total of 205 female athletes were examined to investigate whether female athletes with an ACL injury show decreased neuromuscular control and increased knee valgus. Using a 3D motion analysis system, researchers found that female athletes with a dynamic valgus of 8.4º or more and 27 Nm of knee valgus moment sustained an ACL injury (Hewett et al. 2005). This could be due to the high strain on the ACL which results from a combination of valgus and internal rotational moment, which was found by Shin et al. (2011) to be high enough to cause ACL rupture. These findings may explain the large number of ACL injuries observed among females as they usually land with more knee valgus than males (Kernozek et al., 2005; Chaudhari, Hearn, Leveille, Johnson, & Andriacchi, 2003). However, Hewett et al.'s (2005) study examined absolute moment which may differ when it normalized to body weight. They also did not control for other factors which could potentially influence knee-valgus angle and the incidence of ACL injuries, such as athletes' level of play (Söderman, Pietilä, Alfredson, & Werner, 2002) and training methods (Veugelers et al., 2016). Also, a lack of information about the exact time of injury and the precise ACL loading make the determination of which movement raises ACL strain using videographic analysis difficult (Utturkar et al., 2013). Mazzocca, Nissen, Geary, and Adams (2003) suggest rupture of the MCL creates increased valgus loading which increases ACL loading. This has also been suggested by Shin et al. (2009), who reported that ACL rupture may not occur without MCL damage. However, Mazzocca et al. (2003) and Shin et al. (2009) were cadaveric studies which may differ from actual living organs.

MRI for post-ACL injury patients showed bone bruises in the lateral compartment of the distal femur (Nishimori et al., 2008; Nakamae et al., 2006). The location of these bruises may suggest the presence of high impact forces on the proximal tibia and distal femur. This supports the hypothesis that knee-valgus loading and tibial internal rotation may correlate with ACL injury (Shin et al., 2011). At least 1200N (for females) and 1500–2000N (for males) of force is needed to cause damage to the ACL (Chandrashekar et al., 2006). The most strain on the ACL results from anterior tibial shear, but that force alone cannot cause ACL rupture (McLean et al., 2004a). Therefore, it seems probable that more than one excessive movement is required to apply enough force to cause ACL rupture. For example, anterior tibial shear force combined with knee-valgus and/or rotational moments increase the strain on the ACL significantly, which raises the possibility of ACL injury. This applies more at angle closer to knee extension and supports the proposed mechanism of ACL injury (McLean et al., 2004a).

2.10.2 Knee valgus in relation to Patellofemoral Pain Syndrome

It is commonly believed that knee-valgus dysfunction is influenced by the hip muscles and contributes to the development of PFPS. According to Barton et al. (2011), several functional disorders of the lower limbs, such as a larger Q-angle, are related to PFPS. However, the association between Q-angle and PFPS is debatable. For example, Rauh et al. (2007) found that runners with a Q-angle of more than 20° were more prone to knee injury than those with a normal Q-angle. In contrast, PFPS patients did not show a larger Q-angle and there was no correlation between the onset of PFPS and Q-angle (Park & Stefanyshyn, 2011). The reason for these conflicting findings may be due to other anatomical factors such as pelvic tilt and femoral anteversion, which have been found to influence the magnitude of the Q-angle (Nguyen et al., 2010), and in turn increase the knee-valgus angle. Such factors were not considered in the aforementioned studies.

Myer et al. (2010c) found that athletes who developed PFPS had increased knee-valgus moment in the affected leg, compared to those who did not go on to develop PFPS, which meant that that knee was in a valgus position. Dynamic knee valgus has been associated with the pathogenesis of PFPS in female athletes (Petersen et al., 2010), because it leads to lateralisation of the patella (Petersen et al., 2014). Using MRI, Souza, Draper, Fredericson, & Powers, (2010) evaluated kinematics of the patellofemoral joint in females with PFPS. PFPS. Compared to control group, PFPS subjects demonstrated greater lateral displacement of the patella. Compared to control group, PFPS subjects demonstrated greater lateral displacement of the patella and greater medial femoral rotation.

Biomechanically, when the knee is placed in a loaded situation, for example weight-bearing activities, the hip abductor and hip external rotator are activated to control hip adduction and internal rotation movements, which can result in knee valgus. An inability to do so increases the valgus angle during dynamic activities such as walking and running (Ireland et al., 2003), which increases the contact pressure on the PFJ and may result in PFPS (Elias, Wilson, Adamson, & Cosgarea, 2004a). Impairment of hip-muscle performance may induce hip-joint dysfunction in all planes, because the joint is dependent on a complex muscles group that create appropriate motion and provide its stability during movement (Powers, 2010). Femur-movement abnormality influences the kinematics of the tibiofemoral joint and strains the soft tissues linking the tibia to the distal end of the femur (Powers, 2010). Dynamic knee valgus, a combination of reduced knee flexion, increased hip-internal rotation and high knee-valgus loads (Hewett et al., 2005), correlate with PFPS development (Myer, Ford, Khoury, Succop, & Hewett, 2010a; Boling et al., 2009; Stefanyshyn et al., 2006). Females commonly demonstrate a posture with greater dynamic knee valgus or FPPA than males, which may explain the differences in injury rates (Munro et al., 2012a; Hewett et al., 2004; Ferber et al., 2003; Ford et al., 2003; Zeller et al., 2003). Mascal, Landel, and Powers (2003) demonstrated a correlation between knee-valgus angle and PFPS. They examined the biomechanics of the hip and knee during gait and step-down manoeuvres. Those who demonstrated excessive hip adduction, internal rotation and knee valgus were involved in 14 weeks of endurance training for the hip, pelvis and trunk muscles. At a post-intervention assessment, significant improvements in hip adduction, internal rotation, knee valgus and pain were noted during a step-down manoeuvre. Although pain was reduced, it is not clear whether it was due to an improvement in biomechanics. However, interpretation of these findings suggests a relationship between knee valgus and PFPS.

2.11 The role of hip angles and moments in ACL and PFPS

Abnormal neuromuscular control has been linked to ACL injury. During sport tasks, GRF can reach many times body weight (Kernozek e al., 2005). This force, if not absorbed properly, can result in ACL injury (James, Dufek, & Bates, 2000). Hip-joint stability control helps in distributing the load on the knee joint (Hewett et al., 2006). Hip angles, particularly during landing, may contribute to determining the impact force on the knee joint as increased landing stiffness (described by reduced flexion angle) is associated with less energy absorption and most of the body’s kinetic energy can be absorbed by eccentric contraction of the hip extensor muscles (Schmitz, & Shultz, 2010; McNitt-Gray, Hester, Mathiyakom, & Munkasy, 2001). Dysfunction of the hip may result in alteration to knee loading and increase the risk of injury (Reiman, Bolgla, Lorenza, 2009).

Hewett et al. (1996) reported a significant correlation between valgus collapse, which is a risk factor of ACL injury, and impact force, which also correlated with altered hip angles, during landing. A position of no return, which is a combination of HADD and knee valgus, is the ACL injury mechanism most proposed, particularly in females (Hewett et al., 2005; McLean et al., 2004a). Eccentric contraction of the hip-abductor muscles helps in controlling knee-valgus angle and torque via controlling femoral internal rotation which affects the HADD during weight-bearing activities (Piva et al., 2005). Placement of the joint in such a position may lead to uncontrolled femoral adduction and internal rotation, which increases the dynamic knee's Q-angle (Ireland et al., 2003) and repetitive movement with this dysfunction may cause knee injury (Thijs, Van Tiggelen, Willems, De Clercq, & Witvrouw, 2007). Moreover, a position of no return is triplanar motion, which is resisted by hip extensor, abductor and external rotator muscles (Powers, 2010).

Prospective (Nadler, Malanga, DePrince, Stitik, & Feinberg, 2000) and retrospective (Leetun, Ireland, Willson, Ballantyne, & Davis, 2004; Niemuth et al. 2005) studies have suggested that knee injury is related to issues at the proximal end of the kinetic chain. The hip joint shares the femur with the knee joint and is the most obvious proximal link. Excessive motion of the femur can affect the knee and the soft tissue around it (Powers, 2010). When the hip is adducted, the centre of the knee joint is shifted medially and the tibia abducted, which results in dynamic knee valgus. Increased knee valgus is associated with reduced hip-muscle strength (Hollman et al., 2009; Jacobs, Uhl, Mattacola, Shapiro, & Rayens, 2007; Claiborne et al., 2006; Willson et al., 2006) and associated with various knee injuries such as ACL injury (Hewett et al., 2005) and PFPS (Elias et al., 2004a).

The association between hip position and PFPS during functional activity has been examined in several studies. Some of them are summarised in Table 2.4

With regard to ACL, Houck, Duncan, & Haven, (2005) examined the differences in hip kinematics and kinetics between "non-coper" ACL deficiency subjects and healthy controls while performing sidestep cuts, crossover cuts and steps while proceeding straight. No significant differences between the groups were reported for hip angle in the frontal and transverse planes for all tasks. Sagittal-plane hip moment was higher in the non-copers ACL deficiency group. However, the differences cannot be attributed to the task as there was no interaction effect. In this study, the definitions of "non-copers" and "deficiency" are unclear, which might be a source of bias and thus influence the result. Both partial tear of the ACL and ACL rupture can be considered as deficiencies, but the performances might be different between individuals with a partial tear of the ACL and those with a complete tear. Fitzgerald, Axe, & Snyder-Mackler (2000) found that 76 per cent of ACL deficiency subjects can participate in sport without surgery. "Non-copers" were determined by one or more episode of giving way or those who rated themselves as $\geq 60\%$ on the knee-function questionnaire. Although the knee-function questionnaire was used in previous literature, the validity and reliability of the “giving way” episode in this questionnaire is questionable.

Table 2.4 :Studies that have examined the association between hip position and PFPS during functional activity

Author	Participants	Tasks	Findings
Bolgla et al. (2008)	18 PFPS, 18 controls (all female)	Stair-stepping	No between-group differences in frontal and transverse hip angles
Willson & Davis (2008)	20 PFPS, 20 controls (all female)	Running, SLS, and single-leg jump	PFPS group reported significantly greater HADD angle and lower internal rotation.
Boling et al. (2009)	991 males, 606 females (all midshipmen)	Jump landing	Increase hip-internal rotation angles were reported for those who developed PFPS.
Souza & Power, (2009)	21 PFPS, 20 controls (all females)	Running, drop jump and step down	Significantly greater hip-internal rotation angles were reported for the PFPS group.

SLS = single-leg squat, PFPS = patellofemoral pain syndrome, HADD = hip adduction.

Females with PFPS reported greater HADD angles during running, SLS, single-leg jump (Willson & Davis, 2008) and prolonged running (Dierks, Manal, Hamill, Davis, 2008). The difference in Dierks et al. (2008) was found at the end of the run, which may reflect fatigue. However, such findings were recently supported by Meira and Brumitt (2011). They conducted a systematic review to examine the relationship between hip dysfunction and PFPS. This review covered the period between 1950 and 2010 and included different study designs, such as RCT, case-control, cross-sectional and quasi-experimental. Although there were differences in the designs of the included studies, there was a link between HADD and PFPS. The explanation for this is that an increase in HADD angle leads to an increased Q-angle and relative knee valgus. This, in turn, increases the lateral contact pressure on the PFJ which may lead to PFPS.

Patients with PFPS showed increased HADD moment or decreased hip-abductor muscle strength compared to healthy individuals (Ferber, Kendall, & Farr, 2011; Cichanowski, Schmitt, Johnson, & Neimuth, 2007; Bolgla, Malone, Umberger, & Uhl, 2008). Nevertheless, these studies examined the hip muscles in a side lying position using a handheld dynamometer. According to Bolgla et al. (2008), this position may give a mechanical benefit, since it allows greater arm movement for the examiner and decreases muscle-fibre length. In contrast, DiMattia et al. (2005) found no relationship between HADD moment and hip-abductor muscle strength during SLS.

2.12 The association between ground-reaction force and knee injury

It has been reported that the GRF acts on a 3D plane axis (Horizontal X, Vertical Y and Transverse Z axes) and it is the largest force acting on the body (Winter, 2009). Many studies have linked GRF to ACL injury (Herman et al., 2009; Hewett et al., 2005). The explanation for this is that GRF influences knee-flexion-extension moments, and in turn influences the magnitude of anterior tibial shear which is considered the most direct load on the ACL. Increased tibial shear combined with abnormal frontal-plane movement causes the ACL to experience the greatest load (Pollard, Sigward, & Powers, 2007). If this load exceeds the strength of the ligament, injury may occur. During landing, peak ACL loading occurs at the time of maximum GRF. Sell et al. (2007) recently supported such a finding, during a stop-jump manoeuvre, as they reported that external knee-flexion moment and posterior GRF can predict internal tibial shear, which could be a cause of ACL rupture. It has been reported that VGRF may reach 4.4 times body weight in the activity where jump and landing is not involved (e.g. cycling and sailing) and 4.6 times of body weight during activities that include jumping and landing (e.g. volleyball and basketball) (Kernozek et al., 2005). Hewett et al. (2005) noted a relationship between ACL injury and GRF during landing. Moreover, volleyball, basketball and adolescent football players with ACL injury reported greater peak GRF compared to healthy controls (Myer et al., 2005). This suggests that landing with greater GRF may increase the possibility of ACL injury. However, such findings may not be generalisable to other sports players. Moreover, participants in Myer et al., (2005) study jumped from a box 0.3 metres in height and onto both feet. Therefore, GRF when jumping from a different height or landing on one leg might be different. In contrast, VGRF was reported to be less in magnitude during a jump-landing task for those who developed PFPS compared to those who did not (Boiling et al., 2009). Discrepancies in the results suggest that the role of GRF in knee injury is still not clear and needs to be investigated more, particularly during athletic tasks. However, a study supports the idea that the risk of ACL injury is reduced when the lower-extremity muscles can absorb GRF properly (James, Dufek, & Bates, 2000). Therefore, it seems that GRF is a risk factor for ACL injury and becomes riskier when it interacts with other factors, such as muscle strength and biomechanical alteration.

2.13 ACL injury prevention

The large volume of research on ACL and on analysis of its injury mechanism has led to the identification of many injuries' risk factors. This, in turn, has led researchers to develop different prevention programmes in the hope of finding optimal prevention programmes that can reduce the numbers of ACL injuries as researchers have documented that knee injury can be prevented if an intervention programme is sufficient (Jensen et al., 2012; Petersen, Thorborg, Nielsen, Budtz-Jørgensen, & Hölmich, 2011; Pasanen, Parkkari, Pasanen, & Kannus, 2009;). Finch (2006) suggests that the efficiency of intervention in a controlled experiment does not reflect the actual situation and environment, which means that findings may not be widely adopted and have an impact on public health. However, several controlled studies have shown promising results in that intervention can produce the desired effect, particularly with ACL injury (Waldén, Atroshi, Magnusson, Wagner, & Hägglund, 2012; LaBella et al., 2011; Gilchrist et al., 2008; Soligard et al., 2009; Myklebust et al., 2003; Mandelbaum et al., 2005; Petersen et al., 2005; Olsen, Myklebust, Engebretsen, Holme, & Bahr, 2005; Pfeiffer, Shea, Roberts, Grandstrand, & Bond, 2006). However, a single or limited mode of training was utilized in most of the studies, such as a balance exercise, a plyometric exercise or a combination of these. Moreover, such approaches lasting for a long time (up to 90 minutes) may affect the athlete's actual training schedule. The focus of these studies was mostly on females, which may reduce the generalizability of the findings. Moreover, it is still unknown whether the desired affect is carried over into real-life conditions (Myklebust, Skjølberg, & Bahr, 2013).

To examine the exact effect of such a programme, Myklebust et al. (2013) evaluated the effect of ACL injury-prevention initiatives taken in Norwegian handball. They concluded that ACL injury can be reduced through prevention initiatives, especially when including the coach as a key partner. However, this evaluation study covered prevention initiatives from 1998 to 2011, such as a coach-delivered programme and a physio-delivered programme. Therefore, it is unknown which one of these initiatives is the most effective.

Recently, Taylor, Waxman, Richter, and Shultz, (2013) conducted a meta-analysis to evaluate the effects of these intervention programmes and their duration for ACL injury. Thirteen studies were included, and the results revealed that contact and non-contact ACL injuries were significantly reduced after prevention-training programmes. The outcomes of meta-regression analysis revealed a significant association between greater duration of balance training and higher risk of ACL injury. However, greater duration of static stretching was linked to lower ACL injury risk.

The effect of feedback training on knee injury has also been examined in several studies (Munro & Herrington, 2014; Willy, Scholz, & Davis, 2012; Ford, DiCesare, Myer, & Hewett, 2015; Mizner, Kawaguchi, & Chmielewski, 2008; Herman et al., 2009). These studies confirm the concept of providing critical feedback. These studies also suggest that feedback training can improve some ACL and PFPS risk factors, such as reducing knee valgus angle and moment, increasing knee flexion angle, increasing hip flexion and abduction angles, decreasing hip-internal rotation and adduction angle and reducing VGRF.

The effect of real-time gait retraining on ACL has been examined (Crowell, & Davis, 2011; Barrios, Crossley, & Davis, 2010; Noehren, Scholz, & Davis, 2010). 3D motion analysis was used to provide real-time feedback to modify the risk factors of knee-injury risk. As the focus of recent studies is on feedback training, a summary of some of these studies is shown in Table 2.5.

Table 2.5 : Summary of feedback studies

Authors	Participants	Task	Feedback	Findings
Onate et al. (2005)	51 recreational athletes	Jump, land	Expert, self and combination	Self and combination increase knee flexion and decrease GRF
Walsh et al. (2007)	25 basketballers	DJ	Expert	Reduced knee valgus angle and force
Mizner et al. (2008)	37 athletes (female)	Drop vertical jump	Verbal instruction	Reduced GRF, knee valgus angle and knee valgus moment Increased knee flexion angle
Cronin et al. (2008)	15 volleyballers (female)	Leg-spike jump	Expert	Reduced VGRF
Herman et al. (2009)	58 athletes (female)	DLL	1- Feedback instruction + strengthening 2- Feedback only	In (1), increased hip abduction angle In (2), increased hip flexion, hip abduction, knee flexion and anterior shear force, and decreased GRF.
Dempsey et al. (2009)	12 athletes (male)	Cutting (45°)	Visual and oral feedback	Reduced peak knee valgus moment
Barrios et al. (2010)	8 healthy with varus malalignment	Treadmill walking	Video feedback	Reduced knee adduction angle and knee valgus moment Increased hip internal rotation and hip adduction
Crowell and Davis (2011)	10 runners	Treadmill running	Real-time video	Reduced GRF, force rate and tibial acceleration
Willy et al. (2012)	10 runners with PFPS (female)	SLS, running and step down	Mirror and verbal feedback	Reduced hip adduction and abduction moment Improved pain and function
Munro and Herrington (2014)	28 students	Drop jump and SLL	Self and expert	Reduced FPPA
Ford et al. (2015)	4 athletes (female)	DVJ	Kinetic and kinematic visual feedback	Kinetic visual feedback reduced knee valgus angle and moment

GRF = Ground reaction force, DJ = Drop jump, VGRF = Vertical ground reaction force, PFPS = Patellofemoral pain syndrome, SLS = Single-leg squat, DVJ = Drop-vertical jump, DLL = Double-leg landing, FPPA = Frontal plane projection angle.

Not all prevention programmes have shown a positive effect in reducing ACL injuries. For example, plyometric training focusing on lower-limb alignment was examined during landing from a jump and changing direction while running (Pfeiffer et al., 2006). The results showed no change in ACL injury incidence. Similarly, Myer, Ford, Brent, and Hewett (2007) examined the effect of an intervention programme that included plyometric balance training, core strengthening, speed and resistance training on those who are at high risk of ACL injury (described as those who have knee valgus moment $\geq$ 25.25 Nm). They found that knee valgus moment reduced but not to a level that may prevent injury. However, these studies only included female subjects, which may prevent generalisability to males.

2.14 Functional performance

Over the past decade, researchers have encouraged practitioners to evaluate and treat patients within the context of their function. Therefore, those participating in sport activity or training should be evaluated with consideration of sport-related physical demands (Kivlan & Martin, 2012).

Functional performance can be influenced by injury and a safe return to sport will involve appropriate muscle strength, power, flexibility, endurance, speed and agility (Manske & Reiman, 2013).

Functional performance has been described as the "result of neuromuscular training" (Engelen-van Melick, van Cingel, Tijssen, & Nijhuis-van der Sanden, 2013) and it can provide practitioners with information about the quantity and quality of movement involved in sport and exercise (Reiman & Manske, 2009).

Muscle strength and hop-test distance are examples for quantity of movement while knee-valgus angle and knee-flexion degree during dynamic tasks are examples of quality of movement (Ekegren et al., 2009; Von Porat, Holmström, & Roos, 2008). Optimizing these components is vital for the prevention and rehabilitation of ACL injury and re-injury (Thomeé et al., 2011; Paterno et al., 2010; Renstrom et al., 2008). However, other important physical components such as movement skills and muscle flexibility should be considered when describing functional performance (Reiman & Manske, 2009), as well as joint receptors which play an important role in joint stability (Williams et al., 2001).

Functional performance is crucial for athletes (Prieske et al., 2016). Functional performance, such as sprinting and jumping, can be improved by strength and plyometric training (Ronnestad, Kvamme, Sunde, & Raastad, 2008). There is evidence to support strength training being able to improve strength and functional performance (Wong, Chamari, & Wisloff, 2010). Trunk control is considered pivotal for biomechanical function as it maximizes force generation and reduces joint load in any functional activities (Kibler, Press, & Sciascia, 2006). Therefore, trunk strength has also been described as enhancing functional performance (Kibler, Press, & Sciascia, 2006). This was later supported by several studies that reported a significant relationship between trunk muscles strength and agility, short-distance sprint and jump performance (Sharma, Geovinson, & Singh Sandhu, 2012; Nasser, Huxel, Tincher, & Okada, 2008). Furthermore, a significant improvement in hip-muscle strength and jump performance was reported in adolescent soccer players following stability and strength training for the core muscles (Hoshikawa et al., 2013). Regarding such findings, it seems that core strength might be critical to improve performance. However, Prieske et al. (2015) found only a limited effect of trunk-muscle strength on jump performance. Hence, more investigation on different types of sports and tasks is needed. It is noteworthy that performance in many sports sometimes occurs on unstable surfaces such as landing on uneven turf, and landing or kicking a ball with impedance from an opponent. Accordingly, Behm, Drinkwater, Willardson, and Cowley (2010) suggested that training must imitate the demands of sports. Compared to stable surface conditions, trunk-muscle activity during strength training increases under unstable surface conditions. Therefore, including unstable elements during training could result in better athletes' performance. Only two studies have investigated the changes in performance following core strengthening on stable and unstable surfaces in healthy untrained children (Granacher et al., 2014) and elite soccer players (Prieske et al., 2016). Significant improvements in trunk muscle strength, sprint, kicking and jumping sideways, a Y balance test and a stand-and-reach test were recorded. However, untrained subjects were included in Granacher et al.'s (2014) study, making the application of results to trained subjects disputable, as the adaptive reserve is higher for perception and maximum strength gains are lower in trained subjects (Yarrow, Brown, & Krakauer, 2009).

Furthermore, neuromuscular training, as described by Hewett, Lindenfeld, Riccobene, & Noyes, (1999), is a combination of plyometric agility, weight, balance and sport-specific exercise. The above explanation might be ambiguous to some extent as it does not make a clear distinction between physical and functional performance. As a clinician, it is important to differentiate between them to achieve the best results with clients. For example, a client with good quadriceps muscle strength, which is physical performance, could still have difficulty in performing SLS – for instance – which is a functional performance test, because of poor balance. In other words, functional performance is the ability to perform a task in a form considered right for the person (Reiman & Manske, 2011). Yet, the word (right) and the difference between successful and non-successful functional performance is still unclear. To describe a performance as a 'right' or successful performance, many factors should be considered, such as gender, age and body type, as the right performance for young people might not be like elderly people's performance. Likewise, the right performance for obese people may differ from people who are slim, and so on. This critical point is still vaguely represented in the literature and needs to be further investigated. Therefore, a reference value, which depends on a specific task in a specific sport, is required to determine or describe a 'right' or successful performance (Reiman & Manske, 2009).

2.15 Assessment of functional performance

Functional performance can be assessed in many ways. Currently, the most commonly used methods in the literature are by measuring impairment, self-reported measures and physical performance measures (Hildebrandt et al., 2015; Logerstedt et al., 2014; Reiman & Manske, 2011; Lentz et al., 2012).

Impairment measures can include ROM, muscle strength, joint mobility and joint laxity. Impairment in these components may limit function, which is reflected in functional performance (Jette, 1994). Despite their validity of use, impairment measures may not truly represent the level of functional impairment. For example, limitation in knee ROM, which is an impairment measure, may not mean difficulty in picking up a key from the ground, which is a functional task, as the person may compensate during movement by leaning forward (moving from the trunk instead of the knee).

Self-reported measures are a widely-used method, particularly for pain assessment and the progress of improvement in patients with different diseases or surgeries, such as PFPS (Long-Rossi & Salsich 2010), ACL (Logerstedt et al., 2012) and low back pain (Reneman, Jorritsma, Schellekens,

& Göeken, 2002; Simmonds, Protas, & Jones, 2002). Clearly, this type of measurement is important but it does not usually represent a perfect reflection of functional performance as the findings of these methods are conflicting. Stratford and Kennedy (2006), using self-reported methods with total-knee-arthroplasty patients, found decreased pain and improved functional ability, while the time to perform a functional task increased. In contrast, there was a moderate association between self-reported pain and functional performance in Reneman et al.'s (2002) and Simmonds et al.'s (2002) studies. Therefore, self-reported measures should be used with caution and are better used with other functional assessment methods (Reiman & Manske, 2011).

Physical performance measures are the most common type of functional assessment used to measure different characteristics of functional performance, particularly in post-injury examinations (Hildebrandt et al., 2015; Ross, Langford, & Whelan, 2002; Xergia, Pappas, Zampeli, Georgiou, & Georgoulis, 2013; Sinsurin, Vachalathiti, Jalayondeja, & Limroongreungrat, 2013). Different tests have been used to determine function, such as SLS (Hollman, Galardi, Lin, Voth, & Whitmarsh, 2014) and SLL (Hong, Yoon, Kim, & Shin, 2014). All the aforementioned studies, and others, have used and described these tests as functional tests, while it seems they only measure physical performance. Also, each of them used just one test to measure only one parameter of function (successful return to function) (Reiman & Manske, 2011). As aforementioned, the word 'successful' in our context is still vague. However, due to the good reliability of the SLL test (ICC 0.75–0.97) (Munro et al., 2012a; Alenezi et al., 2014; Myer et al., 2015b), it may be considered a gold-standard functional performance test. Noteworthy is that including one or more different types of SLL test may increase the sensitivity and ability to assess different landing quality, which enhances the ability to understand inconsistencies in performance. This was shown in Reid, Birmingham, Stratford, Alcock, and Giffin's (2007) study. Therefore, clinicians should consider using a battery of landing tests to achieve a better understanding of performance.

2.16 Functional performance test (FPT)

FPT can be defined as the use of a battery of physical skills tasks and tests to assess people's ability to move around, perform daily activities and/or readiness to participate in specific activities or sports (Reiman & Manske, 2011; Reiman & Manske, 2009). At present, it is common to use FPTs in both clinical and sport practice to make decisions about returning to sport (Hildebrandt et al., 2015). FPT has an advantage over the traditional tests, such as special orthopaedic tests, because it evaluates a bodily region or system (Kivlan & Martin, 2012). FPTs are of a closed kinetic-chain nature. Therefore, any movement of any segment in the chain while the distal end of the segment is fixed will influence other segments (Leetun et al., 2004). For instance, a foot fixed to the ground during a cutting manoeuvre may influence the knee joint. However, the exact influence is still not clear. Consequently, closed-chain movement would lead to movement of the hip, knee and ankle joints at the same time. This requires good muscle coordination to control the segments (Clark, 2001a). Measurements should be available in both clinical and field-based sittings in order to facilitate treatment goals, maximize function and evaluate functional performance and the ability to participate in activities at different levels, which is one of the rehabilitation goals (Fitzgerald et al., 2000). Due to the shortage of laboratory-based techniques, such as 3D motion analysis and force-platform measurements, the use of FPTs has increased because they match the reality of sports tasks and do not require a large space or carry a high cost, unlike laboratory-based measurements.

Recently, Smith, DePhillipo, Kimura, Kocher, and Hetzler (2017) examined the ability of a battery of FPTs (triple hop for distance, double-leg lowering manoeuvre, star-excursion balance test, multi-stage fitness test and drop jump) to be used as a preseason screening tool to identify those at risk of lower-extremity injury. One hundred adolescent basketball, volleyball and soccer players were monitored during the sport season (in a six-month surveillance period). They found that those who sustained injury reported lower mean scores on FPTs, suggesting a relationship between FPTs and potential risk. Therefore, the authors suggested that a comprehensive evaluation of FPTs is beneficial to identify those who are prone to injury prior to participation. However, the participants exposure’ during the period examined was not reported. This is a crucial point as it could affect injury incidence. Moreover, the participants were recruited from a single school, which may not represent any other geographic area.

Other studies have utilised a variety of FPTs to assess the risk factors for ACL injury and screening for lower-extremity alignment. Examples of these FPTs are in Table 1.1

Some of these tests are bilateral, which may prevent comparisons of performance between the sound and affected legs. This comparison might be possible with tests that require only one leg to be completed as the sound leg could be used as a control while quantifying the function of the affected leg. Differences in function between the injured and non-injured legs were found in many studies that use tests that require only one leg to be examined (Goerger et al., 2014). However, this is not usually the case as it depends on what we need to compare. For example, a comparison between right and left knee alignment can be achieved using a bilateral test, such as drop-vertical jump.

Cognizance of how risk factors interact with sport tasks' restraints may provide a clearer vision of possible high-risk movements. Further, to use functional tests as a screening tool for those who are susceptible to injury, a better understanding of them is needed. From the literature, it could be concluded that landing, regardless of the type of landing, is the most commonly used test, particularly SLL. As aforementioned, clinicians should consider using a battery of tests to achieve a better understanding of performance. However, to date, no investigation has examined the relationship between kinetic and kinematic variables while performing a battery of SLL tests, which include forward SLL (FSLL), forward SLL off platform (FSLLP), lateral SLL (LSLL), lateral SLL off platform (LSLLP), medial SLL (MSLL) and medial SLL off platform (MSLLP). These are common manoeuvres which can be seen in many sports, such as tennis, squash and volleyball, and these are commonly associated with ACL injury. Such data may provide a better understanding of biomechanical factors that are associated with ACL injury, which, in turn, facilitate the screening of people at risk of ACL injury and their rehabilitation.

2.17 Motion analysis

There are several techniques that can be used to evaluate human biomechanics, such as inertial-motion sensors and marker-less capture. Both require less time preparation, are low cost, consume less power, are transportable, do not need markers and do not need stationary units to collect data, which makes them usable outside the biomechanics laboratory (Castelli, Paolini, Cereatti, & Croce, 2015; Fong & Chan, 2010; Coley, Najafi, Paraschiv-Ionescu, Aminian, 2005). Most of the studies that use these two techniques have examined gait and/or upper-extremity biomechanics. However, Fong and Chan (2010) conducted a systematic review to evaluate the use of inertial-motion sensors in evaluating lower-extremity biomechanics, and they concluded that data-processing and fixation procedures within this methodology are a potential limitation and need to be improved. With regard to marker-less capture, although it has been validated for measuring sagittal-plane kinematics on healthy subjects during gait (Castelli et al., 2015), it still needs validation and standardisation during other planes of movement and other functional tasks.

Video-based motion analysis systems are widely used, particularly during dynamic movements, such as landing, running, squatting, jumping and landing (Munro et al., 2012a; Herrington, Munro, & Comfort, 2015; Ugalde, Brockman, Bailowitz, & Pollard, 2105; Willson et al., 2006; Thijs et al., 2007; Heinert et al., 2008). Advances in technology and the increase in demand for evidence-based practice have led to more accurate measurement tools. High-speed motion analysis technologies provide accurate 3D lower-extremity measurements while performing different sports tasks (Gao, Cordova, & Zheng, 2012; Gao, Cordova, & Zheng, 2012; Zeller et al., 2003; Gao, Cordova, & Zheng, 2012; Ford et al., 2003; McLean et al., 2004a), which can significantly contribute to screening and rehabilitation of related injuries. Although such measurement is the gold standard in movement analysis, as it can accurately describe both multiplane joint angles and moments during functional tasks, the extension to a clinical setting (Willson & Davis, 2008; McLean et al., 2005) or to a larger sample size (Hewett et al., 2005) is limited due to the high financial cost (Nielsen & Daugaard, 2008). Moreover, there are some limitations that should be considered, e.g. the need to apply markers to subjects' skin, which has been suggested influences kinematic measurements. This is because the manual application of markers to bony landmarks, which may lead to a lack of consistency between clinicians for the same subject or between subjects by the same clinician (Queen et al., 2006). The movement of soft tissue underneath markers may also influence the movement of markers. The effect of soft-tissue movement was examined by Benoit et al. (2006). Although skin markers are still reliable, they found that pin-in-bone markers were superior. Therefore, 3D kinematics measurement is prone to errors resulting from soft-tissue movement and this should be considered when interpreting kinematic data.

The nature of clinical measurement requires simple, economic and portable methods. Therefore, such a method was proposed to quantify motion analysis. Accordingly, 2D motion analysis became popular in clinical practice. It only requires a digital video camera and digitizing software. Stensrud, Myklebust, Kristianslund, Bahr, & Krosshaug (2011) reported that 2D motion analysis is universally available, reasonably cheap and typically portable. 2D motion analysis has been used to evaluate lower extremity kinematics in healthy and injured populations (Herrington & Munro, 2010; Willson & Davis, 2008; Stenstrud et al., 2011; Herrington, 2011; Noyes et al., 2005). It is noteworthy that there are some possible errors which should be considered when using 2D motion analysis, e.g. parallax error, which usually occurs when the subject is viewed away from the optical axis of the camera and/or when the observer's line of vision towards the subject changes. To minimize this type of error, the line of sight should align with the centre of motion (Kirtley, 2006). Out-of-plane errors may also occur when the subject moves out of the calibrated plane, which makes measurement to an assumed size incorrect (Payton, 2008).

The digitizing process could be a limitation of 2D analysis as it requires visually identifying the anatomical site of interest, which may result in systematic or random errors. However, such a limitation can be kept to an acceptable level if the calibration and digitizing are done by the same examiner and by using markers on anatomical landmarks. Also, the examiner should have a good knowledge of the underlying musculo-skeletal system to be able to determine anatomical landmarks (Payton, 2008). Another obvious limitation of 2D motion analysis is its inability to capture complex and multiplanar motion, such as knee valgus (Maykut, Taylor-Hass, Paterno, DiCesare, & Ford, 2015). This concern causes researchers to examine the reliability and validity of 2D motion analysis. Although some studies reported promising results, the validity of 2D when compared to 3D is still unclear and needs to be investigated.

Frontal plane projection angle (FPPA), which can be defined as the relative angle of the femur to the tibia, is most commonly used method to evaluate frontal-plane lower-limb kinematics. Different ways can be used to determine the FPPA. Automatic tracking is one of them, which represents an important advance in the practical use of motion analysis. However, little is known about the algorithms of most of the available automatic tracking software, which is essential to optimize the tracking process in different conditions and environments (Magalhaes et al., 2013). Moreover, automatic tracking software is not usually available in clinics and sports clubs. As one of 2D motion analysis' aims is to assist workers in these fields in providing accurate motion analysis for their clients, it is important to examine what they commonly use, which is 2D manual tracking.

Two ways to determine the FPPA using manual tracking are described in the literature. The first one is using the line of the thigh, while the other is using a marker on the anterior superior iliac spine (ASIS). However, the latter one might have an advantage because ASIS is a bony landmark with less soft tissue underneath, which may reduce skin-artefact movement. Although 2D FPPA is the method most used, other 2D methods have been used, such as knee separation distance

(Sigward, Havens, & Powers, 2011; Barber-Westin, Galloway, Noyes, Corbett, & Walsh, 2005; Noyes et al., 2005). However, this method needs both legs for it to be used. Therefore, it is not applicable to tasks that are performed on one leg, such as SLL, taking into account that most knee injuries occur during SLL, a limitation that is crucial when attempting to predict knee injury.

With regard to 2D reliability, some studies have examined the reliability of lower limb biomechanics using 2D FPPA during functional tests. Munro et al. (2012a) examined the 2D FPPA of 20 recreationally active subjects during SLL, SLS and drop jump. Good within-day ICC reliability (≥ 0.59–0.88) and good to excellent between-day ICC reliability (≥ 0.72–0.91) were observed. The authors concluded that 2D analysis is a reliable measurement tool for lower-extremity dynamic KV. Positively, a good standard error of measurement (SEM) (2.72–3.01°) and small detectable difference (7.54–8.93°) were reported in this study. However, this study only examined a healthy population. Therefore, the results cannot be applied to athletes or injured populations. This study did not examine HADD angle, which is a crucial component of most proposed injury mechanisms for ACL (McLean et al., 2004a; Hewett et al., 2005). Therefore, further research is needed to examine FPPA and HADD angles during other tasks such as multidirectional SLL. 2D FPPA has also been used to predict or screen for knee injuries (Munro et al., 2012a; Norris & Olson, 2011; McLean et al., 2005). During step down, 2D video analysis has shown excellent intra-rater reliability for knee valgus and HADD (Hollman et al., 2009). During a performance test, moderate to high reliability for knee valgus (FPPA) was reported (Miller & Callister, 2009). 2D sagittal plane measurement has also shown excellent inter-rater and intra-rater reliability during mechanical lifting (Norris & Olson, 2011).

The findings regarding the validity of 2D motion analysis are conflicting. McLean et al. (2005), found moderate to strong correlation between 2D and 3D measurements when measuring knee-valgus angle during side-step (r^2 = 0.64) and side-jump tasks (r^2 = 0.58). However, lower correlation was found in shuttle runs (r^2 = 0.04). The authors then concluded that 2D motion analysis may offer similar potential to a 3D system when screening for ACL injury risk. Using 2D methods in calculating knee and hip sagittal-plane kinematics was reported to be valid ($r \geq 0.95$) during mechanical lifting (Norris & Olson, 2011). In this study, there was strong positive correlation between 2D and 3D. In contrast, poor correlation was reported between 2D frontal-plane knee kinematics and 3D knee kinematics during single-leg step, ranging between r = -0.23

and 0.34 (Olson et al., 2011). During SLS and single-leg step down, a small link between FPPA and the change in 3D joint kinematics was reported (Olson et al., 2011; Willson & Davis, 2008).

Some authors have suggested that 2D frontal-plane motion of the knee can predict dynamic knee valgus (Mauntel et al., 2014; Sigward et al., 2011; Willson & Davis, 2008; Sigward, Ota, & Powers, 2008;). Moreover, no difference in knee-angle measurements during the gait cycle was observed between 2D and 3D motion analysis systems (Nielsen & Daugaard, 2008). However, this was only for sagittal plane of movement, which may not be generalized to frontal plane of movement or to other tasks rather than gait.

Some other studies have also been conducted to examine the relationship between 2D and 3D motion-analysis methods. Mizner, Chmielewski, Toepke, & Tofte, (2012) suggest that FPPA (r^2 = 0.15) and knee-to-ankle separation ratio (r^2 = 0.35) are a good alternative for 3D dynamic-knee valgus. Willson and Davis (2008) examined the biomechanics of SLS using 2D and 3D motion analysis and found that 2D FPPA reported moderate correlation with 3D pelvic drop, posterior pelvic rotation, femoral adduction, femoral internal rotation, and tibial abduction, the authors then concluded that FPPA can predict 3D knee valgus.

Recently, Glass, Priest, & Hayward, (2008) developed a new technique to calculate 2D FPPA. The difference between this technique and the original one is that the ankle joint works as the fulcrum of the angle instead of the knee, which eliminates the need for an ASIS marker. However, this technique is not commonly used and needs more investigation, particularly with more challenging tasks such as SLL. Moreover, Belyea, Lewis, Gabor, Jackson, & King, (2015) criticized the need for a tripod and computer with traditional 2D methods and examined new methods. Accordingly, they examined the validity of using a hand-held tablet and a motion-analysis application that is available to download from an online store (KinesioCapture app). Moderate to strong positive correlation between FPPA and 3D knee abduction (r = 0.48), and between 2D knee flexion and 3D knee flexion (r = 0.77), was reported. However, holding a tablet in the hands may affect its position, which may lead to error. Furthermore, greater accuracy might be achieved by using a stylus to determine joint angles. This led the author to conclude that 2D measures using the KinesioCapture app might be a suitable alternative for actual 3D joint angle, but care would be needed in its use as it does not represent 3D motion and has limited research on reliability.

Sorenson, Kernozek, Willson, Ragan, & Hove, (2015) examined 31 healthy female subjects to determine the correlation between 2D and 3D kinematic for the knee and hip joints. The findings suggested that 2D knee FPPA and 3D knee frontal-plane kinematics correlated strongly ($r^2 = 0.72$), while 2D and 3D hip kinematics correlated moderately ($r^2 = 0.52$). However, 2D knee FPPA correlated poorly with 3D knee-adduction excursion ($r^2 = 0.06$). These findings were found at IC. Considering that knee-valgus angle increases with knee flexion, such a relationship at maximum knee flexion should be established.

However, the existence of a constant correlation between 3D knee valgus and 2D knee valgus is still questionable. Krosshaug and Bahr (2005) compared 3D and 2D tibiofemoral angles during a side-cut manoeuvre and found no correlation between the two. Accordingly, the reliability and validity of using 2D to measure knee angle during sport manoeuvres need more investigation. Moreover, the relationship between 2D and 3D measurement has not been established in multidirectional SLL. Multidirectional SLL is commonly seen in different sports and is usually used as a screening tool before a return to play. Therefore, establishing the relationship between 2D and 3D variables during this task is essential. This may fill the gap between laboratory measures and players' field testing. If studies are successful in finding a good correlation between 2D and 3D biomechanics, the use of the latter one, which costs significantly more, may be unwarranted.

2.18 Importance of landing examinations

The majority of knee injuries appear to occur during landing on one leg (Quatman et al., 2010; Borotikar et al., 2008; Borotikar et al., 2008; Tillman, Hass, Brunt, & Bennett, 2004; Olsen et al., 2004). The forces and motion of the lower extremities and trunk are greater during unilateral tasks (Stensrud et al., 2011). Although knee injury can occur during both bilateral and unilateral landing, the latter might be more menacing because of the increased demand that results from the absorption of impact on the musculature of a single leg, and the decreased support base (Pappas et al., 2007). Landing on one leg occurs frequently in many sports, such as soccer, basketball, volleyball and tennis. It regularly occurs from different heights and horizontal distances and can cause non-contact knee injury (Yu, Kirkendall, & Garrett, 2002). The landing phase is more important to assess than the take-off phase because it puts high stresses on the limbs, particularly the ACL (Yu et al., 2006; Chappell, Yu, Kirkendall, & Garrett, 2002). Kirkendall and Garrett (2000) reported that most knee injuries happen during landing. This was later supported by Paul et al. (2003). They examined 263 ACL injured subjects and found that 50 per cent of them sustained their ACL injuries during landing events. Despite the large sample size, this was a prospective study. Boden et al. (2009) recently analysed videos for 29 subjects while landing. At IC, they found that more than 55 per cent of subjects landed on one leg. More than 72 per cent of subjects had an ACL injury during SLL. Such findings are in line with the review of the Olympic Committee Current Concepts, which concluded that most of the forces on a single leg with a foot placed in front of the body are a component of knee injury (Renstrom et al., 2008). By analysing videos of ACL injuries and interviewing those who had sustained them, Olsen et al. (2004) found that unilateral landing was the most common injury mechanism, while no injury occurred during DLL. However, the focus of the literature currently is on bilateral landing as a test for injury risk. SLL is a common sport task and previous work has reported alterations to lower extremity biomechanics during this task, such as greater GRF, increased knee-valgus angle, greater knee-extensor moment and reduced hip-extensor moment (Shimokochi et al., 2013; Yeow et al., 2010), which may contribute to increased risk of knee injury. Therefore, an examination of landing on a single leg may provide valuable information that can help in improving activity in daily life, as well as sport performance, because many sport tasks are performed unilaterally (Stålbom, Holm, Cronin, & Keogh, 2007). It may also provide extra insights into the injury mechanism, which may in turn contribute to the development of injury-prevention programmes.

The highest ACL injury incidence rates are reported in multidirectional sports (Boden et al., 2009; Hootman et al., 2007). Although a double-leg task can provide meaningful data, the findings of previous studies support the importance of examining multidirectional SLL which can help in identifying the risk of ACL injury in sports with multidirectional SLL demands (Taylor et al., 2016). Moreover, while SLL is in its own right a common injury mechanism, it also has considerable biomechanical correlation to both step-landing and cutting tasks (two other common mechanisms of injury for knees), with similar hip- and knee-joint angles reported during side and crossover hop-landing (Jones et al., 2014; Ortiz et al., 2011). As previously mentioned, injuries related to SLL could occur in multiple planes, though research relating to multi-plane landing is limited. Prior to undertaking research on more complex tasks or those involving sport-specific activities, it is important to understand the fundamental biomechanics of multi-directional landing;

once this is established, the impact of sport-specific demands or other more complex activities can be studied and understood.

2.19 Gaps in the literature

Most lower-limb injuries are non-contact in nature (Renstrom et al., 2008) and may significantly impact on an athlete's career or a person's function. For example, only 34 per cent of ACL rupture athletes return to full competition and 33 per cent to competition partially (Ardern et al., 2011). Of those who return to sport, 3–15 per cent may get injured again or suffer contralateral ACL injury (Swärd, Kostogiannis, & Roos, 2010). This means that a better understanding of the risk factors, rehabilitation and preparation is needed to allow a safe return to sport (Simoneau & Wilk, 2012). Most studies have been unable to determine the criteria for a return to sport after ACLR (Barber-Westin & Noyes, 2011). To determine such criteria, clinicians should use tests that are practical, reliable, valid, have no or little risk to athletes and have reference values to allow comparison (Myers, Jenkins, Killian, & Rundquist, 2014).

Several studies have been conducted to assess lower-limb biomechanics during different sport tasks and several landing tests have been described in those studies (Table 1.1). However, most of the studies that have examined landing tests were limited to one or two types of landing tests. Due to the good reliability of the SLL test, it may be a gold-standard functional performance test. However, the sensitivity of noticing functional limitations with this test is quite low (38–52%). Combining a battery of SLL tests may raise the sensitivity to 80 per cent (Reid et al., 2007) and raise the ability to understand inconsistencies in landing performance. Therefore, a study examining a combination of an SLL test with other tests is needed.

Moreover, which landing tasks are the most appropriate to evaluate functional performance is still unanswered. As researchers mention, an injured leg might be compensated for by the uninjured one, any task performed bilaterally may hide the functional deficit that occurs after a unilateral lower-extremity injury (Pappas & Carpes, 2012).

Consequently, it seems that examining a battery of tasks performed with one leg might be better and reflect actual intra-limb performance. To the best of the researcher's knowledge, no study has examined the reliability of the biomechanical characteristics of a battery of SLL tests. Therefore, the first aim of this project is to evaluate lower-limb biomechanics during different types of SLL, namely, FSLL, FSLLP, LSLL, LSLLP, MSLL and MSLLP. These tests have been chosen because they are more challenging (puts a larger load on one leg) than double-leg landing and commonly seen in the field of sport. Also, they are commonly used as both rehabilitation exercises and screening tools. Furthermore, unilateral functional limitation may not be evident during bilateral tests (Myer et al., 2011).

Both 2D and 3D motion analysis systems are widely used in research and clinical fields. Each of them has advantages and disadvantages. The gold standard for motion analysis is 3D motion analysis, as it provides accurate and reliable 3D lower extremity measurements while sportspeople are performing different sports tasks (Gao et al., 2012; Sled, Khoja, Deluzio, Olney, & Culham, 2010; Zeller et al., 2003; Ford et al., 2003; McLean et al., 2004a). However, the extension to a clinical setting or to a larger sample size is limited due to the high financial cost and time-consuming nature (Willson & Davis, 2008; Nielsen & Daugaard, 2008). Therefore, 2D might be a good alternative, particularly when examining large populations and/or being used in a clinical environment. Some studies have examined the validity of 2D motion analysis during functional tasks (Mizner et al., 2012; Olson et al., 2011; Glass et al., 2008; McLean et al., 2005; Willson & Davis, 2008; Norris & Olson, 2011). Most of the aforementioned studies examined bilateral tasks and mostly concentrated on the sagittal plane. DLL is less challenging and may mask some important events that occur during SLL which can match the real situation of landing in sports. To the best of the researcher's knowledge, no study has examined the validity of 2D motion analysis in a battery of SLL tasks. Therefore, the second aim of this project is to examine the validity of 2D motion analysis of lower-extremity frontal-plane kinematic variables (FPPA and HADD angles) during multidirectional SLL.

Also currently unknown is what current clinical practice is around the use of these types of SLL as a rehabilitation exercise and screening tool, and what the relationship is between these tasks. Understanding of this is needed as it helps to define significant SLL tasks, which could then be biomechanically analysed for their loading characteristics. Several studies have examined the correlation between biomechanics characteristics during functional tasks. However, most of them have examined limited numbers of female subjects and the correlation between double-leg and single-leg tasks. Furthermore, most of them did not include the calculation of a coefficient of determination (r^2). To the best of the researcher's knowledge, no study has examined the correlation between different types of SLL tasks. Therefore, the third aim of this project is to explore the relationship between the aforementioned tasks and attempt to establish what the current use and practice around them are.

2.20 Project aims

2.20.1 General aim

The overall aim of this project was to examine the lower extremity biomechanics during a battery of SLL tasks in a healthy population to enable a better understanding of potential injury and performance mechanisms

2.20.2 Specific aims

1. To systematically review the available literature investigating the biomechanics of the lower-extremity frontal plane of motion during multidirectional SLL.

2. To examine the reliability of using a 2D motion-analysis system to measure lower-extremity kinematics during multidirectional SLL.

3. To examine the reliability of using a 3D motion-analysis system to measure lower-extremity kinematics during multidirectional SLL.

4. To examine the validity of 2D motion analysis in measuring lower-extremity frontal-plane kinematics during multidirectional SLL in comparison to findings from a 3D motion analysis system.

5. To examine the relationships between biomechanical characteristics during multidirectional SLL tasks using both 2D and 3D motion analysis.

3. Study one: The biomechanics of lower-extremity frontal-plane movement during different directions of single-leg landing: A systematic review

3.1 Background

The SLL test is a functional performance test that is commonly used in both research and clinical practice to evaluate the dynamic stability of the lower extremities, particularly the knee joint (Dos Reis et al., 2015). It is also an important screening tool to identify those who are at risk of lower-extremity injury. SLL is also used to evaluate the progress of rehabilitation regimes for individuals with ACL injury or PFPS (Fukuda et al., 2012; Grindem et al., 2011; MagalhaEs et al., 2010). Most knee injuries occur via a non-contact mechanism in which landing and pivoting are often involved (Agel et al., 2005; Olsen et al., 2004). Lower-limb injury, particularly to the knee joint, needs intensive and appropriate rehabilitation. There is a concern that the injured individual may have limited likelihood of returning to their pre-injury level of participation (Lentz et al., 2012; Shah, Andrews, Fleisig, McMichael, & Lemak, 2010; Swirtun, Eriksson, & Renström, 2006; Lentz et al., 2009; Thorstensson et al., 2009).

Return-to-sport decisions are partly the responsibility of physiotherapists, as they can evaluate the patient's tolerance to sport demands to prevent re-injury (Myklebust & Bahr, 2005). Such a decision needs highly accessible and reliable tools that can assess the demands of the sport that players are practising (Reiman & Manske, 2009).

The use of functional tests became popular to assess athletes' ability and readiness to return to participation, as these have lower limitations than other tests (Reiman & Manske, 2011; Narducci et al., 2011). SLL is one of the tests most used (Bjorklund et al., 2009) and is suggested as being a good indicator of an athlete's readiness to return to sport. It shows good reliability and validity in measuring different components of movement, such as strength, stability, joint mobility, neuromuscular control, balance and agility (Ardern et al., 2011; Reiman & Manske, 2009.). SLL is described as multi-segmental movement that requires coordination (Orishimo et al., 2010) and can place high demands on the lower limbs to absorb GRF (Paterno et al., 2007; Decker et al., 2003). A functional test is usually evaluated by quantity (e.g. distance) and quality (e.g. kinematics and kinetics) information about specific movement. Quality of movement can be determined during landing (Ekegren et al., 2009). Both components are crucial in rehabilitation and avoiding injury and re-injury (Thomeé & Werner, 2011; Renstrom et al., 2008; Paterno et al., 2010).

3.1.1 Rationale

Most of the previous literature is limited to reporting the quantity of movement (e.g. distance, height or time) while the quality of movement is also important but has received less attention. The majority of the studies involved the contralateral leg for comparison, though including a healthy control group might be preferable (Engelen-van Melick et al. 2015). Only a few studies have examined quality of movement. Often, they only examine the sagittal plane, while the frontal plane of movement is also important because of its association with injury (Souza & Powers, 2009; Hewett et al., 2005). Also, the suggested position of injury (position of no return) includes movement that mostly occurs within frontal-plane movement, such as HADD and knee valgus. Furthermore, the majority of relevant literature has examined a bilateral landing task that does not adequately reflect sport-specific movement (Edwards, Steele, & McGhee, 2010; Myer, Ford, & Hewett, 2008b). Bilateral tests may also not prove unilateral functional limitations and may miss important unilateral events that are commonly seen during sport (Myer et al., 2011).

Knee injuries mostly occur when the body's weight is shifted onto a single leg (Olsen et al., 2004). SLL is also a more challenging task (encountering more load than a bilateral task) and matching the sport reality (Myer, Ford, & Hewett, 2004; Olsen et al., 2004). In the literature, different tasks, participant groups, dependent variables and methodologies have been used, which makes a systematic review of this area important. This may help in drawing together evidence to support evidence-based practice (Gopalakrishnan & Ganeshkumar, 2013). It also keeps the knowledge of clinicians updated and helps them to judge the advantages and disadvantages of any intervention (Liberati et al., 2009). Moreover, it can help to guide the direction of future research and be used as evidence to compare or corroborate recent findings.

3.1.2 Objective

Considering the aforementioned limitations in the literature, an aim of this study is to review literature which investigates the frontal-plane biomechanics of the lower extremities during multidirectional single-leg-landing. This will help in establishing what types of SLL tasks have been used, note their findings (i.e. values and reliability) and evaluate the quality of available studies, which in turn can then help to summarise the results of related studies and draw a conclusion about gaps in the literature.

3.2 Methods

3.2.1 Search Strategy

A comprehensive electronic search of PubMed, MEDLINE via EBSCO, CINAHL via EBSCO, SPORTDiscus via EBSCO, EMBASE, AMED (Allied and Complementary Medicine Index), PEDro (physiotherapy Evidence database), Google Scholar and Healthsource database was conducted to collect as many related articles as possible. The search was also expanded to include a manual search of reference lists of all relevant studies to identify any further related studies not found in the original search. The search terms were customised to suit all databases and used a combination of the following terms: "single leg landing", "landing", "hop test", "single leg hop", "functional test", "performance test", "return to sport", "biomechanics", "kinematics" and "kinetic".

3.2.2 Inclusion and exclusion criteria

This review is limited to human subjects and includes all articles written in English, in full texts, examining frontal-plane biomechanics (both kinematics and kinetics) of single-leg landing in any direction, namely forward, lateral and medial (or synonyms). Landing on one leg was chosen because most lower-extremity injuries and real landings occur on a single leg, and examining bilateral tasks may not be useful for unilateral deficits as it may mask the functional limitations of the lower extremities involved during screening and rehabilitation (Myer et al., 2011). Also, is increases the demands on the limb due to the increasing landing impact on the musculature as the base of support decreases (Pappas et al., 2007). It is limited to the frontal plane of movement because the sagittal plane of movement is widely covered in the literature, while the frontal plane of movement is closely associated with injury and the suggested position of injury (position of no return) includes movement that mostly occurs in the frontal plane of movement, such as HADD and knee valgus (Souza & Powers, 2009; Hewett et al., 2005).

Studies were excluded if they were written in a language other than English, were an abstract only, examined the quantity of a task only (distance, height or time) or examined a bilateral task. The ‘word’ biomechanics is a term that mainly describes the motion and forces around body parts. Both components are important as force is the cause of motion. As the majority of previous studies focus on either kinetics or kinematics alone, such studies were also excluded in order to understand the full picture of joint biomechanics and match some of the objectives of the SR, which are to explore studies that examine both of them in an SLL task and summarize their findings in order to compare them with our findings

No restriction was applied on country, gender, age, type of sport, population or recruitment method. Also, studies that examined lower-limb biomechanics post-intervention were excluded. The search was limited to between 1995 and 2015. The rationale for selecting this period is that Lichtenstein, Yetley, and Lau (2008) suggest that a systematic review needs to cover twenty years at least. Moreover, the use of functional tests became more common during this period.

3.2.3 Study identification

Initially, the researcher reviewed the titles of all studies that were collected via the search strategy and then excluded all duplicates. After this, two reviewers (the researcher and Ziyad Nematallah, a PhD student) independently reviewed the titles and abstracts. Any unrelated studies were excluded. Then, full texts of all articles which potentially met the inclusion criteria were obtained. In accordance with predefined inclusion criteria, the reviewers reviewed the full texts. In cases where there was insufficient information to determine whether a study was eligible for inclusion or not, such as an abstract only, the full text was requested directly from corresponding authors via a Researchgate account or email to identify such information. All studies were subjected to this study’s inclusion and exclusion criteria.

3.2.4 Data extraction

Initial data were extracted from all papers that potentially met the inclusion criteria using JBI-SUMARI data-extraction tools (Appendix I). This tool was designed by the Joanna Briggs Institute to help researchers in the health field appraise and synthesise the suitability of evidence. It includes information about study design, participant details, inclusion and exclusion criteria and descriptions of interventions and outcomes.

3.2.5 Assessment of methodological quality and risk of bias

A modified version of the Downs and Black checklist was used to evaluate the quality of methods and risk of bias for all included articles (Hart et al., 2015). 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 & Black, 1998). Therefore, 15 scores (all of them were reliable in the original version), were included in the version used in the current study. A score of 12 or more suggests high methodological quality while 10–11 suggests moderate quality and less than 10 scores suggests low quality (Munn, Sullivan, & Schneiders, 2010).

3.3 Results

3.3.1 Search strategy

The results of the search strategy and a hand search are presented in a PRISMA flow diagram. As Figure 3.1 shows, a total of 4,028 papers were identified. Duplicate studies were then excluded (n = 1,860). To exclude clearly irrelevant papers, the titles and abstracts of all studies were critically reviewed by applying the search terms. A total of 1726 studies excluded as they were irrelevant. On reviewing the full texts of 442 studies, 433 articles were excluded because they had one or more of the flowing; examined an SLL test with regard to quantity, examined the sagittal plane only, examined the biomechanics of the ankle only, used an external support (e.g. orthosis), examined tasks other than landing, examined the effect of intervention on landing, examined kinematics or kinetics only, examined a bilateral task, were written in a language other than English or were unrelated systematic reviews or theses. Therefore, the full texts of nine studies were retained for review.

3.3.2 Studies descriptions and appraisals

The ability to evaluate research quality is a crucial component of any systematic review. There were no RCTs that met the inclusion criteria. The demographics of participants who were examined in the included studies are listed in Table 3.1. Variables of interest that were examined in the included studies are described in Tables 3.2 & 3.3. A summary of the included studies' descriptions is presented in Table 3.4.

Most of the studies reported ≤ 9 scores, which indicates low quality. Three studies reported scores of 9 or 10, indicating moderate quality. Only one study reported a score of 12, indicating high quality. The critical-appraisal process for assessment of the included studies' methodologies is summarised in Table 3.5. The studies included in this review examined 252 subjects, including 179 women, 13 of which were ACLR patients and 12 were PFPS patients. Groups of female subjects were examined in four studies (Ortiz et al., 2011; Jones et al., 2014; Dos Reis et al., 2015; Myer et al., 2015b), while a group of male subjects was examined in one study (Marshal et al., 2015).

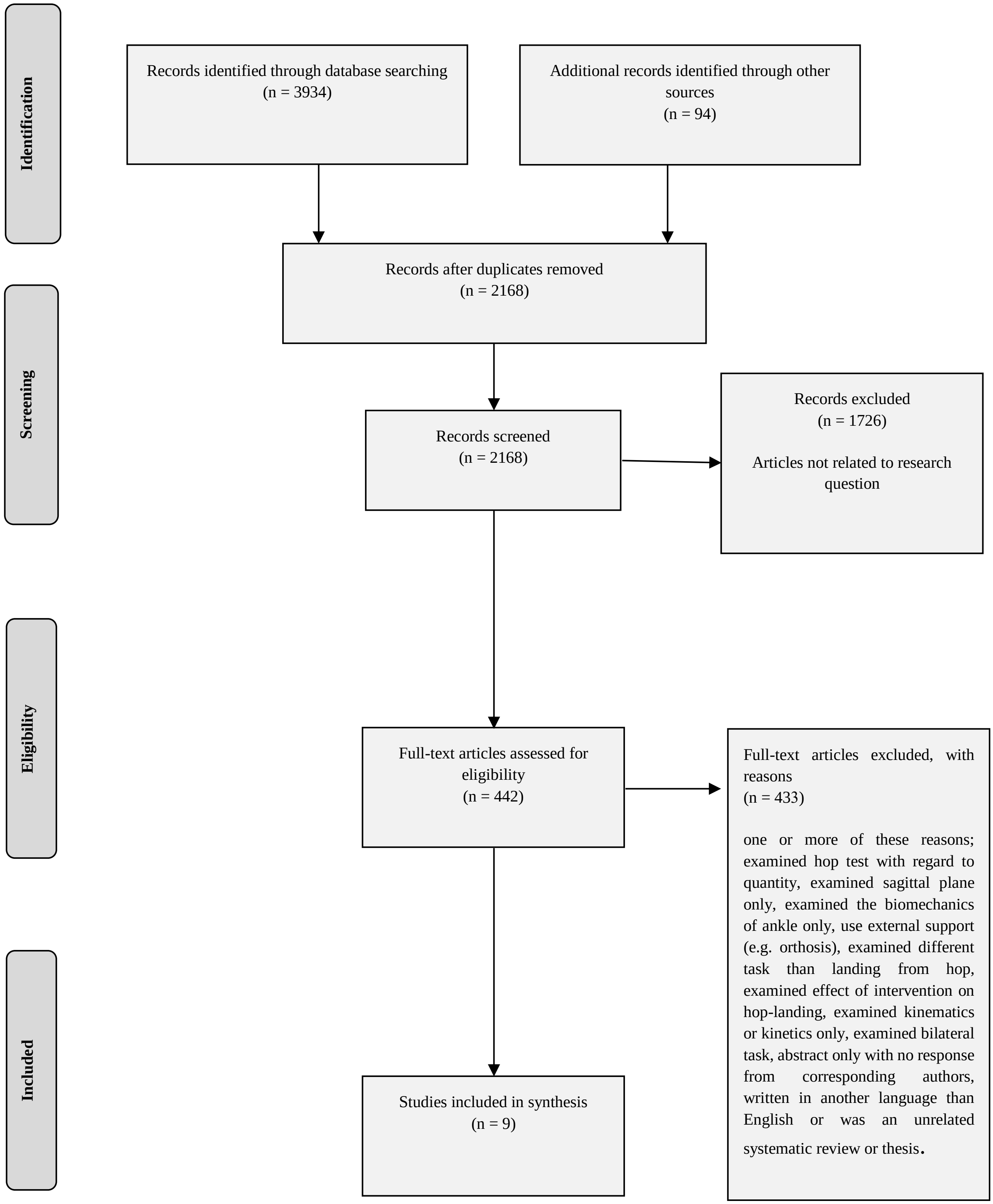


Figure 3.1: PRISMA flow diagram

Table 3.1 : Subjects' demographics (mean ± SD) for included studies

Study	Number	Age (years)	Height (cm)	Weight (kg)	Sporting participation	Level
	16 M	28.8 ± 3.9	181.7 ± 7.4	81 ± 10.4		
Pappas et al. (2007)	16 F	28.2 ± 5.4	167 ± 5.9	59 ± 5.8	Recreational athletes	University & college
	12 M	25 ± 4	177.3 ± 5.8	71.3 ± 5.5		
Orishimo et al. (2009)	21 F	27 ± 5	167.5 ± 4.9	57.9 ± 6.3	Ballet dancers	Professional
	13 F (ACLR)	25.4 ± 3.1	167.5 ± 5.9	63.2 ± 6.7		
Ortiz et al. (2011)	15 F (healthy)	24.6 ± 2.6	164.7 ± 6.3	58.4 ± 8.9	Physically active	Unknown
	7 M	25 ± 6.4	171 ± 6.7	69.7 ± 10.7		
Alenezi et al. (2014)	8 F	26 ± 3.5	163 ± 5.4	63 ± 8	Athletes	Recreational
Jones et al. (2014)	20 F	21 ± 3.9	163 ± 8	58.4 ± 6.4	Soccer players	Unknown
	20 M	27 ± 6	184 ± 7	73.5 ± 9.4		
	20 F	25 ± 5	170 ± 7	56.9 ± 6	Ballet dancers	Professional or modern
Orishimo et al. (2014)	20 M	22 ± 2	185 ± 7	78.8 ± 13.6		
	20 F	20 ± 2	176 ± 8	67.6 ± 7.5	Team-sport athletes	Colligate
	12 F (PFPS)	23.5 ± 2.1	171 ± 13	55.3 ± 4.8		
Dos Reis et al. (2015)	20 F (no pain)	23.1 ± 3.3	165.5 ± 12	55.9 ± 7.1	Physically active	Unknown
Myer et al. (2015b)	12 F	15.3 ± 0.6	169 ± 4	58.36 ± 6	Varsity & volleyball players	high school
Marshal et al. (2015)	20 injury free	20.4 ± 1.0	186 ± 8	98.4 ± 9.9	Rugby players	Elite

M = Male, F= Female, ACLR = Anterior cruciate ligament reconstruction, PFPS = patellofemoral pain syndrome.

In four studies, subjects of both genders were examined (Pappas et al., 2007; Orishimo et al., 2009; Alenezi, Herrington, Jones, & Jones, 2014; Orishimo, Liederbach, Kremenic, Hagins, & Pappas, 2014). Landing off a 30cm platform was examined in all studies except Pappas et al. (2007) and Myer et al. (2015b), who used platforms of 40 cm and 31cm, respectively, while no platform was used in two studies (Ortiz et al., 2011; Dos Reis et al., 2015). All studies collected both kinematic and kinetic data using a 3D motion analysis system and a force platform. In three studies, motion data and force platform data were sampled at 240 Hz and 1200 Hz, respectively (Pappas et al., 2007; Jones et al., 2014; Alenezi et al., 2014), while two studies used 250 Hz and 2500 Hz for sampling motion and force data, respectively (Orishimo et al., 2009; Orishimo et al., 2014).

Table 3.2: Summary of kinematic variables of interest reported by included articles

Authors	***H*FLEX/EXT**	***H*ADD/ABD**	***H*INT/EXT**	***K*FLEX/EXT**	***K*VAR/VAL**	***K*INT/EXT**
Pappas et al. (2007)	×	✓	×	✓	✓	×
Orishimo et al. (2009)	✓	✓	×	✓	✓	×
Ortiz et al. (2011)	✓	✓	×	✓	✓	×
Alenezi et al. (2014)	✓	✓	✓	✓	✓	✓
Jones et al. (2014)	×	✓	✓	×	✓	✓
Orishimo et al. (2014)	×	×	×	×	✓	×
Dos Reis et al. (2015)	✓	✓	✓	✓	✓	×
Myer et al. (2015b)	✓	✓	✓	✓	✓	✓
Marshal et al. (2015)	✓	✓	×	✓	✓	×

All variables are angles, HFLEX/EXT= Hip flexion/extension, HADD/ABD = Hip adduction/abduction, HINT/EXT = Hip internal rotation/external rotation, KFLEX/EXT = Knee flexion/extension, KVAR/VAL = Knee varus/valgus, KINT/EXT = Knee internal rotation/external rotation.

Table 3.3: Summary of kinetic variables of interest reported by included articles

Authors	HFLEX/EXT	HADD/ABD	HINT/EXT	KFLEX/EXT	KADD/ABD	GRF
Pappas et al. (2007)	×	×	×	×	×	✓
Orishimo et al. (2009)	✓	✓	×	✓	✓	✓
Ortiz et al. (2011)	×	×	×	✓	✓	×
Alenezi et al. (2014)	✓	✓	×	✓	✓	✓
Jones et al. (2014)	×	×	×	×	✓	×
Orishimo et al. (2014)	×	✓	×	×	✓	×
Dos Reis et al. (2015)	✓	✓	×	✓	✓	×
Myer et al. (2015b)	✓	✓	✓	✓	✓	×
Marshal et al. (2015)	✓	✓	×	✓	✓	✓

All variables are moments, apart from GRF, HFLEX/EXT= Hip flexion/extension, HADD/ABD = Hip adduction/abduction, HINT/EXT = Hip internal rotation/external rotation, KFLEX/EXT = Knee flexion/extension, KADD/ABD = Knee adduction/abduction, GRF = Ground reaction force.

Marshal et al. (2015) used 200 Hz and 1000 Hz for sampling motion and force data, respectively. In the study by Myer et al. (2015b), three different sampling rates were used as the study was conducted across three different centres. In the first centre, 1200 Hz was used for force data sampling, while motion sampling was not mentioned. In the second and third centres, 200 Hz and 240 Hz were used to sample motion data, while 1000 Hz and 1200 Hz were used for force-data sampling, respectively.

Table 3.4 : Description of the included studies

Author	Task	Methods	Variables	Result	
Pappas et al. (2007)	SLL 40 cm platform	3D 8 cameras 1 force platform	Peak knee flexion (°) HADD (°) Knee valgus (°) VGRF (times BW)	72.2 ± 12.2 -8.4 ± 6 0.96 ± 5 3.2 ± 1.3	No gender difference
Orishimo et al. (2009)	SLL 30 cm platform	3D 8 cameras 1 force platform	Hip flexion (°) (+) HADD (°) (+) Hip extension Mom (Nm/kg) (+) Hip abduction Mom (Nm/kg) (+) Knee flexion (°) (+) Knee abduction (°) (+) Knee extension Mom (Nm/kg) (+) Knee adduction Mom (Nm/kg) (+) VGRF (times BW)	M (20 ± 16.6) F (28.7 ± 10.2) M (4.8 ± 5.3) F (0.9 ± 5.4) M (1 ± 0.7) F (0.8 ± 0.6) M (1.3 ± 0.4) F (1 ± 0.5) M (59.2 ± 12.5) F (58.7 ± 5.5) M (-3.2 ± 4.3) F (-1.7 ± 11.1) M (1.6 ± 0.5) F (1.4 ± 0.5) M (-0.6 ± 0.3) F (-0.4 ± 0.4) M (4.2 ± 0.7) F (3.9 ± 0.5)	
Ortiz et al. (2011)	Side-to-side hop (divided into side hop & crossover hop)	3D 4 cameras 2 force platforms	Hip flexion (°) HADD (°) Knee extension Mom (Nm/kg) Knee valgus Mom (Nm/kg)	Side hop (39.90) Crossover hop (14.08) Side hop (3.99) Crossover (8.54) Side hop: Control group (2.96) ACLR group (3.97). Crossover hop: Control group (7.62) ACLR group (2.13) *. Side hop: Control group (1.16), ACLR group (6.96) * Crossover hop: Control group (1.16) ACLR group (5.59) *.	
Alenezi et al. (2014)	SLL 30 cm platform	3D 10 cameras 1 force platform	Hip flexion (°) HADD (°) Hip flexion Mom (Nm/kg) HADD Mom (Nm/kg) Knee flexion (°) Knee valgus (°) Knee flexion Mom (Nm/kg) Knee valgus Mom (Nm/kg) VGRF (times BW)	Within day (49.83) SEM (3.26) Between days (50.19) SEM (2.97) Within day (8.56) SEM (1.53) Between days (7.70) SEM (1.29) Within day (-2.39) SEM (0.21) Between days (-2.51) SEM (0.29) Within day (-1.93) SEM (0.16) Between days (-2.01) SEM (0.11) Within day (70.27) SEM (3.35) Between days (70.27) SEM (3.27) Within day (-9.36) SEM (1.44) Between days (-8.89) SEM (1.14) Within day (3.33) SEM (0.11) Between days (3.35) SEM (0.11) Within day (-0.51)) SEM (0.08) Between days (-0.57) SEM (0.08) Within day (4.42) SEM (0.24) Between days (4.45) SEM (0.25)	
Jones et al. (2014)	SLL 30 cm platform	3D Cameras number (N/A) 1 force platform	HADD (+) /Abduction (-) (°) Hip internal (+) external (-) rotation (°) Peak-knee abduction (-) (°) Knee internal (+) external (-) rotation (°) Knee abduction Mom (Nm/kg)	3 ± 5 10 ± 8 -7 ± 6 5 ± 6 0.28	

Continued

Continued Table 3.4

Author	Task	Methods	Variables	Result
Orishimo et al. (2014)	SLL 30 cm platform	3D 8 cameras 1 force platform 20 reflective markers	Knee abduction (-) adduction (+) (°) Peak-knee flexion (°) HADD Mom (Nm/kg) Peak-knee flexion Mom (Nm/kg)	Dancers M (6) F (2.5), Athletes M (5) F (-3) * Dancers M (54.3 ± 6.3) F (57 ± 6.1), Athletes M (54.2 ± 9.1) F (56 ± 5.4) Dancers M (3.2) F (2.2) *, Athletes M (3.1) F (3) Dancers M (2.8 ± 0.4) F (2.5 ±0.4), Athletes M (2.8 ± 0.6) F (2.9 ±0.3)
Dos Reis et al. (2015)	SLTH	3D 8 cameras 1 force platform 23 reflective markers	Hip flexion (°) HADD (°) Hip-internal rotation (°) Knee flexion (°) Knee adduction (°) Hip abduction Mom (Nm/kg) Hip extension Mom (Nm/kg) Knee abduction Mom (Nm/kg) Knee extension Mom (Nm/kg)	Control (58.6 ± 3.7) PFPS (54.4 ± 5.4) * Control (6.9 ± 0.6) PFPS (10.3 ± 0.6) * Control (8.9 ± 0.9) PFPS (12.5 ± 3.3) * Control (56.7 ± 4.9) PFPS (47.8 ± 2.8) Control (7.8 ± 3) PFPS (8.4 ± 2.2) Control (1.8 ± 0.5) PFPS (2.2 ± 0.2) * Control (2.9 ± 0.5) PFPS (2.8 ±0.5) Control (0.9 ± 0.3) PFPS (2.1 ± 0.4) * Control (2.8 ± 0.4) PFPS (1.9 ± 0.3) *
Myer et al. (2015b)	SLCD 31 cm platform	3D 10 cameras in the 1st centre, 18 cameras in the 2nd centre, 8 cameras in the 3rd centre 1 force platform 43 reflective markers	Hip flexion (°) HADD (°) Hip-internal rotation (°) Knee flexion (°) Knee abduction (°) Knee-internal rotation (°) Hip flexion Mom (Nm/kg) HADD Mom (Nm/kg) Hip internal Mom (Nm/kg) Knee flexion Mom (Nm/kg) Knee abduction Mom (Nm/kg) Knee internal Mom (Nm/kg)	1st centre/ LT (58.6) RT (59.2), 2nd centre/ LT (53.2) RT (51.7), 3rd centre/ LT (51.7) RT (53.2) 1st centre/ LT (9.9) RT (14.2), 2nd centre/ LT (8) RT (10.2), 3rd centre/ LT (8.7) RT (14) 1st centre/ LT (-4.5) RT (-5.7), 2nd centre/LT (-6) RT (-6.7), 3rd centre/ LT (-5.8) RT (-6.7) 1st centre/ LT (65.9) RT (68.2), 2nd centre/ LT (61.3) RT (65.5), 3rd centre/ LT (66.2) RT (65.3) 1st centre/LT (9) RT (10.6), 2nd centre/ LT (8.9) RT (9.2), 3rd centre/ LT (8.3) RT (12.4) 1st centre/ LT (6.8) RT (2), 2nd centre/ LT (9.3) RT (9.5), 3rd centre/ LT (9.1) RT (8.9) 1st centre/ LT (103.9) RT (105.3), 2nd centre/ LT (80.9) RT (83.9),3rd centre/ LT (91.7) RT (80.5) 1st centre/ LT (90.6) RT (104.3), 2nd centre/ LT (90.9) RT (92), 3rd centre/ LT (85.9) RT (95.1) 1st centre/ LT (43.8) RT (47.6), 2nd centre/ LT (48.8) RT (40.3), 3rd centre/ LT (37.9) RT (49.6) 1st centre/ LT (134.2) RT (129.3), 2nd centre/ LT (141.2) RT (145.3), 3rd centre/ LT (144.9) RT (138.8) 1st centre/ LT (10.3) RT (6.7), 2nd centre/ LT (5.4) RT (15.7), 3rd centre/ LT (10.4) RT (11.3) 1st centre/ LT (1.5) RT (1.2), 2nd centre/ LT (8.8) RT (3.5), 3rd centre/ LT (2.9) RT (5.4)

Continued

Continued Table 3.4

Author	Task	Methods	variables	Result
Marshal et al. (2015)	SLL SLHH 30 cm platform	3D 8 cameras 1 force platform	knee flexion (+)/extension (-) (°) knee varus (+)/valgus (-) (°) hip flexion (+)/extension (-) (°) HADD (+)/abduction (-) (°) hip extension (+)/flexion (-) (Nm/kg) hip abduction (+)/adduction (-) (Nm/kg) knee extension (+)/flexion (-) (Nm/kg) knee valgus (+)/varus (-) (Nm/kg)	Dominant (66.6 ± 8.8), Non-dominant (66.3 ± 8) Dominant (4.3 ± 5.6), Non-dominant (7.6 ± 8.5) Dominant (59.3 ± 10.9), Non-dominant (59.4 ± 9.1) Dominant (9.3 ± 5.6), Non-dominant (10 ± 3) Dominant (5.4 ± 2), Non-dominant (5 ± 1.3) Dominant (2.7 ± 0.7), Non-dominant (2.2 ± .8) Dominant (3.1 ± 0.4), Non-dominant (3.1 ± 0.3) Dominant (1.9 ± 0.4), Non-dominant (2 ± 0.5)

SLL = Single-leg landing, cm = Centimetre, 3D = Three-dimensional, SLTH = Single-leg triple hop, SLCD = Single-leg cross drop, SLHH = Single-leg hurdles hop, HADD = Hip adduction, VGRF = Vertical ground reaction force, BW = Body weight, Mom = Moment, Nm/kg = Newton meter per kilogram. M = Male, F = Female, PFPS = Patellofemoral pain syndrome, LT = Left, RT = Right, N/A = not available.

Only one study used a sampling frequency determined in a pilot study (Dos Reis et al., 2015). This study used 100 Hz and 400 Hz for motion and force data, respectively. With regard to kinematic data filtration, a low-pass fourth-order Butterworth filter at a cut-off frequency of 12 Hz was used in four studies (Jones et al., 2014; Alenezi et al., 2014; Dos Reis et al., 2015; Myer et al., 2015b), while a cut-off frequency of 6 Hz was used in two studies (Pappas et al., 2007; Ortiz et al., 2011). A cut-off frequency of 15 Hz was used by Marshal et al. (2015), while two studies used a cut-off frequency of 10 Hz (Orishimo et al., 2009; Orishimo et al., 2014). Kinetic data were filtered using a cut-off frequency of 25 Hz in two studies (Jones et al., 2014; Alenezi et al., 2014), while cut-off frequencies of 6 Hz and 15 Hz were used in Ortiz et al. (2011) and Marshal et al. (2015), respectively. The remaining studies did not mention a cut-off frequency for kinetic data.

Table 3.5 : Methodological quality rating scores with the Modified Downs and Black Scale

	Item number															
Study	**1**	**2**	**3**	**4**	**5**	**6**	**7**	**8**	**9**	**10**	**11**	**12**	**13**	**14**	**15**	**Total**	**Methodology Quality**
Pappas et al. (2007)	1	0	1	0	0	1	1	0	1	1	1	0	1	0	0	**8**	Low
Orishimo et al. (2009)	1	1	1	1	0	1	1	0	1	0	1	0	1	0	0	**9**	Low
Ortiz et al. (2011)	1	1	1	1	1	1	0	1	1	1	1	1	1	0	0	**12**	High
Alenezi et al. (2014)	1	1	1	0	0	1	1	0	1	1	1	0	1	0	0	**10**	Moderate
Jones et al. (2014)	1	1	0	0	0	1	1	0	1	1	1	1	0	0	0	**8**	Low
Orishimo et al. (2014)	1	1	1	0	1	1	1	0	1	0	1	1	1	1	0	**11**	Moderate
Dos Reis et al. (2015)	1	1	0	0	1	1	1	0	1	1	1	0	1	1	0	**10**	Moderate
Myer et al. (2015b)	1	1	0	0	0	1	0	0	1	1	1	1	0	0	0	**7**	Low
Marshal et al. (2015)	1	1	1	0	0	1	1	0	1	1	1	0	1	0	0	**9**	Low

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

A power calculation was performed in two studies (Orishimo et al., 2014; Dos Reis et al., 2015). Whilst the landing strategy was clearly described in most of the studies, three studies asked the subjects to perform the task with hands crossed over their chest (Orishimo et al., 2009; Alenezi et al., 2014; Marshal et al., 2015). Statistical analysis was appropriate and clearly described in all studies. However, only three studies reported confidence intervals (Alenezi et al., 2014; Dos Reis et al., 2015; Marshal et al., 2015). Shoes were standardized in most of the studies, while subjects were examined barefoot in one study (Dos Reis et al., 2015). Orishimo et al. (2009) examined ballet-dancer subjects while wearing sport shoes. The surgery or graft type was not controlled in the study of Ortiz et al. (2011). Also, the length of time after surgery was 1–16 years. Only three studies reported VGRF (Pappas et al., 2007; Orishimo et al., 2009; Alenezi et al., 2014), which is considered a risk factor for knee injury (Yu et al., 2006; Boling et al., 2009).

3.4 Discussion

SLL tasks are commonly seen in different types of sport and usually used as a rehabilitation exercise and screening tool before returning to sport. Therefore, the purpose of this review was to review the literature that investigated the biomechanics (kinematics and kinetics) of the lower extremities during any task that includes landing on one leg in any direction. Lower-limb biomechanics during athletic tasks are shown to be associated with knee injury (Malinzak et al., 2001). While frontal-plane movement seems to be linked to injury, it presents the worst scenario for knee loading when combined with sagittal- and/or transverse-plane movement (Shin et al., 2011; Olsen et al., 2004). In the literature, the sagittal plane of movement has had the most research during SLL. Due to the marked differences between those studies in terms of objectives, methods, interventions and outcomes, it seems difficult to draw a conclusion about their findings for SLL biomechanics. This review was limited to studies that examined the frontal plane of movement as the position of injury (position of no return), including movements that mostly occurred in the frontal plane such as HADD and knee abduction. Most studies examined kinematics only, kinetics only and/or the sagittal plane only. These were excluded because they did not match the inclusion criteria for this review.

All included studies examined biomechanics during SLL in a forward direction except for Ortiz et al. (2011), Myer et al. (2015b) and Marshal et al. (2015) who evaluated side-to-side hopping, single-leg cross-landing and single-leg hurdle hop (laterally), respectively. Some studies employed some other tasks, such as SLS (Alenezi et al., 2014), cutting and pivoting (Jones et al. 2014), running cut (Marshall et al., 2015) and DLL (Pappas et al., 2007). However, this review was limited to any task that includes landing on a single leg; therefore, other tasks will not be discussed.

Whilst the landing strategy was clearly described in most of the studies, three studies asked the subject to perform the task with hand crossed over the chest (Orishimo et al., 2009; Alenezi et al., 2014; Marshal et al., 2015) which was not matching the nature of landing. Subjects were examined barefoot in the study of Dos Reis et al. (2015) which is not typical for such a task. Orishimo et al. (2009) examined ballet dancer subjects while wearing sport shoes, while it is known that the nature of ballet is performed with specific footwear (ballet shoes or slipper). The surgery type or graft type were not controlled in the study of Ortiz et al. (2011) which may affect the subject's performance (Wagner, Kääb, Schallock, Haas, & Weiler, 2005). Also, the range of time after surgery was (1-16) years, which may make the performance varied between subjects. A minimum of 6 trials for each task was performed in Jones et al. (2014) study, so fatigue effect may present.

Orishimo et al. (2014) did not evaluate hip kinematic which is suggested to be a risk factor for knee injury (Powers, 2010; McLean et al., 2004a; Hewett et al., 2005). Moreover, selection bias might be present in this study as the dancers were selected for body type and the ability to perform balance exercises. This may lead to exclusion for those with poor performance. All studies appropriately interpreted the findings and linked it to the clinical practice in a logical manner.

The task in the study of Ortiz et al. (2011) may not be exactly a side hop as they defined the task as hopping toward the opposite side of the weight-bearing leg. So, the angle of hop may vary between subjects, which may affect the landing mechanics. Most of those who examined SLL asked subjects to drop off the platform directly onto a force platform. So, it seems that such a task is considered a drop landing. Apart from the abovementined study, all studies examined one task and one direction of SLL, except for Marshall et al. (2015) who examined SLL and a single-leg hurdle hop. Although the SLL might be the gold standard due to its good reliability (ICC 0.86 - 0.95) (Munro et al., 2012a; Alenezi et al., 2014; Myer et al., 2015b), the sensitivity of noticing functional limitations with this test is quite low (38–52%). Moreover, Narducci et al. (2011) suggest that the functional ability that required them to return to sport could not be evaluated using one isolated test. Employing other landing tests may increase the sensitivity of noticing functional limitations to 80 per cent (Reid et al., 2007). In addition, it may include other functional variables in multiplane movement which meet specific sport demands (Narducci et al., 2011). Clinicians usually use other directions in SLL tests, such as a sideways landing which is not fully examined in the included studies. Therefore, examining these tests is beneficial and can contribute to the literature.

Functional tests are usually used as rehabilitation exercises and as a screening tool for both injury and return to sport (Reiman and Manske, 2009; Reiman and Manske, 2011; Narducci et al., 2011). However, no study has examined the ability of SLL or a battery of SLL tests to predict readiness to return to sport. It would be clinically beneficial to have FPTs that have been validated within return-to-sport evaluations. This may help clinicians to decide which test or battery of tests should be used for making decisions about patients returning to sports. The shortage of literature to assist practitioners suggests that they may have to make decisions based only on intuition and experience (Narducci et al., 2011). When considering the huge variability in human movement between individuals, particularly in high-speed movement such as landing, large sample sizes are required (Holden, Boreham, & Delahunt, 2016). However, most of the reviewed studies included less than 20 subjects per sub-group, which may indicate insufficient power.

The reliability of SLL (Alenezi et al., 2014) and single-leg cross landing (Myer et al., 2015b) has been established in two studies. Alenezi et al. (2014) examined 15 recreational athletes, 8 females and 7 males (age 26 ± 3.5, 25 ± 6.4 years; height 163 ± 5.4, 171 ± 6.7 cm; mass 63 ± 8, 69.7± 10.7, respectively). Participants were captured using 3D motion analysis. They reported within-day reliability (combined average ICC = 0.90) to be higher than between days (combined average ICC = 0.78). Moreover, they reported excellent within-day reliability for lower-limb kinematic and kinetic variables during SLL, ranging between 0.80 and 0.97, with CI ranging between 0.39 and 0.99, except for knee-internal rotation angle and HADD moment which showed moderate to good reliability at 0.78 with CI = (-0.33 - 0.93) and 0.63 with CI = (-0.08 - 0.88), respectively. Positively, SEM was reported for all variables which range between 1.22° and 4.16° for kinematic variables and between 0.01 and 0.13 Nm/kg for kinetic variables. Between-days reliability for kinematic variables was fair to excellent for all variables (0.48–0.96) (CI = 0.31–0.98). Between-days SEM ranged between 0.11° and 3.27°. The calibration anatomical system technique (CAST) model was employed in this study, it offers improved anatomical relevance compared to other models and reduces skin artefacts (Cappozzo, Catani, Della Croce, & Leardini, 1995). Myer et al. (2015b) reported between-centre reliability for single-leg cross-landing using 3D motion analysis. Female participants ($n = 25$), high-school volleyball players, were recruited for this study. However, only 12 subjects completed the study. Kinematic variables exhibited good reliability with a coefficient of multiple correlation (CMC) ≥ 0.75, apart from lateral-trunk flexion, which exhibited poor reliability. SEM for sagittal-plane hip and knee motion was 9.3° and 7.3°, respectively. However, the frontal plane showed less SEM (hip = 4.6°; knee = 2.5º). Kinetic variables for the sagittal plane were highly reliable, with CMC ≥ 0.79. However, transverse-plane variables showed moderate to good reliability, with CMC > 0.72. Frontal-plane moment was also reliable, with CMC > 0.71. The data-collection process of the study by Myer et al. (2015b) was conducted in three different centres. Each centre used their own instrument. The differences that exist between the centres' instrumentation, such as differences in numbers of cameras, sampling frequencies and cut-off frequencies may affect the results. However, it seems that this might be unavoidable but matches the objective of the study, which was to examine between-centres reliability. Moreover, the level of activity may influence the results, as volleyball players might be able to land better than players of other sports because volleyball includes such a task, which may make the players accustomed to doing it.

In a comparison between groups and sexes, Orishimo et al. (2014) examined the biomechanics of SLL in 40 professional dancers (20 male and 20 female) and 40 collegiate team athletes (20 male and 20 female). Females athletes landed with greater knee valgus than those in other groups. HADD moment was lower in female dancers than those in the other group. However, the effects of group, gender and interaction were not statistically different. Such findings support Pappas et al.'s (2007) study, which found that females landed with greater knee-valgus angle during SLL and DLL compared to males. In addition, Orishimo et al. (2009), who used multivariate analysis of variance (MANOVA), the same as Orishimo et al. (2014), reported no gender differences in SLL kinematics (Table 3.4) and VGRF (4.2 ± 0.7 BW for males, 3.9 ± 0.5 BW for females) between male and female ballet dancers. Although joint moments and VGRF were normalized to body mass, differences in age and weight between team-sport subjects and female dancer in Orishimo et al.'s (2014) study may still influence the findings. Also, sport-team athletes, who were considered as collegiate athletes in this study, may not be suitable to compare with professional dancers. In such a case, the level of experience and training, as well as body type, may play a crucial role in findings variations. Shoe-differences in Orishimo et al.'s (2009) study may influence VGRF. Nevertheless, dancers of a lesser level of experience and training may exhibit different landing biomechanics.

Ortiz et al. (2011) compared uninjured females (n = 15) with ACLR females (n = 13). Subjects were captured while performing side-to-side hopping, which was divided into SLL 10 times repeatedly and side-to-side across two lines. Both groups exhibited similar knee- and hip-joint kinematics during both tasks. During crossover hopping, hip-flexion and adduction angles were greater (41.08° and 8.54°, respectively) than during side-to-side hopping (38.90° and 3.99°, respectively) in both groups. Knee extension and adduction moment were greater in the control group during crossover hopping, while they were greater during side-to-side hopping for the ACLR group. However, a 60 Hz sampling rate, with a 6 Hz low-pass filter, was employed in this study. Although it seems logical, given the data of interest, it may cause variability in kinematic data. Even though uninjured females in comparison to ACLR females may show injury-predicting factors during sport tasks, it seems valuable if male subjects were also employed as a control group to represent the right biomechanics and neuromuscular control during tasks. Moreover, the author addressed the level of performance of ACLR participants in this study being greater than the average of those post-ACL. Thus, the findings might be applicable only to women with the same level of performance.

Pappas et al. (2007) examined the biomechanics of SLL and DLL and the effect of gender. Recreational athletes, of which there were 32, half of them female, were examined while performing SLL and DLL off a 40 cm platform. The results showed that SLL was performed with increased knee valgus (0.96 ± 5 for SLL and -1.4 ± 5.9 for DLL), decreased knee flexion (peak) (72.2 ± 12.2 for SLL and 93.3 ± 17.6 for DLL) and decreased hip adduction (-8.4 ± 6 for SLL and -1.13 ± 3.3 for DLL. Compared to men, women exhibited greater knee valgus and GRF in both types of landings. This may explain the gender differences in ACL injury incidence. However, no differences were found in the interactions between gender and landing type. Positively, the recruited subjects in this study were matched in age and sport activity to hours per week. However, height and weight were significantly different between the genders, which may influence the findings, particularly GRF which was normalised to body weight.

In a cross-sectional study comparing between females with and without PFPS, Dos Reis et al. (2015) examined the biomechanics of single-leg triple hop (SLTH), specifically, the transition period between the first two hops. Twenty women physically active and age-matched were recruited for each group from an outpatient rehabilitation programme. Subjects were captured using a 3D motion analysis system while performing SLTH. Kinematic and kinetic data were collected. The result showed that the PFPS group landed with greater hip adduction (10.3 ± .6 for PFPS group and 6.9 ± .6 for a control group) and internal rotation (12.5 ± 3.3 for PFPS group and 8.9 ± 0.9 for a control group) and decreased knee (PFPS = 47.8 ± 2.8, control group = 56.7 ± 4.9) and hip (PFPS = 54.4 ± 5.4, control group = 58.6 ± 3.7) flexion. Knee- (PFPS = 2.1 ± 0.4, control group = 0.9 ± 0.3) and hip-abductor (PFPS 2.2 ± 0.2, control group = 1.8 ± 0.5) internal moment was also greater in the PFPS group.

Positively, a power calculation was performed prior to this study. However, it was calculated depending on the maximum knee flexion reported in previous studies. Therefore, the other variables may still be underpowered. Moreover, as this study examined only the transition period between the first two hops, peak-knee angle might be greater. This transition period might be less important to assess because the landing phase was found to be more stressful, particularly for the ACL (Chappell et al., 2002). Moreover, Kirkendall and Garrett (2000) reported that most knee injuries happened during landing and greater GRF was reported during the landing phase (Paterno et al., 2007; Decker et al., 2003). The task was performed barefoot, which might not be typical for such a task. It might be better if shoes were standardised between subjects.

Biomechanical symmetry in Rugby Union players was examined by Marshal et al. (2015). Twenty elite rugby players were recruited for this study (age = 20.4 ± 1 years, mass = 98.4 ± 9.9 kg, height = 1.86 ± 0.08 m). Participants were captured using 3D motion analysis while performing SLL, single-leg hurdle hops (SLHH) (laterally) and running cut. As running cut is not an area of interest in this review, it will not be discussed. In this study, kinematic and kinetic data for the frontal and sagittal planes of movement were collected. There were differences between the limbs for pelvic contralateral drop in the drop landing and hurdle hop (dominant = - 12.1° ± 4°, non-dominant = - 8.9° ± 3.4°) (dominant = - 1.4° ± 4.7°, non-dominant = 3.1° ± 4.1°), respectively. The rest of the variables showed no differences in limb symmetry in drop landing and hurdle hop. However, the limb symmetry index (LSI) ranged between 0–143 per cent in drop landings and 1–264 per cent in hurdle hops. Landing in this study was performed with arms across the chest to minimize the effect of arm movement. However, in a real situation, arm movement is unrestricted. The researcher suggested the results of this study to be normative data for this task. Nevertheless, data from only 20 subjects who participate in a single sport might not be applicable to larger or different sport populations.

To find the relationship between different tasks in terms of dynamic knee valgus, Jones et al. (2014) conducted a study on 20 female soccer players (age = 21 ± 3.9 years, mass = 58.4 ± 6.4 kg, height = 1.65 ± 0.08 m) who were captured using a 3D motion analysis system while performing SLL from a 30 cm platform, cutting and pivoting. Kinematic and kinetic data were collected. The kinematic and kinetic results for SLL are presented in Table 3.4. They also reported that strong correlation was found between tasks for knee-abduction angle (r = 0.63 - 0.86). With regard to knee-abduction moments, only moderate correlation was found between cutting and SLL (r = 0.46), pivoting and SLL (r = 0.43), pivoting and cutting (r = 0.56). However, all correlations were statistically significant, suggesting that poor performance in SLL may be associated with poor performance in other tasks. However, this study only included female soccer players, which may limit the generalizability of the findings to males or people in other sports. Furthermore, the correlation between tasks only checked for the right leg. A comparison of both legs (dominant and non-dominant) might add valuable information to this study.

3.5 Conclusion

This review has found that only SLL in a forward direction has been tested in most of the literature using a 3D motion analysis system. The findings of this systematic review suggest that there is a lack of evidence about the biomechanics of other directions of SLL and about the utility of 2D motion analysis to evaluate the biomechanics of multidirectional SLL. Also, there is a shortage of literature showing to what extent the different directions of SLL correlate. This project will consider examining multidirectional SLL and the correlation between different directions of SLL using 2D and 3D motion analysis. This may contribute to a better understanding of the similarities and differences between them, which, in turn, may fill a gap between the research environment and field and the clinical environment, and allow examining patients in clinics and players in the field.

4. Study two: Within-day and between-days reliability of lower-limb biomechanics using two-dimensional and three-dimensional movement analysis systems during multidirectional single-leg landing

4.1 Study aims

1- To examine within-day and between-days reliability for lower-extremity biomechanics using 2D motion analysis during multidirectional SLL.
2- To establish SEM for 2D biomechanical measurements during multidirectional SLL.
3- To examine within-day and between-days reliability for lower-extremity biomechanics using 3D motion analysis during multidirectional SLL.
4- To establish SEM for 3D biomechanical measurements during multidirectional SLL.

4.2 Background

Abnormal lower-limb mechanics during a variety of sporting manoeuvres can result in non-contact injury (Willson & Davis, 2008). This theory has been examined in several different studies. Hewett et al. (2005) reported that during bilateral jump landing, knee-valgus angle and moment can predict the risk of ACL injury in female athletes. Such findings were supported by McLean, Huang, Su, and Van Den Bogert, (2004b), who found that knee-valgus moment is the most sensitive component to change in the moment pattern during a cutting manoeuvre. Therefore, investigating lower-limb biomechanics during high demand sport tasks may provide a better understanding and improve rehabilitation for non-contact lower limb injury.

SLL is a common task performed in many sports. Furthermore, it is easy to implement in clinical and/or sport training, to then be used in evaluating functional performance and quality of movement (Myer et al., 2015b). Some studies have shown evidence that using an SLL test allows differentiation between injured and uninjured legs in an ACL injury population and between patient and control groups (Eastlack ,1999). Fitzgerald et al. (2000) also showed evidence that SLL can predict the risk of knee instability post-ACL injury.

The literature indicates that a variety of different SLL tests are used as screening tools for a return to sport and as an exercise for injury rehabilitation. Xergia et al. (2013) also support landing being the most popular test used to determine a return to sport, as it is a functional task which gives information about neuromuscular deficits (Paterno et al., 2010).

Different landing tests are examined in the literature. However, the examination of functional performance after lower-limb injury is mostly limited to the use of only a single test, which has been reported to result in low sensitivity in noticing functional limitations (38–52%). Combining SLL tests with other landing tests, such as sideways landing, may increase the sensitivity to 80 per cent and raise the ability to understand inconsistencies in performance (Reid et al., 2007). Sideways landing is a common task that can be seen in different sports, and even in normal daily activities. It is also commonly used as both a screening tool and a rehabilitation exercise. Although forward SLL has been examined in the literature, the focus is on the sagittal plane of movement and its loading, which suggests the importance of investigating the biomechanics of the frontal plane of movement and its loading. With regard to sideways SLL, only MacLean et al. (2005) and Sorenson et al. (2015) have examined such a task, and they only reported kinematic data. Consequently, examining both the kinematics and kinetics of SLL in different directions should be done.

The gold standard for examining lower-limb biomechanics is a 3D motion analysis system, which has been used most research (Gao et al., 2012; Sled et al., 2010; Zeller et al., 2003; Ford et al., 2003; McLean et al., 2004a). This system allows accurate data collection for multiplane joint and multiple plane biomechanics during functional tasks. Although this system is very important for doing research and gives valuable information about lower-limb biomechanics, its extension to clinical settings (Willson & Davis, 2008) or larger sample sizes (Hewett et al., 2005) is limited due to the high financial cost (Nielsen & Daugaard, 2008) and the knowledge required to operate the system. Therefore, 2D motion analysis became popular in clinical practice. It only requires a digital video camera and digitizing software (Stenstrud et al. 2011). Many studies have used a 2D motion analysis system and concluded that 2D is reliable for calculating lower-limb kinematics (Stenstrud et al., 2011; Herrington, 2011; Noyes et al., 2005; Willson & Davis, 2008; Herrington & Munro, 2010).

Considering the importance and accuracy of 3D motion analysis alongside the accessibility and portability of 2D motion analysis, it can be concluded that each of them is important and needs to be investigated. This may fill a gap between the research environment, the field and the clinical environment, and allow examining patients in clinics and players in the field.

Regardless of the method, outcome measurements must provide consistent and repeatable values with small measurement errors in order to be valuable (Rankin & Stokes 1998). Therefore, examining the reliability and SEM of each of the aforementioned methods is essential and can provide researchers and clinicians with valuable information.

For motion analysis, both within-day and between-days reliability are important. Most published papers have reported that biomechanical variables of the lower extremities show an ICC ranging between 0.59 and 0.98 for both 3D and 2D motion analysis during different tasks, such as running, SLS, landing, side step, stair ascent and descent, single-limb step-down and SLL (Munro et al., 2012a ; Alenezi et al., 2014; Norris & Olson, 2011; Sled et al., 2010; Gao et al., 2012; Ford et al., 2003; McLean et al., 2004a; Mclean et al., 2005;; Hollman et al., 2009; Miller & Callister, 2009; Ferber, McClay Davis, Williams, & Laughton, 2002; Zeller et al., 2003). However, comparisons between these studies and interpretations of their results present some limitations regarding reliability and validity because of their using different methods and screening tasks. Calculation of reliability and SEM allows clinicians to differentiate the changes in performance caused by variability and true difference.

To the best of the researcher's knowledge, no study has examined reliability by presenting SEM for both kinematics and kinetics during a battery of SLL tasks. Therefore, the aim of this study was to establish reliability and SEM for 2D and 3D motion analysis systems during multidirectional SLL. The results of this study can then provide information about the levels of errors that may be inherent to examinations, and this, in turn, can determine the validity of these results (Kottner et al., 2011).

4.3 Study hypothesis

Based on previous literature, the hypothesises below were formulated.

Alternative hypotheses

H_1: There will be within-day and between-days agreement between repeated measurement scores for all 2D and 3D variables throughout all tasks.

H_2: Within-day measurement will be more reliable than between-days measurement for all 2D and 3D variables throughout all tasks.

H_3: Reliability of GRF measurement will be greater than reliability of kinematic and kinetic data in all tasks.

Null hypotheses

$H0_1$: There will be no agreement between repeated measurement scores for all 2D variables throughout all tasks.

$H0_2$: There will be no agreement between repeated measurement scores for all 3D variables throughout all tasks.

4.4 Methods

4.4.1 Participants

Twelve subjects were voluntarily recruited from the staff and student population of the University of Salford. Sample demographics are presented in Table 4.1. Participants were adult, healthy, moderately active (defined as the practice of any sport or exercise for at least half an hour, three times a week for at least the last six months), with normal balance [able to stand on one leg for 30 seconds with eyes closed (Atwater et al., 1990)], no history of lower extremity, pelvis or back injury, or surgery, one year prior to the study and able to perform the test's task independently. The age range was limited to 18–35 years as this is the expected age range for most athletes in most sports; athletes are more prone to injury and they are mostly the ones to whom our study would be applicable (Griffin, 2001). In this context, injury is any musculoskeletal complaint that can limit the subject’s ability to perform regular exercise. Individuals who had any pathology, injury or surgery of the lower extremities, which affects their physical activity, or cardiovascular, balance, neurological or pulmonary conditions were excluded from this study.

Table 4.1 : Sample demographics for reliability study

		Number	Mean	SD	Minimum	maximum
Age (years)	Male	6	28.3	5.7	20	35
	Female	6	26.8	2.9	24	31
	All	12	27.6	4.4	20	35
Height (m)	Male	6	1.7	0.03	1.68	1.76
	Female	6	1.6	0.03	1,59	1.7
	All	12	1.67	0.04	1.59	1.76
Body mass (kg)	Male	6	70	2.7	66	74
	Female	6	62	9.7	53	80
	All	12	66	7.9	53	80

SD= Standard deviation. M= Metre. Kg= kilogram.

Each subject was given an information sheet and a signed consent form was obtained from participants who agreed to take part in this study. Ethical approval for this study was granted from the College of Health and Social Care Research Ethics Panel (Appendix II).

Data were collected by testing each subject on several typical athletic tasks, all of which are described below (see section 4.4.5). All subjects were asked to refrain from exercise one day prior to the testing day to avoid any muscular discomfort or tension which might confound the results (Munro & Herrington, 2011).

4.4.2 System calibration

In motion analysis, the accuracy of the calibration process plays an important role in determining the accuracy of the data collected (Richards, 2008). Therefore, the calibration process was adhered to strictly.

4.4.2.1 3D system calibration

Calibration is necessary for the system to collect kinetic and kinematic data. Therefore, calibration was done according to the manufacturer's guidelines. Two pieces of equipment were used to complete this process. The first one is known as a reference object, which is an L-shaped metal frame (figure 4.1A) with four markers attached to it. This frame is placed on the corner and parallel to the Y and X axes of the force platform. The distances between the markers and the origin of force platform coordinate system are predefined and calculated automatically and linked to the software (Winter, 2009). The frame (reference object) is used to define the origin of the laboratory co-ordinate system, together with X, Y and Z axes (medial/lateral, anterior/posterior and vertical, respectively). The second piece of equipment is a T-shaped wand (figure 4.1B) with two markers on it. The examiner randomly moves this wand around the testing place while the L-shaped frame is still on the force platform to determine the position and orientation of the 15 cameras relative to the coordinate system (Payton, 2008). The calibration process was complete within one minute. When the process was complete, the residual results for the cameras and the standard deviation of the T-shaped wand length must be below 1 mm. If it was more than 1 mm, the calibration was repeated until a correct result was gained.

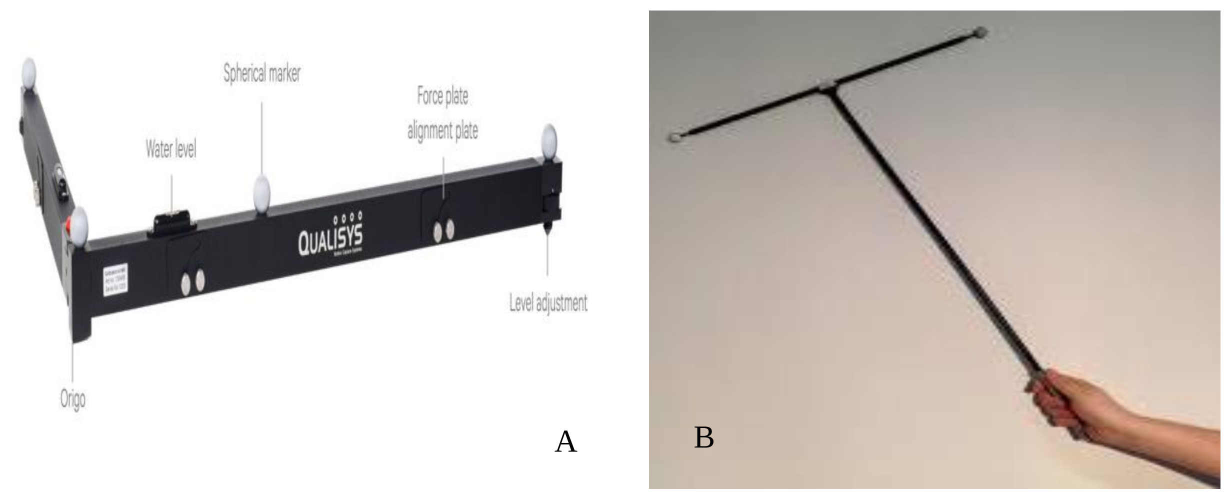


Figure 4.1: A: L-shaped calibration frame (reference object), B: T-shaped calibration wand

4.4.2.2 2D system calibration

A 2D system calibration is necessary because to ensure accurate data by making the size of the object known in the calibrated area at a known distance (Payton, 2008). Therefore, a vertical and horizontal calibration frame (120x120cm) was placed in front of the subject (just between the subject and the force platform). Each subject was asked to hold the frame while recording for three seconds. This video was used as a reference for the distance of the calibrated area when analysing 2D trials using Quintic software (figure 4.2) (Brewin & Kerwin, 2003).

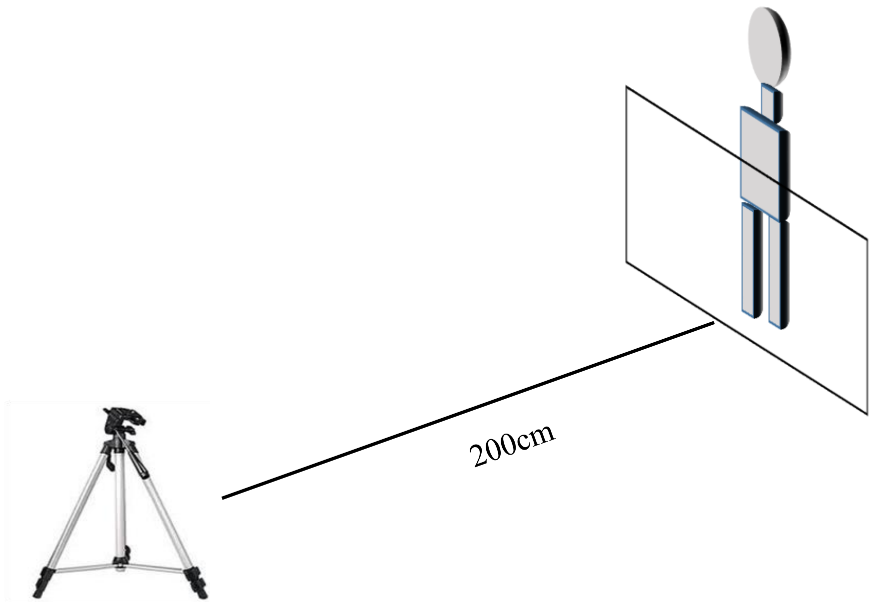


Figure 4.2: 2D system calibration technique

4.4.3 Markers placement

Using hypo-allergic double-sided tape, reflective markers were attached to the subjects on these bony land marks: ASIS, posterior superior iliac spines (PSIS), iliac crest, greater trochanter, medial and lateral femoral condyles, medial and lateral malleoli, posterior calcanei, the head of the 1st, 2nd and 5th metatarsals, antero-lateral aspect of the thigh and shank (four semi-rigid plates, each one consists of 4 reflective markers secured with elastic bands). These markers were used to define the anatomical reference frame and centres of joint rotation. To determine the 3D orientation and position of each marker, at least two cameras should identify the marker at the same time during capture (Cappozzo, Della Croce, Leardini, & Chiari, 2005). Moreover, three non-collinear markers should be seen from each segment to determine its location. During the movement trials, a CAST model was used (figure 4.3). This model was first created by Cappozzo et al. (1995) and suggested to be superior to other systems, such as the modified Helen Hayes markers system, as the former one enhances anatomical relevance and attaches markers to the centres of segments rather than close to joints, which minimizes skin-movement artefacts. Despite skin-movement artefacts, the position and orientation of the joint should not change as the relationship between mounted segment markers and the joint in static calibration initialization can still be determined mathematically. This information should also be able to identify the original position of the joint during dynamic trials after removing the mounted joint markers. Such a method to estimate tibial rotation during walking was suggested by Manal et al. (2000). In addition, CAST is considered to be a gold-standard system as it allows the movement of each segment to be tracked independently by allowing six degrees of freedom (DOF) (3 rotational and 3 translational), compared to only three DOF for the knee and 2 DOF for the ankle by the Helen Hayes markers system. CAST also employs large quantities of markers with small distances in-between, which helps to avoid the propagation of errors that result from the segment measurement of the Helen Hayes marker system, due to less accurate distal segment movement (Cereatti, Camomilla, Vannozzi, & Cappozzo, 2007).

For 2D data collection, a minimum of three markers is needed to measure a joint angle. Therefore, in addition to an ASIS marker, two markers were also attached to the midpoint of the knee joint (midway between the lateral and medial femoral epicondyles) and the middle of the ankle mortise using double-sided adhesive tape. FPPA was defined as the angle between ASIS, the midpoint of the knee joint and the middle of the ankle mortise. The HADD angle was determined by the angle between the two ASIS and the midpoint of the knee joint.

4.4.4 Digital video data collection for knee and hip biomechanics

4.4.4.1 Three-dimensional motion capture

To collect kinematic and kinetic variables for the lower extremities, 15 infrared cameras (Qualisys, Gothenburg, Sweden) sampling in a 100 Hz motion-analysis system, with four force platforms (AMTIBP400600, USA) fixed to the ground of the landing area (sampling at 1000 Hz) were used. These cameras emit light that reflects back to them from the markers and then defines the 2D position of each marker. The system then defines the 3D position by calculating the 2D position relative to the cameras (Kaufman & Sutherland, 2006).

The size of the capture volume is crucial as it may affect the resolution of the system and, in turn, affect the accuracy of the data positions collected. Therefore, the camera configuration should minimize the blind space around the capture volume in the camera's field of view (Richards, 2008;

Pantano et al., 2005). The variables of interest in the current study were collected in the stance phase of multidirectional SLL. Hence, a linear camera configuration around force platforms (figure 4.4) was employed, which covers all the examined movement, as this configuration gives a larger data-collection volume because not all cameras should see the reference frame (Richards, 2008).

Figure 4.3: Calibration Anatomical Systems Technique (CAST)

4.4.4.2 Two-dimensional analysis

One digital video camera (Casio EX-F1, Japan) sampling at 30 Hz was used to videotape subjects when performing test tasks (Table 4.2), it was positioned on a tripod at a horizontal distance of 200 cm, a height of 80 cm in front of the force platform (Willson & Davis, 2008), perpendicular to the frontal plane of motion with an in-built spirit level to keep the perpendicular position, consequently eliminating any unexpected sources of error and maximizing the methodological reliability (Pownall, Moran, & Stewart, 2008). In order to standardize the camera position between participants, the zoom lens of the camera was set at a standard 10x optical zoom in all trials (Munro et al., 2012). This camera was used to collect 2D FPPA and 2D HADD angles (figure 4.4). In the literature, different distances between camera and subjects are used (2.4 m, 3m, 2.5, 10m by Schurr, Marshall, Resch, & Saliba, 2017; Munro et al., 2012, Ghulam, Herrington, Comfort, & Jones, 2015). However, there was no clear criteria when choosing these distances. Payton (2008) reported that the camera should be placed as far as possible from the participants in order to reduce movement outside the plane of performance error (perspective error). Considering the laboratory space, 2 metres was chosen because different distances were tried initially and 2 metres was considered the most suitable to avoid reflections of the 2D cameras in the 3D cameras and adequate for the camera's zoom lens (12x) to keep a balance between perspective error and the quality of the images.

Regarding sampling, some authors suggest 50–100 Hz is suitable for tasks such as running. However, increasing the frame rate may improve the quality of measurements. The camera used in this study can be sampled at 300 Hz. However, it needs extra lights when use this sampling rate, which was not realistic because the limited space of the laboratory would bring the positions of the extra lights within the view of the 3D cameras and reflect in them. Therefore, 30 Hz was the best choice in this case to minimise noise in the 3D data. However, increasing the frame rate might improve the quality of measurements.

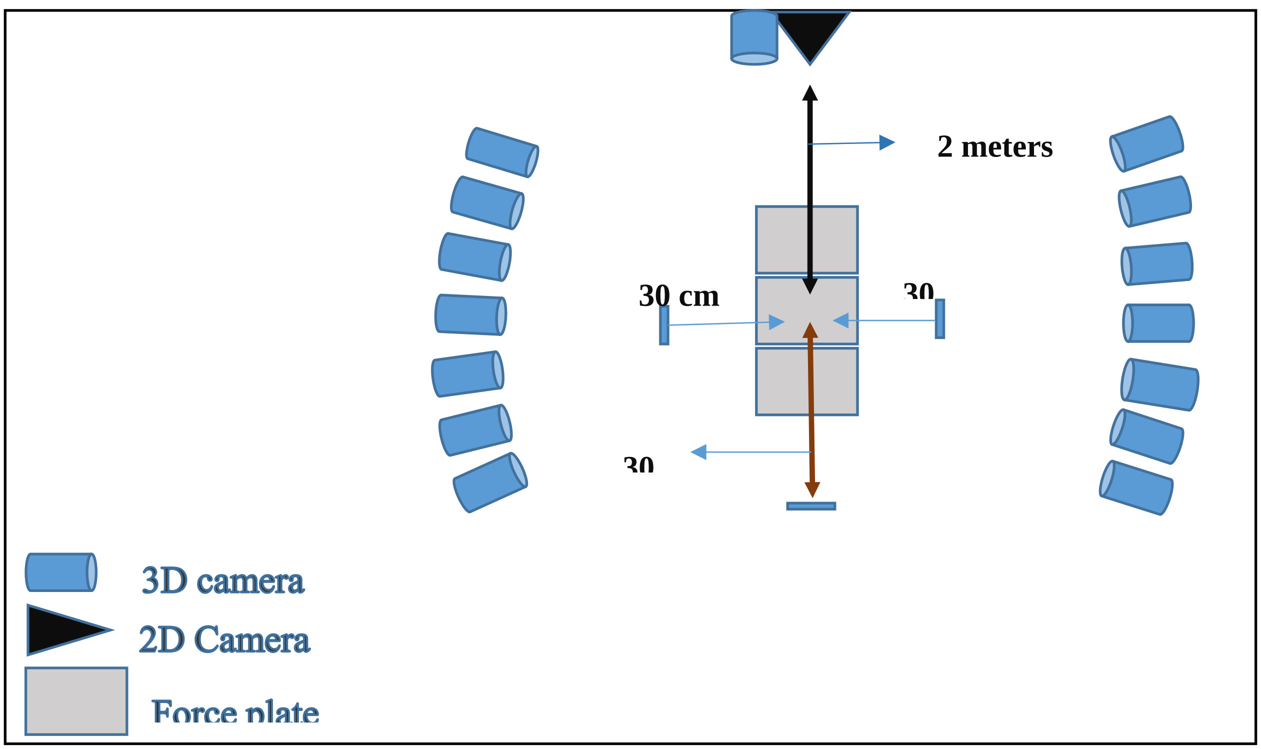


Figure 4.4: Plan of the procedure set-up and cameras configuration

4.4.5 Study procedure

For each subject, 2D knee and hip kinematics and 3D knee and hip kinematics and kinetic and VGRF were collected for both legs while the subjects performed these tests: FSLL, FSLLP, LSLL, LSLLP, MSLLL, and MSLLP. These tests are described in detail below (Table 4.2).

On arrival of a subject at the laboratory, personal data and a past medical history were collected to confirm participants met the inclusion criteria. All subjects were asked if they had read the information sheet and any questions regarding it were answered. A signed consent form was obtained from each participant. A questionnaire was completed for each participant regarding their activity level and health status (Appendix III).

Each subject's body weight and height were measured using an electronic floor weighing scale (Marsden M-420) and a height-measure rod (Seca, UK). Subjects were then asked to change into shorts and remove their shoes and socks. Then, each subject was asked to wear standard shoes (New Balance, UK) which were available to fit all sizes. This was offered because footwear is found to influence lower-limb biomechanics, particularly anterior tibial translation, utilized coefficient of friction (Hong, Jeong, Lee, Yoon, & Shin, 2013), knee-valgus angle and knee-valgus moment (Hong et al., 2014). Consequently, shoe standardization would have the same influence on all subjects. However, it may influence some subjects' performance because it might not allow the proper placement and anatomical alignment of each subject's feet. Also, it might not suit all participants, depending on shoe weight, cushioning, heel thickness and type of arch. These factors may influence shock, impact and stability, which in turn might result in performance changes (Logan, Hunter, Hopkins, J. Feland, & Parcell, 2010; Knapik, Trone, Tchandja, & Jones, 2014).

Subjects then completed a warm-up protocol adapted from Ortiz et al. (2011) (5 toe raises, 5 half squats and 5 continuous vertical jumps) to reduce the risk of physical discomfort and avoid injuries that might occur during the tests (Woods, Bishop, & Jones, 2007). After this, reflective markers were placed on the subjects (see section 4.4.3).

Before starting any of the tests, the researcher gave verbal instructions and demonstrated all test tasks; then, subjects had to perform sufficient practice trials for each of the tests to become familiar with them (Phillips & Van Deursen, 2008). After finishing the practice trials, static standing trials were conducted. Using 3D markers (see section 4.4.3), each subject was captured in a static position while standing over the force plate with equal distribution of the weight on the lower limbs, and it was ensured that the upper limbs were away from the markers to avoid covering them. In this trial, all reflective markers should be visible to the cameras. Qualisys software was used to track anatomical markers prior to extraction for post-processing software. Moreover, from these static trials, a kinematic model was generated by defining seven skeletal segments (pelvis, 2 thighs, 2 shanks and 2 feet) (McLean et al., 2004a).

After the static trials, the anatomical markers (iliac crest, greater trochanter, medial and lateral femoral condyles, medial and lateral malleoli) were detached while retaining the tracking markers [ASIS, PSIS, posterior calcanei, the head of the 1st, 2nd and 5th metatarsals, antero-lateral aspect of the thigh and shank (Four semi-rigid plates, each one consisting of 4 reflective markers tightened with elastic bands)]. The markers on the midpoint of the knee joint and the middle of the ankle mortise, which would be used for 2D measurements, were retained as well.

Then, each subject was captured while performing testing tasks, as described below (Table 4.2). The tasks were performed within and without the 30 cm platform. The rationale for this is that clinicians use both. Moreover, landing in sport occurs from different heights. Therefore, including both may match the reality of different sports and cover their demands. This platform height is standardised in the literature and has been used in many studies, and it may approximate to the average height that people can jump (Orishimo et al., 2009; Ortiz et al., 2011; Alenezi et al., 2014; Jones et al., 2014; Orishimo et al., 2014; Dos Reis et al., 2015; Marshal et al., 2015). The distance between the starting point and the middle of the force platform was standardised at 30 cm for all subjects, because it would be safe for them to perform tests without any risk of injury or fatigue and that distance would ensure all subjects could land on the force platform. Such standardisation helps to increase the internal validity of the experiment, though it may affect the external validity because of differences in the ability to perform the tests (from different distances) from one subject to another.

Tests were conducted in a random order (Philip & Van Deursen, 2008) by asking each subject to choose a folded piece of paper with the name of a test. This helped to minimise bias. Subjects had to achieve three successful trials for each test, from a maximum of five (Gustavsson et al., 2006). The average of these three trials was taken as per the findings of Ortiz, Olson, Libby, Kwon, and Trudelle-Jackson (2007), who examined the number of trials needed to reach acceptable reliability when measuring the biomechanics of a single-leg task, they found that three trials gave good reliability.

A trial was considered successful if the contact phase of the task occurred on the force plate in the field of view of all cameras. Unsuccessful trials were counted and noted but not processed. A data-collection sheet for this purpose was prepared for each participant to regulate the process of trials' data collection and record which trials would be accepted for analysis and which would not when the researcher watched the videos. A tick (√) was drawn in front of accepted trials and a cross (×) in front of unaccepted ones. A 30-second rest period was allowed between each trial for each test (Kea et al., 2001) and 2–5 minutes in-between tests (Corriveau, Hébert, Prince, & Raîche, 2000). Each subject was examined twice on the same day, with an hour between, and another examination a week later.

Table 4.2: Test procedure

Test	Instructions	Notes
FSLL	Participants were asked to stand on both legs at the start point. Then to jump forward and land on the right leg in the middle of the force plate, keeping their eyes open and focused forward, balance as fast as possible, keep still as much as possible for 5 seconds and then relax. Their arms were free to move depending on participants' comfort. No instructions were given about the landing technique to avoid a coaching effect. The same procedure was repeated for the left leg (Fig. 4.5A).	The start point was shown by tape placed on the floor, in front and 30 cm away from the centre of the force platform.
LSLL	The same procedure as FSLL but subjects were asked to jump laterally from the start point and land on the right leg. The same procedure was repeated to land on the left leg but the force platform and starting point were to the left of the subject (Fig. 4.5 C).	The starting point for this test was show by tape placed on the floor, beside and 30cm away from the centre of the force plate.
MSLL	The same procedure as LSLL but the force plate was on the left of the subjects, who jumped towards the force plate and landed on their right leg. The same procedure was repeated as a mirror image for the left leg (Fig. 4.5 E).	The starting point for this test was shown by tape placed on the floor, beside and 30cm away from the centre of the force plate.
FSLLP	The same procedure as FSLL but from a platform (Fig. 4.5 B).	Height of the platform is 30 cm.
LSLLP	The same procedure as LSLL but from a platform (Fig. 4.5 D).	Height of the platform is 30 cm.
MSLLP	The same procedure as MSLL but from a platform (Fig. 4.5 F).	Height of the platform is 30 cm.

FSLL= forward single-leg landing, LSLL = lateral single-leg landing, MSLL = medial single-leg landing, FSLLP = forward single-leg landing off a platform, LSLLP = lateral single-leg landing off a platform, MSLLP = medial single-leg landing off a platform.

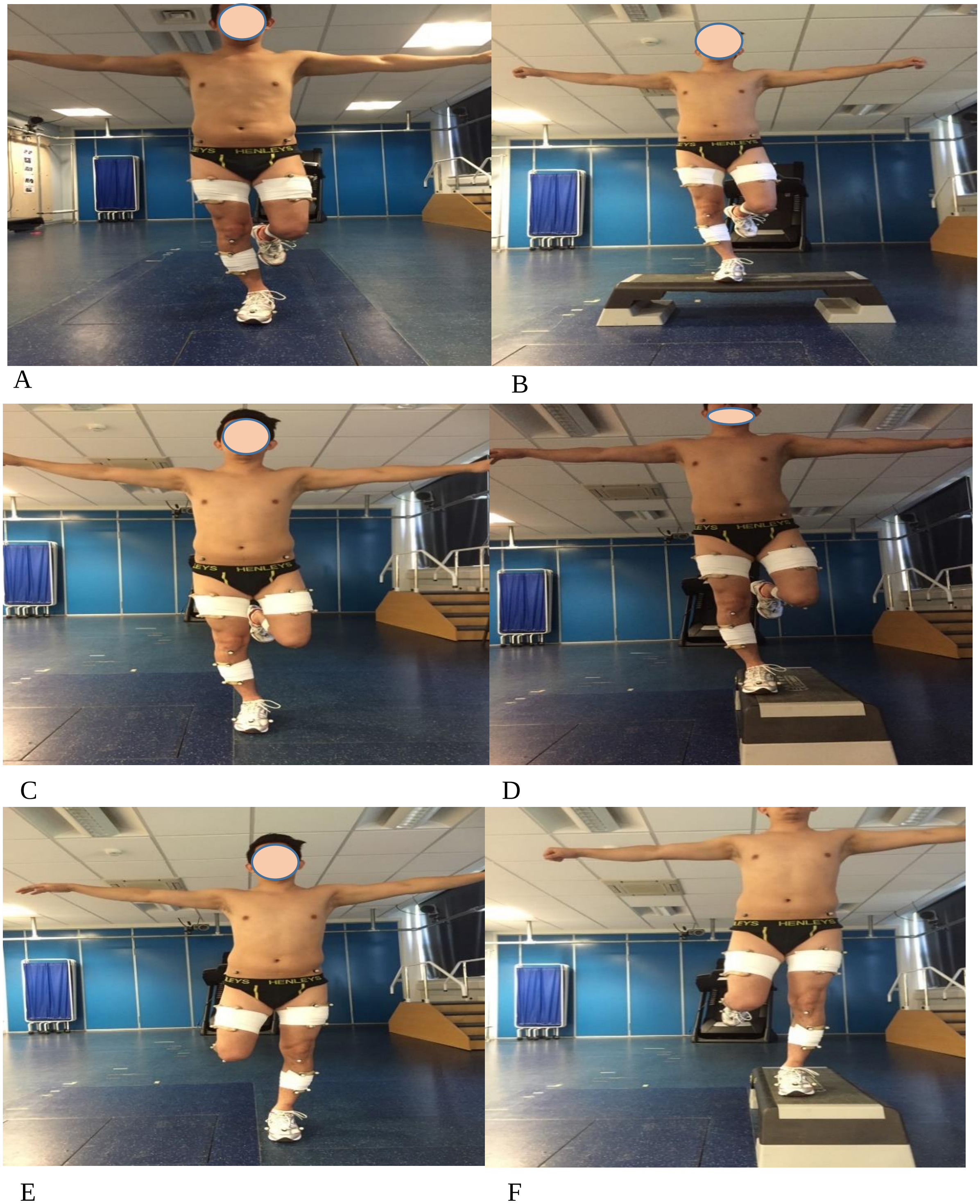


Figure 4.5: Test tasks procedure

4.4.6 Data processing

4.4.6.1 3D data processing

To calculate kinematic and kinetic data, each successful trial was processed using Qualisys Track Manager Software (Version 2.8, Beta Build 835). The markers were labelled and then exported as a C3D file to Visual 3D software (Version 4.21, C-Motion Inc., Rockville, MD, USA). A Butterworth 4th order bi-directional low-pass filter was used for filtration of motion and force measurements with cut-off frequencies of 12 Hz (for motion data) and 25 Hz (for force data). Such filtration is commonly used in motion-analysis research (Munro & Herrington, 2014; Yu, Gabriel, Nobel, & An, 1999). Moreover, Yu et al. (1999) estimated such cut-off frequencies for a Butterworth low-pass digital filter, which became the basis of the cut-off frequencies selected in the current study. The goal of data filtration is to reduce random noise by smoothing the data with no effect on the signal. This is true when using a Butterworth filter. However, there is limited information in the literature that enables researchers to choose the best filtration, thus a pilot study was conducted on the data of four subjects and the aforementioned filtration showed the best data signals. The segments of the lower extremities were modelled as conical frustra, which means that the internal parameters are estimated from anthropometric data (Dempester et al., 1959 as cited in Alenezi et al., 2014).

An X-Y-Z Euler rotation sequence (Table 4.3, Fig. 4.6) was used to calculate joint angles, while 3D inverse dynamics was used to calculate joint kinetic data. All joint moments were normalized to body weight and shown as external moments.

Table 4.3: X-Y-Z Euler rotation sequence

Rotation sequence	**Movement represented**
X	Flexion-extension
Y	Abduction-adduction/valgus-varus
Z	Internal-external rotation.

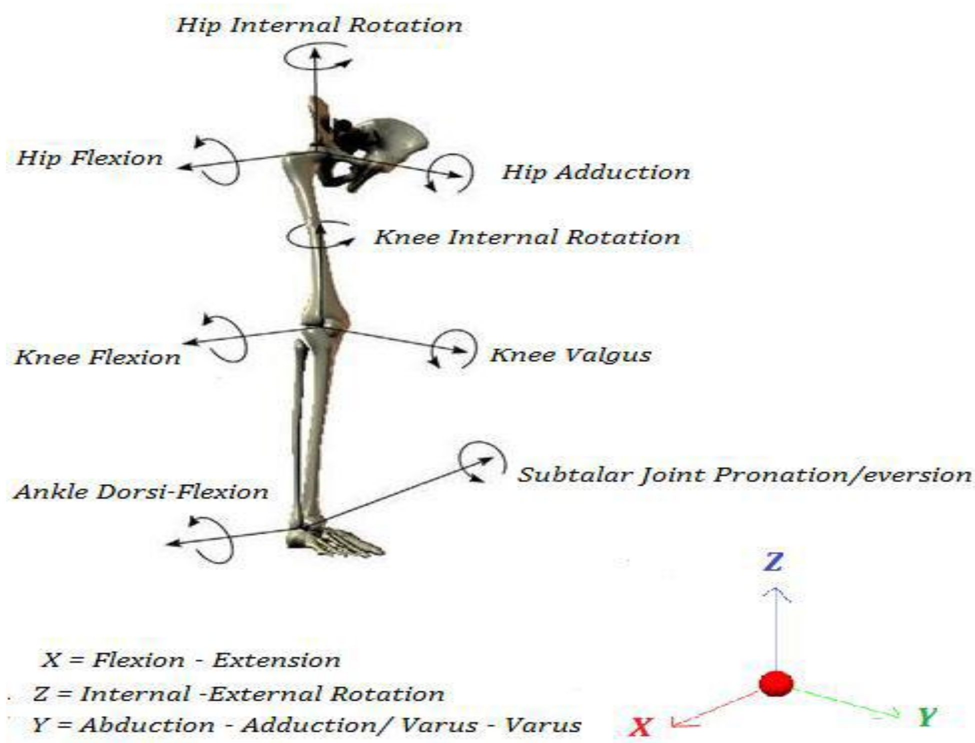


Figure 4.6: Lower-extremity segments and joint rotation denotation

Adapted from Mclean et al. (2005b)

During the dynamic trials, six DOF movements were defined for each segment using the CAST model. A static trial was processed with all markers (anatomical and tracking) (see section 4.4.3) using Qualisys software before extraction to post-processing software (Visual 3D). The positions of the anatomical markers act as reference positions to identify segment movement, by tracking markers during dynamic trials.

The model used in the current study consisted of seven rigid segments attached to the joint (figure 4.7). The position of each segment was described by the six variables that each segment was considered to contain. Three of these variables describe the origin, and the others described the rotation in 3D space. Precisely, three variables describe segment translation within three perpendicular axes (vertical, medial-lateral and anterior-posterior), and three variables describe rotation about each axis of the segment (frontal, sagittal and transverse).

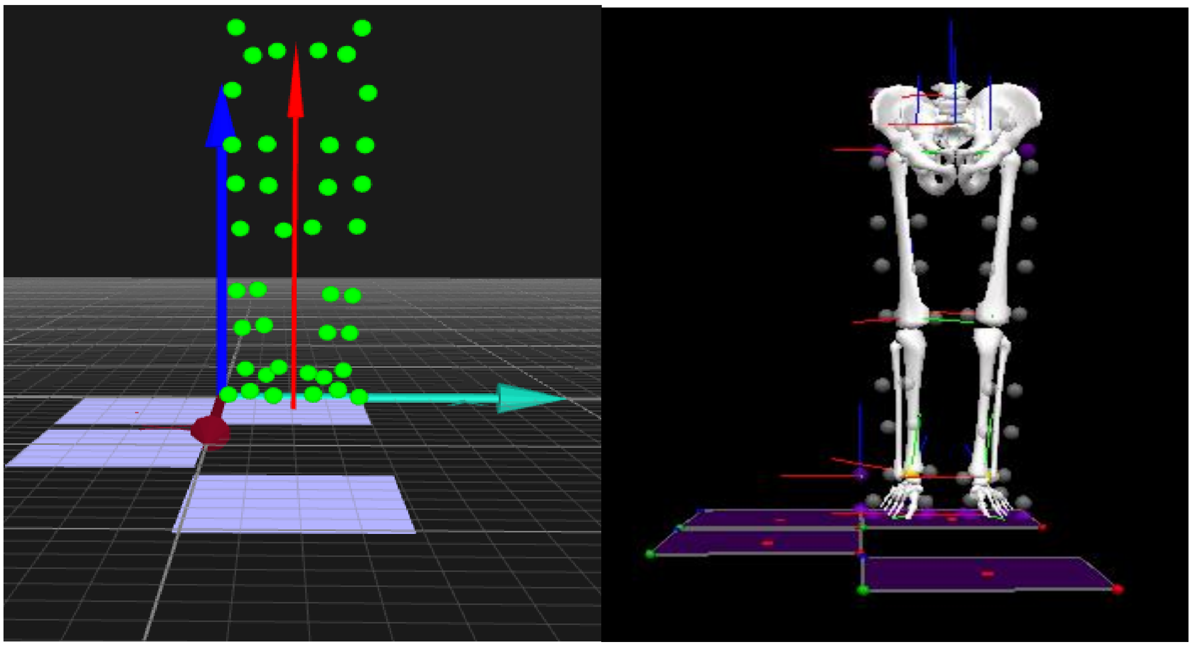


Figure 4.7: QTM™ static models (left), and Visual 3D™ bone model (right)

Each participant's body mass and height (in kilograms and metres, respectively) were entered into Visual 3D software to be used in kinetic measurement calculations. To determine the proximal and distal joint/radius and tracking markers, pelvis, thigh, shank and foot segments were modelled. However, the centre of the hip joint is automatically calculated using ASIS and PSIS markers and by using the regression equation from Bell, Brand, & Pedersen's (1989) study, which found such methods can predict the true position of the centre of the hip joint with about 95% certainty.

An event was then created from initial contact to 15° ascending following the maximum of knee flexion for each leg in each task. The rationale for this was to ensure that maximum knee flexion was included in the event. Figure 4.8 illustrates an example of event creation. The variables of interest were then exported from Visual 3D into Excel, to be used later in final analysis.

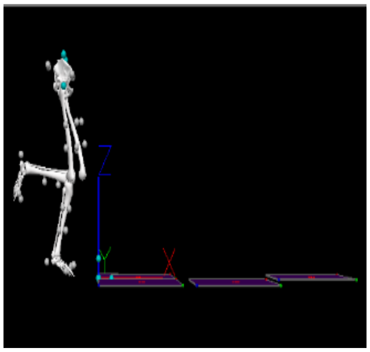
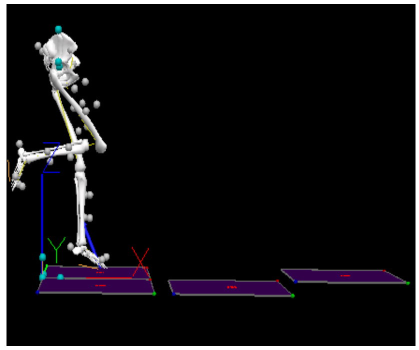
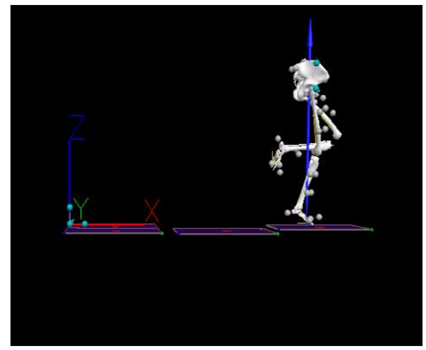
Starting position	Event	
	start	End
		

Figure 4.8: Event creation during the task

4.4.6.2 2D data processing

The data collected during the multidirectional SLLs for each participant were transferred from the camera to a computer. The 2D kinematic data were analysed using Quintic Biomechanics Software (v21, Quintic, Sutton Coldfield, UK). The video captured for calibration was uploaded to the software. Then, the horizontal and vertical lengths of the calibration frame were defined using a designation tool. To determine FPPA, each SLL trial was reviewed in very slow motion, frame by frame, until Peak FPPA was considered to have been observed. This was considered true through a process of two steps. The first step was to stop the video one frame before the point when the subject started to extend and transit from knee flexion after landing (Mizner et al., 2012). The second step was to review the video from the stop point in the first step and go back until initial contact with the ground, and then angles were measured in the frame where the marker on the midpoint of the knee joint was nearest to the opposite leg. These two steps ensured that maximum FPPA was calculated. At this point, the zoom tool was used to determine the centres of placement markers (ASIS, midpoint of the knee joint and midpoint of the ankle mortise). Using the shapes tool in Quintic Biomechanics software, a straight-line passed through the centre of the reflective markers on the middle of the knee joint and the ankle mortise. Using the angle tool of the software, a line was drawn from the centre of ASIS to the centre of the marker placed on the midpoint of the knee joint. Another line was drawn from the latter marker to the marker placed on the midpoint of the ankle mortise. The angle between these lines was defined as FPPA. Figure 4.9 is an illustration of this process. A negative value represents knee valgus, which means the marker on the mid-joint of the knee moves towards the midline of the body, while a positive value represents knee varus, which means the marker on the mid-joint of the knee moves outside the midline of the body. To calculate the right HADD angle, a line was drawn from left ASIS to right ASIS, and another line from right ASIS to the marker on the midpoint of the right knee joint. To calculate the left HADD angle, a line was drawn from right ASIS to left ASIS, and another line from left ASIS to the marker on the midpoint of the left knee joint. Figure 4.10 is an illustration of this process. A positive value means HADD and a negative value means hip abduction. All 2D trials were of the same trials that were accepted for 3D analysis, but captured by a 2D digital camera.

Figure 4.9: 2D frontal plane projection angle during FSLL

Figure 4.10: 2D hip adduction angle during FSLL off a platform

4.4.7 Main outcome measures

Based on what has been discussed in sections 2.5 and 2.7, the main outcomes were:

1. 2D FPPA and HADD angle.
2. 3D peak-knee valgus, hip-adduction, knee-flexion and hip-internal-rotation angles
3. Peak-knee valgus, hip-adductor and knee-extensor external moments.
4. Peak VGRF.

4.4.8 Statistical analysis

All statistical analyses were performed using the Statistical Package for the Social Sciences software (version 21, IBM SPSS Statistics).

All variables in all three visits were tested for normality using a Shapiro Wilk's test. The means of peak-joint angles and moments of successful three trials from the first and second visits were used to evaluate within-day reliability, and the averages of three trials from the first and third visits were used to assess between-days reliability. Intraclass correlation coefficient (ICC) was used to assess the consistency or conformity of measurements (Shrout & Fleiss, 1979). ICC was chosen because it is usually used as a reliability coefficient for evaluating items that are considered to be in the same class or category. ICC compares the covariance of scores with total variance (Yaffee, 1998). It is also used to take into account systematic bias and random error. The nature of ICC means it can be used when a retest is compared with a test. A suitable form of the ICC was chosen as per the guidelines from Shrout and Fleiss (1979). Therefore, model 3.3 was utilised. The first number indicates the use of a two-way mixed model, which means the rater who performed all measurements was the same. Thus, the findings cannot be generalised to other examiners. While the second number indicates the number of averaged measurements (3 trials), which means the result is not applicable to a single measure (Portney & Watkins, 2009). Generally, an ICC of 0.81 or more is considered an indication of excellent reliability, while scores between 0.61 and 0.8 indicates good agreement; a value between 0.41 and 0.60 indicates moderate reliability. A value less than 0.40 is an indication of a less than satisfactory level of agreement (Landis & Koch, 1977).

Although ICC is commonly used in biomedical research to evaluate reliability, on its own it may not provide a comprehensive assessment of the level of reliability and should be combined with a confidence interval (CI). Therefore, test-retest reliability for all measurements performed was associated with 95% CI. Since ICC does not calculate the amount of disagreement between measurements, standard error of measurement (SEM), which is defined as the variance between results, should be calculated. Calculating SEM can help in determining a real change in outcomes, rather than measurement error. A high ICC with a relatively small SEM is a sign of good reliability. Consequently, in addition to ICC with 95% CI, SEM was also calculated using the formula of Denegard and Ball (1993): SEM = (SD (pooled) $\sqrt{1 - \mathrm{ICC}}$). SEM is presented in the same units as the variables tested (degrees for joint kinematics, Newton-metre/kilograms for kinetics) (Blankevoort et al., 2013). % SEM was calculated as [(SEM ÷ actual value) * 100].

4.5 Results

4.5.1 Test of normality

The majority of variables in the three visits reported a P value greater than 0.05, indicating normality. Appendix IV illustrates the results of normality tests for all variables of both legs during all tasks.

4.5.2 2D reliability

Descriptive data, mean and standard deviation (SD) for first, second and third visit measurements for 2D variables during all tasks are presented in Table 4.4.

Mostly, within-day reliability reported greater ICC values than between-days reliability. FSLL showed slight superiority over other tasks.

4.5.2.1 Within-day reliability

Within-day reliability was shown to be good to excellent for all 2D variables in all tasks, ranging between ICC = 0.77–0.97, which was generally greater than between-days ICCs. This result suggests consistency between measurements when examined by one rater. Therefore, the null hypothesis was rejected. Table 4.5, illustrates the ICC and CI for all variables in all tasks.

Within-day SEM, as illustrated in Table 4.6, ranged between 0.65° and 1.88°, which, in general, was less than between-days SEM. However, %SEM ranged between 8.8% and 29.9%.

4.5.2.2 Between-days reliability

Between-days ICCs ranged between 0.62 and 0.96, indicating good to excellent reliability for all 2D variables in all tasks (Table 4.5). Consequently, the null hypothesis was rejected.

Between-days SEMs are shown in Table 4.6, they ranged between 0.69° and 2.7°. %SEM ranged between 9.2% and 41.8%.

Table 4.4: Mean and SD for first, second, and third visit measurements for 2D variables during all tasks

Variables	**Mean (SD)**																	
	Right leg																	
	FSLL			**LSLL**			**MSLL**			**FSLLP**			**LSLLP**			**MSLLP**		
	V1	V2	**V3**	**V1**	**V2**	**V3**	**V1**	**V2**	**V3**	**V1**	**V2**	**V3**	**V1**	**V2**	**V3**	**V1**	**V2**	**V3**
FPPA (°)	-9.8 (4.5)	-8.4 (4.1)	-8.1 (3.6)	-7.5 (4.8)	-5.8 (5.2)	-6.7 (4.3)	-13.7 (5.8)	-13.6 (4.8)	-11.7 (4.6)	-9.9 (5.3)	-9.3 (5.9)	-9.8 (4.8)	-7.4 (3.9)	-7.2 (3.9)	-7.4 (3.2)	-16.3 (7.1)	-15.9 (5.8)	-16.6 (5.7
HADD (°)	7.4 (7.3)	8.3 (4.6)	5.8 (7.7)	5.1 (6)	4.4 (7)	4.6 (6.3)	9.2 (5.4)	8.7 (4.9)	9.2 (4.5)	7.4 (8.2)	7.2 (8.6)	8.2 (6.8)	6.3 (4.9)	4.7 (3.9)	6.6 (3.8)	11.1 (5)	10.1 (5.1)	10.1 (7.9)
	Left leg																	
FPPA (°)	-6.7 (4.2)	-6.1 (4.1)	-6.5 (3.5)	-4.2 (1.6)	-4.3 (2.3)	-4.3 (1.5)	-8.4 (5.6)	-8.9 (4.8)	-7.3 (4.9)	-9.1 (4.8)	-8.7 (4.1)	-8.2 (4.1)	-4.6 (2.2)	-4.9 (2.6)	-3.4 (1.6)	-10.5 (5.2)	-11.7 (6)	-9.4 (5.7)
HADD (°)	6.1 (5.3)	5.5 (5.8)	5.1 (5.4)	4.5 (4.6)	3.6 (4.3)	3.3 (5.1)	6.7 (5.9)	6.6 (4.3)	6.9 (4.5)	7.4 (7.3)	6.9 (6.2)	7.5 (5.1)	5.2 (4.5)	3 (4.7)	4.1 (4.8)	7.4 (5.9)	6.4 (6.1)	7.9 (6.4)

SD = standard deviation, FSLL = forward single-leg landing, LSLL = lateral single leg landing, MSLL = medial single-leg landing, FSLLP = forward single-leg landing off a platform, LSLLP = lateral single-leg landing off a platform, MSLLP = medial single-leg landing off a platform, V1 = visit one, V2 = visit two, V3 = visit three, FPPA = frontal plane projection angle, HADD = hip adduction angle * All variables are angles in °.

4.5.3 3D Reliability

Descriptive data means and standard deviations (SD) for first, second and third visit measurements for 3D kinematic and kinetic variables during all tasks are presented in Tables 4.7 & 4.8, respectively.

In general, within-day reliability reported greater ICC values than between-days reliability. Joint kinematics were more reliable than joint kinetics. Therefore, the null hypothesis was rejected. Comprehensive views of all variables in all tasks suggest that forward SLL and medial SLL off platform are generally more reliable than other tasks.

Table 4.5: Within-day and between-days ICC and 95% CI for 2D variables during all tasks

ICC (95% CI)				
Variables	**Right leg**		**Left leg**	
	Within day	**Between days**	**Within day**	**Between days**
FSLL				
FPPA	0.95 (0.85–0.98)	0.90 (0.66–0.97)	0.94 (0.82–0.98)	0.86 (0.53–0.96)
HADD	0.94 (0.80–0.98)	0.92 (0.74–0.97)	0.91 (0.70–0.97)	0.89 (0.63–0.97)
LSLL				
FPPA	0.90 (0.65–0.97)	0.87 (0.57–0.96)	0.77 (0.23–0.93)	0.70 (0.40–0.98)
HADD	0.93 (0.77–0.98)	0.88 (0.60–0.96)	0.92 (0.73–0.97)	0.88 (0.59–0.96)
MSLL				
FPPA	0.94 (0.81–0.98)	0.91 (0.7–0.97)	0.90 (0.66–0.97)	0.84 (0.44–0.95)
HADD	0.96 (0.89–0.99)	0.93 (0.78–0.98)	0.91 (0.69–0.97)	0.85 (0.48–0.95)
FSLLP				
FPPA	0.97 (0.91–0.99)	0.96 (0.87–0.98)	0.87 (0.55–0.96)	0.82 (0.39–0.95)
HADD	0.95 (0.82–0.98)	0.93 (0.75–0.98)	0.96 (0.86–0.98)	0.93 (0.77–0.98)
LSLLP				
FPPA	0.90 (0.67–0.97)	0.67 (0.36–0.88)	0.91 (0.69–0.97)	0.62 (0.29–0.86)
HADD	0.94 (0.81–0.98)	0.85 (0.49–0.95)	0.88 (0.59–0.96)	0.81 (0.34–0.94)
MSLLP				
FPPA	0.93 (0.75–0.98)	0.92 (0.73–0.97)	0.93 (0.75–0.98)	0.88 (0.59–0.96)
HADD	0.85 (0.48–0.95)	0.81 (0.35–0.94)	0.96 (0.89–0.99)	0.93 (0.75–0.98)

ICC = Intraclass Correlation Coefficient, CI = Confidence Interval, FPPA = Frontal plane projection angle, HADD = Hip adduction, FSLL = Forward single-leg landing, FSLLP = Forward single-leg landing off a platform, LSLL = Lateral single-leg landing, LSLLP = Lateral single-leg landing off a platform, MSLL = Medial single-leg landing, MSLLP = Medial single-leg landing.

Table 4.6: Within-day and between-days means and SEMs for 2D variables during all tasks

Variables	Mean (SEM ° and %)			
	Right leg		Left leg	
	Within day	Between days	Within day	Between days
FSLL				
FPPA (°)	-9.15 (0.93, 10.1%)	-8.98 (1.24, 13.8%)	-6.39 (0.96, 14.3%)	-6.60 (1.36, 20%)
HADD (°)	8.6 (1.15,13.3%)	8.73 (1.28, 14%)	6.31 (1.42, 22%)	6.41 (1.36, 21.2%)
LSLL				
FPPA (°)	-6.71 (1.35, 20.1%)	-7.15 (1.59, 22.2%)	-4.24 (0.88, 20.7%)	-4.27 (0.81, 18.9%)
HADD (°)	4.69 (1.65, 35%)	4.89 (2.05, 42%)	4.04 (1.21, 29.9%)	3.85 (1.61, 41.8%)
MSLL				
FPPA (°)	-13.71 (1.26, 9.1%)	-12.75 (1.51, 11.8%)	-8.67 (1.57, 18%)	-7.84 (2.01, 25.6%)
HADD (°)	9.02 (1.00, 11%)	9.25 (1.27, 13.7%)	6.68 (1.47, 21%)	6.84 (1.94, 28.3%)
FSLLP				
FPPA (°)	-9.64 (0.94, 9.7%)	-9.89 (0.97, 9.8%)	-8.86 (1.54, 17.3%)	-8.62 (1.82, 21%)
HADD (°)	8.66 (1.46, 16.8%)	9.09 (1.46, 16%)	7.72 (1.15, 14.8%)	7.77 (1.48, 19%)
LSLLP				
FPPA (°)	-7.35 (0.65, 8.8%)	-7.44 (0.69, 9.2%)	-5.04 (1.03, 20.4%)	-4.31 (1.0, 23.2%)
HADD (°)	5.5 (1.04, 18.9%)	6.49 (1.63, 25.1%)	6.09 (1.54, 25%)	4.63 (1.94, 41.9%)
MSLLP				
FPPA (°)	-16.11 (1.64, 10.1%)	-16.5 (1.74, 10.5%)	-11.08 (1.42, 11.1%)	-9.94 (1.81, 18.2%)
HADD (°)	10.59 (1.88, 17.7%)	10.58 (2.7, 25.5%)	6.86 (1.16, 16.9%)	7.6 (1.56, 20.5%)

SEM = Standard error of measurement, Within-day mean = the mean of visit 1 mean and visit 2 mean, Between-days mean = the mean of visit 1 mean and visit 3 mean, FPPA = Frontal plane projection angle, HADD = Hip adduction, FSLL = Forward single-leg landing, FSLLP = Forward single-leg landing off a platform, LSLL = Lateral single-leg landing, LSLLP = Lateral single-leg landing off a platform, MSLL = Medial single-leg landing, MSLLP = Medial single-leg landing off a platform

Table 4.7: Mean and SD for first, second and third visit measurements for 3D kinematic variables during all tasks

Variables	Mean (SD)																	
	Right leg																	
	FSLL			**LSLL**			**MSLL**			**FSLLP**			**LSLLP**			**MSLLP**		
	V1	**V2**	**V3**	**V1**	**V2**	**V3**	**V1**	**V2**	**V3**	**V1**	**V2**	**V3**	**V1**	**V2**	**V3**	**V1**	**V2**	**V3**
K/VAL (°)	-0.9 (3.6)	-1.3 (3.8)	-1.9 (4.5)	-3.6 (3.1)	-4.7 (4.1)	-4.7 (3.3)	-2.1 (5.7)	-1.5 (5.8)	-1.9 (5.7)	-1.3 (3.7)	-0.9 (4.7)	-1.5 (4.4)	-2.5 (3.5)	-3.3 (4.6)	-3.2 (4.1)	-2.9 (5.6)	-4.5 (6.6)	-4.3 (4.8)
K/FLX (°)	60.6 (7.2)	60 8 (6.9)	61.5 (8.8)	59.1 (8.6)	59.6 (5.7)	58.3 (7.8)	59.4 (7.5)	60.3 (7)	57.2 (9)	65.2 (7.7)	65.6 (8.1)	67.1 (7.2)	62.7 (5.8)	64.8 (7.7)	64.9 (11)	65.2 (6.5)	64.2 (8.4)	69.8 (14.8)
HADD (°)	8.3 (4.7)	9 (5.6)	8,9 (6.7)	5.7 (6.4)	6.3 (6.2)	5.9 (6.5)	8.7 (5.2)	9.2 (6)	8.5 (5.5)	10.1 (6.4)	8.2 (7.1)	10 (6.3)	7.5 (6.6)	6.2 (5.8)	9.3 (6)	9.9 (5.7)	9.5 (6.3)	11 (4.9)
H/INT (°)	9.7 (6.1)	9.6 (5.8)	9.4 (6.6)	11.4 (6.6)	12.7 (8.2)	10.3 (6.6)	10.6 (8.2)	10.8 (8.6)	10.3 (8.4)	9.9 (7.9)	11.6 (8.4)	9.5 (8.8)	12 (10.3)	10.7 (10.3)	12 (9.6)	11.9 (6.4)	12.1 (7)	10.5 (5.7
	Left leg																	
K/VAL (°)	-0.9 (3)	-1.8 (3)	-1.5 (2.4)	-2.9 (2.8)	-3.8 (3)	-3.9 (3.1)	-1.8 (2.7)	-1.7 (3.8)	-2.4 (2.9)	-2 (2.8)	-3.2 (3.4)	-2.9 (3.9)	-2.2 (1.8)	-4.4 (2.9)	-3.4 (2.9)	-3.2 (3.4)	-4.1 (3.9)	-4.2 (3.5)
K/FLX (°)	57.9 (7.3)	57.6 (8.8)	56.7 (7.2)	55.8 (9.1)	55.6 (9.2)	56.2 (9.1)	56.3 (7.9)	56.1 (8,2)	56.2 (11.7)	58.9 (12.8)	59.1 (10.6)	58.4 (8.7)	59.5 (6.2)	58.9 (9.)	60.5 (10.1)	61.4 (10.4)	62.4 (10)	62.5 (9.6)
HADD (°)	6.8 (5.6)	6.4 (5.4)	7.5 (5.6)	5.5 (5.2)	5.8 (5.6)	4.9 (5.1)	7.2 (6.8)	6.9 (5.6)	7 (5)	6.9 (6.8)	7.7 (6.2)	7.2 (6)	6.7 (5.8)	4.2 (5.6)	4.4 (5.4)	7.9 (6.9)	8.4 (7.03)	8.6 (6.4)
H/INT (°)	8.4 (5)	9 (5.5)	7.4 (5.4)	8.8 (6.5)	8.6 (6.8)	9.3 (7.7)	9.1 (4.8)	7.3 (5.8)	7.1 (6.5)	8.1 (6)	7.9 (6.2)	8.8 (6.6)	9.4 (6.3)	7.8 (7.3)	9.2 (7.7)	11.2 (6.1)	9.9 (6.8)	9.9 (6.6)

SD = standard deviation, FSLL = forward single-leg landing, LSLL = lateral single-leg landing, MSLL = medial single-leg landing, FSLLP = forward single-leg landing off a platform, LSLLP = lateral single-leg landing off a platform, MSLLP = medial single-leg landing off a platform, V1 = visit one, V2 = visit two, V3 = visit three. K/VAL = knee valgus, K/FLX= knee flexion, HADD = hip adduction, H/INT = hip internal rotation. * All variables are angles in °.

Table 4.8: Mean and SD for first, second and third visit measurements for 3D kinetics variables and GRF during all tasks

		FSLL			**LSLL**			**MSLL**		**Right leg**	**FSLLP**			**LSLLP**			**MSLLP**	
	V1	V2	**V3**	**V1**	**V2**	**V3**	**V1**	**V2**	**V3**	**V1**	**V2**	**V3**	**V1**	**V2**	**V3**	**V1**	**V2**	**V3**
K/VALM (Nm/Kg)	0.14 (0.17)	0.13 (0.12)	0.1 (0.11)	0,24 (0.11)	0.26 (0,13)	0.35 (0.15)	0.26 (0.2)	0.23 (0.17)	0.23 (0.16)	0.26 (0.13)	0.29 (0.14)	0.21 (0.15)	0.36 (0.15)	0.31 (0.14)	0.25 (0.15)	0.5 (0.3)	0.4 (0.3)	0.5 (0.3)
K/EXM (Nm/Kg)	2.8 (0.4)	2.7 (0.5)	2.7 (0.4)	2.4 (0.49)	2.5 (0.5)	2.5 (0.42)	2.5 (0.36)	2.4 (0.37)	2.6 (0.49)	3 (0.49)	3.01 (0.5)	3.3 (0.6)	2.7 (0.38)	2.8 (0.4)	2.8 (0.41)	3 (0.48)	2.9 (0.62)	3.01 (0.51)
HADDM (Nm/Kg)	-1.5 (0.33)	-1.3 (0.34)	-1.2 (0.35)	-1.4 (0.29)	-1.5 (0.27)	-1.5 (0.36)	-1.32 (0.25)	-1.4 (0.29)	-1.35 (0.18)	-1.68 (0.4)	-1.96 (0.43)	-1.68 (0.51)	-1.7 (0.43)	-1.71 (0.33)	-1.76 (0.38)	-1.69 (0.6)	-1.76 (0.56)	-1.71 (0.57)
H/INTM (Nm/Kg)	-0.85 (0.26)	-0.9 (0.26)	-0.76 (0.18)	-0.76 (0.19)	-0.76 (0.22)	-0.71 (0.25)	-0.71 (0.23)	-0.67 (0.21)	-0.78 (0.35)	-0.97 (0.15)	-1.1 (0.2)	-1.1 (0.21)	-0.99 (0.28)	-1.1 (0.27)	-0.96 (0.28)	-0.96 (0.29)	-1.04 (0.31)	-1.1 (0.33)
GRF (times WB)	2.35 (0.32)	2.32 (0.31)	2.4 (0.29)	2.41 (0.44)	2.44 (0.37)	2.5 (0.46)	2.39 (0.28)	2.44 (0.37)	2.41 (0.32)	3.48 (0.43)	3.37 (0.51)	3.78 (0.38)	3.57 (0.52)	3.53 (0.53)	3.4 (0.75)	3.35 (0.61)	3.51 (0.55)	3.46 (0.68)
										Left leg								
K/VALM (Nm/Kg)	0.03 (0.07)	0.02 (0.08)	.012 (0.09)	0.15 (0.08)	0.14 (0.11)	0.14 (0.09)	0.06 (0.09)	0.05 (0.13)	0.09 (.15)	0.17 (0.14)	0.15 (0.14)	0.21 (0.18)	0.14 (0.13)	0.19 (0.13)	0.18 (0.1)	0.2 (0.13)	0.27 (0.21)	0.19 (0.1)
K/EXM (Nm/Kg)	2.8 (0.5)	2.7 (0.6))	2.7 (0.53)	2.35 (0.4)	2.3 (0.51)	2.31 (0.56)	2.3 (0.59)	2.4 (0.46)	2.3 (0.5)	3 (0.55)	3.2 (0.6)	3.2 (0.54)	2.76 (0.46)	2.9 (0.68)	2.8 (0.69)	2.9 (0.75)	2.8 (0.61)	2.8 (0.65)
HADDM (Nm/Kg)	-1.6 (0.26)	-1.8 (0.28)	-1.7 (0.21)	-1.88 (0.28)	-1.73 (0,21)	-1.8 (0.24)	-1.64 (0.25)	-1.61 (0.26)	-1.69 (0.28)	-1.92 (0.2)	-1.97 (0.23)	-1.9 (0.13)	-2 (0.19)	-2.3 (0.66)	-1.98 (0.09)	-1.96 (0.28)	-2 (0.4)	-2.1 (0.37)
H/INTM (Nm/Kg)	-0.93 (0.23)	-1 (0.34)	-1.1 (0.36)	-0.95 (0.37)	-0.91 (0.24)	-0.9 (0.32)	-0.89 (0.22)	-0.86 (024)	-0.83 (0.17)	-1.1 (0.25)	-1.22 (0.31)	-1.2 (0.29)	-1.15 (0.2)	-1.07 (0.27)	-1.3 (0.4)	-1.15 (0.21)	-1.21 (0.19)	-1.2 (0.29)
GRF (times BW)	2.35 (0.36)	2.38 (0.27)	2.34 (0.3)	2.3 (0.35)	2.37 (0.36)	2.36 (0.36)	2.47 (0.37)	2.53 (0.37)	2.54 (0.35)	3.4 (0.44)	3.6 (0.47)	3.66 (0.38)	3.5 (0.59)	3.6 (0.81)	3.6 (0.54)	3.53 (0.58)	3.45 (0.52)	3.47 (0.46))

SD = standard deviation, FSLL = forward single-leg landing, LSLL = latera single-leg landing, MSLL = medial single-leg landing, FSLLP = forward single-leg landing off a platform, LSLLP = lateral single-leg landing off a platform, MSLLP = medial single-leg landing off a platform, V1 = visit one, V2 = visit two, V3 = visit three. K/VALM = knee valgus moment, K/EXM= knee extensor moment, HADDM = hip adduction moment, H/INTM = hip internal rotation moment, BW = Body weight, Nm/kg = Newton meter per kilogram.

4.5.3.1 Within-day reliability

Within-day reliability for 3D variables in all tasks reported ICCs ranging between 0.61 and 0.98 for most of the variables, which suggests good to excellent within-day agreement when examined by one rater. Moderate within-day agreement, ranging between 0.44 and 0.60, was reported for left-leg HADD moment during FSLL, LSLL and LSLLP, both leg-knee valgus moments during FSLLP, left-knee-extensor moment during LSLLP, and left-leg knee-valgus moment during MSLL.

Less than satisfactory between-days agreement was reported for knee-valgus moment and knee-adduction moment during most of the tasks, particularly those performed off a platform. All 3D variables ICCs are presented in: Table 4.9 for FSLL, Table 4.10 for FSLLP, Table 4.11 for LSLL, Table 4.12 for LSLLP, Table 4.13 for MSLL and Table 4.14 for MSLLP.

Within-day SEMs, as illustrated in Tables 4.15–4.20, ranged between 0.63° and 3.3° for kinematic variables, with left-knee-flexion angle during LSLLP exhibiting the greatest SEM (3.3°). The smallest SEM (0.63°) was reported for the left-leg HADD angle during FSLLP. However, it appears larger when considering %SEM as it ranges between 4% and 90%.

Kinetic variables reported within-day SEM ranging between 0.01 and 0.34 Nm/Kg. The lowest was reported for right-leg HADD moment and left-leg GRF during MSLL, while the highest was reported for GRF during landing laterally on the left leg and for left-knee-extensor moment during LSLLP. %SEM ranged between 0.7% and 80%.

Table 4.9: Interclass Correlations (ICCs) and Confidence Intervals (CIs) for 3D variables during forward single-leg landing

Variable	Right leg		Left leg	
	Within day	Between days	Within day	Between days
	ICC (95% CI)			
Knee-valgus angle	0.96 (0.87–0.98)	0.92 (0.74–0.97)	0.91 (0.70–0.97)	0.89 (0.63–0.96)
Knee-flexion angle	0.85 (0.50–0.95)	0.84 (0.46–0.95)	0.89 (0.64–0.97)	0.82 (0.37–0.94)
Hip-adduction angle	0.92 (0.72–0.97)	0.83 (0.42–0.95)	0.97 (0.90–0.99)	0.95 (0.82–0.98)
Hip-internal rotation angle	0.98 (0.93–0.99)	0.97 (0.91–0.99)	0.94 (0.81–0.98)	0.86 (0.54–0.96)
Knee-valgus moment	0.83 (0.42–0.95)	0.62 (0.29–0.86)	0.61 (0.28–0.85)	0.61 (0.28–0.85)
Knee-extensor moment	0.88 (0.6–0.96)	0.77 (0.51–0.92)	0.91 (0.71–0.97)	0.90 (0.66–0.97)
Hip-adduction moment	0.97 (0.89–0.99)	0.79 (0.3–0.94)	0.60 (0.26–0.85)	0.59 (0.25–0.84)
Hip-internal rotation moment	0.83 (0.41–0.95)	0.78 (0.24–0.93)	0.73 (0.45–0.9)	0.67 (0.36–0.88)
Ground-reaction force	0.96 (0.86–0.98)	0.71 (0.42–0.9)	0.8 (0.57–0.93)	0.72 (0.43–0.9)

Table 4.10: Interclass Correlations (ICCs) and Confidence Intervals (CIs) for 3D variables during forward single-leg landing off a platform

Variable	Right leg		Left leg	
	Within day	Between days	Within day	Between days
	ICC (95% CI)			
Knee-valgus angle	0.94 (0.79–0.98)	0.88 (0.6–.96)	0.83 (0.42–.95)	0.82 (0.6–0.94)
Knee-flexion angle	0.88 (0.59–0.96)	0.79 (0.55–0.93)	0.93 (0.76–0.98)	0.87 (0.57–0.96)
Hip-adduction angle	0.96 (0.88–0.99)	0.92 (0.72–0.97)	0.99 (0.97–0.99)	0.97 (0.89–0.99)
Hip-internal rotation angle	0.98 (0.93–0.99)	0.97 (0.92–0.99)	0.94 (0.79–0.98)	0.93 (0.76–0.98)
Knee-valgus moment	0.58 (0.24–0.84)	0.54 (0.19–0.82)	0.59 (0.25–0.84)	0.39 (0.03–0.74)
Knee-extensor moment	0.92 (0.73–0.97)	0.69 (0.39–0.89)	0.84 (0.47–0.95)	0.78 (0.53–0.92)
Hip-adduction moment	0.88 (0.58–0.96)	0.85 (0.48–0.95)	0.62 (0.29–0.86)	0.35 (0.00–0.71)
Hip-internal rotation moment	0.67 (0.36–0.88)	0.38 (0.02–0.73)	0.82 (0.6–0.94)	0.85 (0.66–0.95)
Ground-reaction force	0.91 (0.69–0.97)	0.56 (0.21–0.83)	0.75 (0.48–0.91)	0.64 (0.32–0.87)

Table 4.11: Interclass Correlations (ICCs) and Confidence Intervals (CIs) for 3D variables during lateral single-leg landing

Variable	Right leg		Left leg	
	Within day	Between-days	Within day	Between-days
ICC (95% CI)				
Knee-valgus angle	0.89 (0.74–0.96)	0.82 (0.6–0.94)	0.9 (0.76–0.97)	0.85 (0.66–0.95)
Knee-flexion angle	0.84 (0.64–0.95)	0.79 (0.55–0.93)	0.95 (0.87–0.98)	0.95 (0.87–0.98)
Hip-adduction angle	0.94 (0.81–0.98)	0.9 (0.68–0.97)	0.93 (0.78–0.98)	0.91 (0.7–0.97)
Hip-internal rotation angle	0.96 (0.86–0.98)	0.96 (0.89–0.99)	0.93 (0.77–0.98)	0.88 (0.59–0.96)
Knee-valgus moment	0.64 (0.32–0.87)	0.45 (0.09–0.77)	0.72 (0.43–0.9)	0.48 (0.12–0.79)
Knee-extensor moment	0.81 (0.58–0.93)	0.66 (0.34–0.84)	0.94 (0.79–0.98)	0.88 (0.72–0.96)
Hip-adduction moment	0.91 (0.78–0.97)	0.86 (0.68–0.95)	0.56 (0.21–0.83)	0.42 (0.06–0.75)
Hip-internal rotation moment	0.79 (0.55–0.93)	0.9 (0.67–0.97)	0.9 (0.76–0.97)	0.93 (0.76–0.98)
Ground-reaction force	0.88 (0.59–0.96)	0.88 (0.61–0.96)	0.92 (0.75–0.98)	0.9 (0.67–0.97)

Table 4.12: Interclass Correlations (ICCs) and Confidence Intervals (CIs) for 3D variables during lateral single-leg landing off a platform

Variable	Right leg		Left leg	
	Within day	Between days	Within day	Between days
ICC (95% CI)				
Knee-valgus angle	0.77 (0.51–0.92)	0.77 (0.22–0.93)	0.63 (0.3–0.86)	0.56 (0.21–0.83)
Knee-flexion angle	0.78 (0.53–0.92)	0.57 (0.23–0.83)	0.82 (0.6–0.94)	0.79 (0.55–0.83)
Hip-adduction angle	0.92 (0.73–0.97)	0.85 (0.66–0.95)	0.93 (0.77–0.98)	0.91 (0.71–0.97)
Hip-internal rotation angle	0.97 (0.91–0.99)	0.9 (0.67–0.97)	0.95 (0.84–0.98)	0.94 (0.82–0.98)
Knee-valgus moment	0.67 (0.36–0.88)	0.47 (0.11–0.78)	0.67 (0.36–0.88)	0.61 (0.28–0.85)
Knee-extensor moment	0.77 (0.21–0.93)	0.72 (0.43–0.9)	0.53 (0.18–0.81)	0.5 (0.15–0.8)
Hip-adduction moment	0.90 (0.66–0.97)	0.89 (0.63–0.97)	0.44 (0.08–0.76)	0.46 (0.10–0.78)
Hip-internal rotation moment	0.70 (0.40–0.89)	0.93 (0.76–0.98)	0.84 (0.45–0.95)	0.55 (0.2–0.82)
Ground-reaction force	0.80 (0.57–0.93)	0.80 (0.57–0.93)	0.80 (0.32–0.94)	0.76 (0.5–0.92)

Table 4.13: Interclass Correlations (ICCs) and Confidence Intervals (CIs) for 3D variables during medial single-leg landing

Variable	Right leg		Left leg	
	Within day	**Between days**	**Within day**	**Between days**
	ICC (95% CI)			
Knee-valgus angle	0.97 (0.89–0.99)	0.83 (0.62–0.94)	0.90 (0.76–0.97)	0.76 (0.5–0.92)
Knee-flexion angle	0.81 (0.58–0.93)	0.57 (0.23–0.83)	0.88 (0.60–0.96)	0.78 (0.53–0.92)
Hip-adduction angle	0.89 (0.64–0.97)	0.83 (0.62–0.94)	0.98 (0.95–0.99)	0.96 (0.90–0.99)
Hip-internal rotation angle	0.98 (0.95–0.99)	0.79 (0.55–0.93)	0.93 (0.83–.98)	0.89 (0.64–0.97)
Knee-valgus moment	0.93(0.78–0.98)	0.83 (0.62–0.94)	0.67 (0.36–0.88)	0.58 (0.24–0.084)
Knee-extensor moment	0.88 (0.72–0.96)	0.58 (0.24–0.84)	0.87 (0.56–0.96)	0.83 (0.62–0.94)
Hip-adduction moment	0.87 (0.70–0.96)	0.80 (0.57–0.93)	0.79 (0.55–0.93)	0.60 (0.26–0.85)
Hip-internal rotation moment	0.63 (0.30–0.86)	0.79 (0.55–0.93)	0.71 (0.42–0.90)	0.70 (0.40–0.89)
Ground reaction force	0.89 (0.63–0.97)	0.89 (0.63–0.97)	0.85 (0.48–0.95)	0.65 (0.33–0.87)

Table 4.14: Interclass Correlations (ICCs) and Confidence Intervals (CIs) for 3D variables during medial single-leg landing off a platform

Variable	Right leg		Left leg	
	Within day	**Between days**	**Within day**	**Between days**
	ICC (95% CI)			
Knee-valgus angle	0.98 (0.93–0.99)	0.94 (0.80–0.98)	0.79 (0.55–0.93)	0.71 (0.42–0.90)
Knee-flexion angle	0.88 (0.72–0.96)	0.87 (0.56–0.96)	0.89 (0.62–0.96)	0.84 (0.64–0.95)
Hip-adduction angle	0.95 (0.83–0.98)	0.89 (0.63–0.96)	0.92 (0.73–0.97)	0.85 (0.66–0.95)
Hip-internal rotation angle	0.98 (0.96–0.99)	0.97 (0.90–0.99)	0.97 (0.91–0.99)	0.92 (0.74–0.97)
Knee-valgus moment	0.84 (0.45–0.95)	0.81 (0.58–0.93)	0.56 (0.21–0.83)	0.34 (0.10–0.66)
Knee-extensor moment	0.91 (0.78–0.97)	0.89 (0.74–0.96)	0.94 (0.85–0.98)	0.92 (0.81–0.97)
Hip-adduction moment	0.89 (0.61–0.96)	0.75 (0.48–0.91)	0.93 (0.83–0.98)	0.73 (0.45–0.90)
Hip-internal rotation moment	0.89 (0.64–0.97)	0.82 (0.60–0.94)	0.92 (0.72–0.97)	0.89 (0.62–0.96)
Ground-reaction force	0.88 (0.59–0.96)	0.89 (0.63–0.97)	0.94 (0.80–0.98)	0.94 (0.81–0.98)

4.5.3.2 Between-days reliability

Between-days ICCs ranged between 0.42 and 0.97, indicating moderate to excellent reliability for all 3D variables in all tasks. Less than satisfactory between-days agreements ranging between 0.34 and 0.39 were reported for left-leg knee-valgus moment during FSLL and MSLL, left-leg HADD moment and right-leg hip-internal rotation moment during FSLL. All 3D variables' ICCs are presented in: Table 4.9 for FSLL, Table 4.10 for FSLLP, Table 4.11 for LSLL, Table 4.12 for LSLLP, Table 4.13 for MSLL and Table 4.14 for MSLLP.

Between-days SEMs for all variables during all tasks are presented in Tables 4.15–4.20. Kinematic variables reported SEMs ranging between 0.86° and 6.6°, with left-knee valgus angle during FSLL being the smallest, while right-knee flexion angle during MSLLP was the greatest. %SEM ranges between 3.3% and 90%.

Kinetic variables reported between-days SEMs ranging between 0.05 and 0.29 Nm/Kg. The lowest was reported for left-knee valgus moment during FSLL while the highest was for left-knee extensor moment during MSLLP. %SEM ranges from 0.4% to 77%.

Table 4.15: Within-day and between-days means, SDs and SEMs for 3D variables during FSLL[a]

FSLL												
Variables	**Right leg**						**Left leg**					
	Within-day			**Between-days**			**Within-day**			**Between-days**		
	mean	**SD**	**SEM° (%)**	**Mean**	**SD**	**SEM° (%)**	**Mean**	**SD**	**SEM° (%)**	**Mean**	**SD**	**SEM° (%)**
Knee -valgus angle	-1.12	3.61	0.72 (64%)	-1.42	3.9	1.1 (77%)	-1.34	2.9	0.87 (64%)	-1.2	2.6	0.86 (71%)
Knee-flexion angle	60.7	6.8	2.6 (4.2%)	61	7.7	3.1 (5.1%)	57.8	7.7	2.6 (4.4%)	57.3	7	2.7 (4.7%)
Hip-adduction angle	8.7	5	1.4 (16%)	8.7	5.5	2.3 (26.4%)	6.6	5.3	0.92 (17.3%)	7.2	5.4	1.2 (16%)
Hip-internal rotation angle	9.7	5.7	0.81 (8.3%)	9.6	6	1 (10.4%)	8.7	5	1.2 (13.7%)	8	5	1.9 (23%)
Knee-valgus moment	0.14	0.14	0.05 (35%)	0.12	0.14	0.09 (75%)	0.03	0.08	0.05 (16%)	0.02	0.08	0.05 (25%)
Knee-extensor moment	2.8	0.5	0.2 (7.1%)	2.7	0.42	0.2 (7.4%)	2.8	0.52	0.16 (5.7%)	2.8	0.5	0.2 (7.1%)
Hip-adduction moment	-1.4	0.3	0.05 (3.5%)	-1.3	0.3	0.14 (10.7%)	-1.7	0.3	0.19 (11.1%)	-1.7	0.2	0.13 (7.6%)
Hip-internal rotation moment	-0.9	0.3	0.12 (13.3%)	-0.8	0.21	0.1 (12.5%)	-0.9	0.3	0.2 (22.2%)	-0.99	0.3	0.2 (20.2%)
Ground-reaction force	2.3	0.3	0.06 (2.6%)	2.4	0.3	0.16 (6.6%)	2.4	0.3	0.13 (5.4%)	2.4	0.3	0.16 (6.6%)

[a] All angles in degrees. All moments in Newton metres per kilogram, Ground reaction force = *body weight, SEM = Standard error of measurement. Within-day mean = the mean of visit 1 mean and visit 2 mean. Within-day SD = the average of visit 1 and visit 2 standard deviations. Between-days mean = the mean of visit 1 mean and visit 3 mean. Between-days SD = the average of visit 1 and visit 2 standard deviations, FSLL = forward single-leg landing

Table 4.16: Within-day and between-days means, SDs and SEMs for 3D variables during FSLLP[a]

FSLLP												
Variable	**Right leg**						**Left leg**					
	Within day			**Between days**			**Within day**			**Between days**		
	mean	**SD**	**SEM° (%)**	**Mean**	**SD**	**SEM° (%)**	**Mean**	**SD**	**SEM° (%)**	**Mean**	**SD**	**SEM° (%)**
Knee-valgus angle	-1.1	4.1	1 (90%)	-1.5	3.9	1.4 (93%)	-2.6	2.9	1.2 (46%)	-2.5	3.2	1.4 (56%)
Knee -flexion angle	65.4	7.6	2.6 (3.9)	66.1	7.2	3.3 (4.9%)	59	11.2	2.7 (4.5%)	58.7	10.3	3.7 (6.3%)
Hip-adduction angle	9.1	6.5	1.3 (14.2%)	10	6.1	1.7 (17%)	7.3	6.3	0.63 (8.6%)	7.1	6.2	1.1 (15.4%)
Hip-internal rotation angle	10.8	7.9	1.12 (10.3%)	9.8	8.1	1.4 (14.2%)	8	6	1.5 (18.7%)	8.5	6	1.6 (18.8%)
Knee-valgus moment	0.3	0.13	0.08 (26%)	0.2	0.14	0.09 (45%)	0.2	0.14	0.09 (45%)	0.2	0.15	0.12 (60%)
Knee-extensor moment	3	0.5	0.14 (4.6%)	3.2	0.5	0.3 (9.3%)	3.1	0.5	0.2 (6.4%)	3.1	0.5	.23 (7.4%)
Hip-adduction moment	-1.8	0.4	0.14 (7.7%)	-1.7	0.4	0.15 (8.8%)	-1.9	0.2	0.12 (6.3%)	-1.9	0.2	0.16 (8.4%)
Hip-internal rotation moment	-1	0.2	0.11 (11%)	-1	0.2	0.16 (16%)	-1.2	0.3	0.13 (5.9%)	-1.2	0.3	0.12 (10%)
Ground-reaction force	3.4	0.5	0.15 (4.4%)	3.6	0.4	0.27 (7.5%)	3.5	0.44	0.22 (6.2%)	3.6	0.4	0.24 (6.6%)

[a] All angles in degrees. All moments in Newton meter per kilogram, Ground reaction force = *body weight. SEM = Standard error of measurement. Within-day mean = the mean of visit 1 mean and visit 2 mean. Within-day SD = the average of visit 1 and visit 2 standard deviations. Between-days mean = the mean of visit 1 mean and visit 3 mean. Between-days SD = the average of visit 1 and visit 2 standard deviations. FSLLP = forward single-leg landing off a platform

Table 4.17: Within-day and between-days means, SDs and SEMs for 3D variables during LSLL[a]

LSLL												
Variable	**Right leg**						**Left leg**					
	Within day			**Between days**			**Within day**			**Between days**		
	mean	**SD**	**SEM**	**Mean**	**SD**	**SEM**	**Mean**	**SD**	**SEM**	**Mean**	**SD**	**SEM**
Knee-valgus angle	-4.2	3.5	1.7 (40%)	-4.18	3.1	1.3 (31%)	-3.4	2.8	0.86 (25.2%)	-3.4	2.8	1.1 (32.3%)
Knee-flexion angle	59.4	6.7	2.7 (4.5%)	58.7	7.9	3.6 (6.1%)	55.7	8.8	2 (3.5%)	56	8.7	1.9 (3.3%)
Hip-adduction angle	6	6	1.5 (25%)	5.8	6.1	1.9 (32%)	5.7	5.2	1.4 (24.5%)	5.2	5	1.5 (28.8%)
Hip-internal rotation angle	12.1	7.1	1.4 (11.5%)	10.9	6.4	1.3 (11.9%)	8.8	6.4	1.7 (19%)	9.1	6.8	2.4 (26%)
Knee-valgus moment	0.3	0.12	0.1 (33%)	0.3	0.13	0.1 (33%)	0.15	0.09	0.05 (33%)	0.15	0.08	0.06 (40%)
Knee-extensor moment	2.5	0.5	0.22 (8.8%)	2.5	0.4	0.23 (9.2%)	2.3	0.4	0.1 (4.3%)	2.3	0.5	0.17 (7.3%)
Hip-adduction moment	-1.4	0.3	0.09 (6.4%)	-1.5	0.3	0.1 (6.6%)	-1.8	0.2	0.13 (7.2%)	-1.8	0.2	0.15 (8.3%)
Hip-internal rotation moment	-0.8	0.2	0.09 (11.2%)	-0.7	0.2	0.06 (8.5%)	-0.9	0.3	0.09 (10%)	-0.9	0.3	0.08 (8.8%)
Ground-reaction force	2.4	0.4	0.14 (5.8%)	2.4	0.4	0.14 (5.8%)	2.4	0.34	0.1 (4.1%)	2.4	0.34	0.11 (4.5%)

[a] All angles in degrees. All moments in Newton metres per kilogram, Ground reaction force = *body weight. SEM = Standard error of measurement. Within-day mean = the mean of visit 1 mean and visit 2 means. Within-day SD = the average of visit 1 and visit 2 standard deviations. Between-days mean = the mean of visit 1 mean and visit 3 mean. Between-days SD = the average of visit 1 and visit 2 standard deviations. LSLL = lateral single-leg landing

Table 4.18: Within-day and between-days means, SDs and SEMs for 3D variables during LSLLP[a]

LSLLP												
Variable	**Right leg**						**Left leg**					
	Within day			**Between days**			**Within day**			**Between days**		
	mean	**SD**	**SEM**	**Mean**	**SD**	**SEM**	**Mean**	**SD**	**SEM**	**Mean**	**SD**	**SEM**
Knee-valgus angle	-3	3.9	1.9 (63%)	-2.9	3.7	1.8 (62%)	-3.3	2.3	1.4 (42%)	-2.8	2.3	1.5 (53%)
Knee-flexion angle	63.7	6.5	3 (4.7%)	63.8	8.1	5.3 (8.3%)	59.3	7.7	3.3 (5.5%)	60	7.8	3.6 (6%)
Hip-adduction angle	6.8	6	1.7 (25%)	8.4	6	2.3 (27.3%)	5.5	5.5	2.3 (12.6%)	5.6	5.3	1.6 (28.5%)
Hip-internal rotation angle	11.4	9.9	1.7 (14.9%)	12	9.6	3 (25%)	8.7	6.6	1.5 (17.2%)	9.4	6.8	1.7 (18%)
Knee-valgus moment	0.3	0.14	0.1 (33%)	0.3	0.15	0.1 (33%)	0.2	0.13	0.1 (50%)	0.2	0.12	0.07 (35%)
Knee-extensor moment	2.8	0.4	0.2 (7.1%)	2.8	0.4	0.21 (7.5%)	2.9	0.5	0.34 (11.7%)	2.8	0.5	0.35 (12.5%)
Hip-adduction moment	-1.7	0.4	0.13 (7.6%)	-1.7	0.4	0.13 (7.6%)	-2.1	0.4	0.3 (14.2%)	-2	0.14	0.1 (5%)
Hip-internal rotation moment	-1	0.3	0.16 (16%)	-1	0.3	0.08 (8%)	-1.1	0.2	0.08 (7.2%)	-1.2	0.3	0.2 (16.6%)
Ground-reaction force	3.6	0.5	0.22 (6.1%)	3.5	0.6	0.3 (8.5%)	3.6	0.7	0.31 (8.6%)	3.6	0.54	0.26 (7.2%)

[a] All angles in degrees. All moments in Newton meter per kilogram, Ground reaction force = *body weight. SEM = Standard error of measurement. Within-day mean = the mean of visit 1 mean and visit 2 mean. Within-day SD = the average of visit 1 and visit 2 standard deviations. Between-days mean = the mean of visit 1 mean and visit 3 mean. Between-days SD = the average of visit 1 and visit 2 standard deviations. LSLLP = lateral single-leg landing off a platform.

Table 4.19: Within-day and between-days means, SDs and SEMs for 3D variables during MSLL[a]

	MSLL											
Variable	**Right leg**						**Left leg**					
	Within day			**Between days**			**Within day**			**Between days**		
	mean	**SD**	**SEM**	**Mean**	**SD**	**SEM**	**Mean**	**SD**	**SEM**	**Mean**	**SD**	**SEM**
Knee-valgus angle	-1.8	5.6	0.79 (43%)	-2	5.5	1.3 (65%)	-1.8	3.1	1.4 (77%)	-2.1	2.7	1.5 (71%)
Knee-flexion angle	60	7	2.4 (4%)	58.3	7.9	2.8 (4.8%)	56.2	7.7	2.6 (4.6%)	56.2	9.4	3.8 (6.7%)
Hip-adduction angle	8.9	5.4	1.2 (13.4%)	8.7	5.2	1.7 (19.5%)	7	6	1.7 (24.2%)	7.1	5.7	2.2 (30%)
Hip-internal rotation angle	10.7	8	1.13 (10.5%)	10.5	8	1.4 (13.3%)	8.2	5.1	0.9 (10.9%)	8	5.5	1.6 (20%)
Knee-valgus moment	0.25	0.18	0.07 (28%)	0.25	0.18	0.08 (32%)	0.1	0.12	0.08 (80%)	0.1	0.11	0.09 (90%)
Knee-extensor moment	2.4	0.4	0.12 (5%)	2.5	0.4	0.13 (5.2%)	2.4	0.5	0.12 (5%)	2.3	0.5	0.14 (6%)
Hip-adduction moment	-1.4	0.3	0.01 (0.7%)	-1.3	0.2	0.1 (7.6%)	-1.6	0.2	0.05 (3.1%)	-1.7	0.3	0.15 (8.8%)
Hip-internal rotation moment	-0.7	0.21	0.07 (10%)	-0.7	0.3	0.12 (17.1%)	-0.9	0.22	0.06 (6.6%)	-0.9	0.2	0.07 (7.7%)
Ground-reaction force	2.4	0.31	0.11 (4.5%	2.4	0.3	0.1 (4.1%)	2.5	0.4	0.01 (0.4%)	2.5	0.4	0.01 (0.4%)

[a] All angles in degree. All moments in Newton meter per kilogram, Ground reaction force = *body weight. SEM = Standard error of measurement. Within-day mean = the mean of visit 1 mean and visit 2 mean. Within-day SD = the average of visit 1 and visit 2 standard deviations. Between-days mean = the mean of visit 1 mean and visit 3 mean. Between-days SD = the average of visit 1 and visit 2 standard deviations. MSLL = medial single-leg landing

Table 4.20: Within-day and between-days means, SDs and SEMs for 3D variables during MSLLP[a]

	MSLLP											
Variable	**Right leg**						**Left leg**					
	Within day			**Between days**			**Within day**			**Between days**		
	mean	**SD**	**SEM**	**Mean**	**SD**	**SEM**	**Mean**	**SD**	**SEM**	**Mean**	**SD**	**SEM**
Knee-valgus angle	-3.7	5.9	1 (27%)	-3.6	5	2.1 (58%)	-3.7	3.6	1.1 (29.7%)	-3.7	3.3	1.6 (43%)
Knee-flexion angle	64.7	7.2	3.1 (4.8%)	67.5	10	6.6 (9.8%)	62	9.8	3.4 (5.4%)	62	9.6	4.5 (7.3%)
Hip-adduction angle	9.7	5.8	1.9 (19.5%)	10.5	5.1	2.1 (20%)	8.2	6.7	.94 (11.4%)	8.2	6.4	1.3 (15.8%)
Hip-internal rotation angle	11.9	6.4	.91 (7.6%)	11.2	5.8	2.7 (24%)	10.6	6.2	1.6 (15%)	10.5	6.6	2.2 (20.9%)
Knee-valgus moment	0.5	0.3	0.08 (16%)	0.5	0.3	0.12 (24%)	0.24	0.2	0.11 (45%)	0.2	0.1	0.07 (35%)
Knee-extensor moment	3	0.5	0.17 (5.6%)	3	0.5	0.32 (10.6%)	2.9	0.7	0.25 (8.6%)	2.9	0.7	0.29 (10%)
Hip-adduction moment	-1.7	0.6	0.22 (8.1%)	-1.7	0.6	0.27 (15.8%)	-2	0.3	0.14 (7%)	-2	0.3	0.19 (9.5%
Hip-internal rotation moment	-1	0.3	0.18 (18%)	-1	0.3	0.14 (14%)	-1.2	0.2	0.11 (9.1%)	-1.2	0.2	0.11 (9.1%)
Ground-reaction force	3.4	0.6	0.2 (5.8%)	3.4	0.6	0.2 (5.8%)	3.5	0.53	0.21 (6%)	3.5	0.5	0.3 (8.5%)

[a] All angles in degrees. All moments in Newton meter per kilogram, Ground-reaction force = *body weight. SEM = Standard error of measurement. Within-day mean = the mean of visit 1 mean and visit 2 mean. Within-day SD = the average of visit 1 and visit 2 standard deviations. Between-days mean = the mean of visit 1 mean and visit 3 mean. Between-days SD = the average of visit 1 and visit 2 standard deviations. MSLLP = medial single-leg landing off a platform

4.6 Discussion

The aims of this study were to:

1. Examine within-day and between-days reliability for lower-extremity biomechanics using 2D motion analysis during multidirectional SLL.
2. Establish SEMs for 2D biomechanical measurements during multidirectional SLL.
3. Examine within-day and between-days reliability for lower-extremity biomechanics using 3D motion analysis during multidirectional SLL.
4. Establish SEMs for 3D biomechanical measurements during multidirectional SLL.

From a sport physiotherapy point of view, testing single-leg tasks helps in detecting lower-extremity instabilities better than double-leg tasks (Pappas et al., 2007). Many studies have examined different types of singe-leg tasks such as SLS (Munro et al., 2012a; Mauntel et al., 2014; Stickler et al., 2015; Whatman et al., 2011), side step and side jump (McLean et al., 2005), single-limb step-down (Hollman et al., 2009), SLL (Zeller et al., 2003; McLean et al., 2004a; Hass et al., 2005; Yu, Lin, & Garrett, 2006; Pappas et al., 2007; Yeow et al., 2010; Munro et al., 2012a), jumping (Willson & Davis, 2009), lunge hop and step down (Whatman et al., 2011).

Multidirectional SLL is commonly used as a rehabilitation exercise and as a screening tool post-injury. However, the reliability of these tasks using both 2D and 3D motion analysis is still unclear and needs to be established. Therefore, this study hypothesised that there would be within-day and between-days agreement between repeated measurement scores for all 2D and 3D variables during all examined tasks. This study also hypothesised that within-day agreement would be greater than between-days agreement.

The results showed that 2D and 3D variables in all tasks reported good to excellent within-day reliability ranging between ICC = 0.77–0.97 and 0.61–0.98, respectively. Most of the 2D and 3D variables showed between-days reliability ranging between 0.67–0.96 and 0.42–0.97, respectively (Tables 4.5, 4.9–4.14). Less than satisfactory between-days agreement was reported for 3D left-knee valgus moment and left-knee adduction moment during most of the tasks, particularly those being performed off platform. Although other studies have reported low reliability for these variables (Alenezi et al., 2014; Myer et al., 2015b) our findings are even lower than these studies. There might be several reasons for this finding.

First, there is the nature of the task, it is highly demanding and challenging because it increases the demand and load on the limb due to increasing the landing impact on the musculature, which results from a samller base of (Pappas et al., 2007). This may lead to high between-subjects variability between trials. Variability has been found to increase when the height of the landing increases (James et al., 2000). Therefore, the variability in the current study may be due to examining SLL from different heights and/or arms and using contralateral leg-swing strategies. Some studies have suggested that high variability in movement may be a potential risk factor in gymnastics (Comfort, Colclough, & Herrington, 2016). In contrast, James et al. (2000) and Bartlett, Wheat, & Robins (2007) suggest that movement variability allows better distribution of the load among different tissues, which in turn may prevent overuse injury. However, the actual relationship between movement variability and musculoskeletal injury is still unclear. To determine whether variability is a risk factor or not, a reference value for variability should be known to decide if the variability in any task is within normal limits or not. However, such a value has not been established in the literature. Moreover, joint moment was calculated using inverse dynamics that may lead to noise which, in turn, may effect the calculated linear and angular velocities and acceleration of segments and then affect the consistency of moment scores (Blajer, Dziewiecki, & Mazur, 2007).

Second, the current study examined lateral and medial SLL, which was observed to include trunk movement towards the side of the leg landed on. Given the weight distribution of body segments, it is found that the trunk makes up the largest percentage. Tözeren (1999) reported that the trunk weight is 48.3% of total body weight for women and 50.8% for men. Movement of this weight above the lower part of the body while one leg is fixed on the ground may place high stress on the muscles and joints and may influence the moment. In addition, movement of the trunk could shift the centre of mass onto the landing limb and thus affect the frontal-plane moment (Powers, 2010; Dos Reis et al., 2015).

Third, the dominant leg might affect performance as a less than satisfactory level of agreement was found for one leg only (mainly the left leg). Most of the participants in the current study were right-dominant (defined as the preferred leg to kick the ball). Ortiz et al. (2016) reported less reliability for the non-dominant leg during DVJ.

Fourth, although a suitable rest time between tests was allowed for each participant, the possibility of fatigue still exists in this study as it includes many tasks, and each task may be performed many times.

Finally, participants were moderately active but the types of activities they practised was unknown. Therefore, their sport might not include highly demanding tasks such as those examined in the current study, which may make it difficult for them to perform. A better level of agreement could be achieved when examining participants who practise sport or do activities that include SLL.

The findings of the current study indicate good to excellent within-days agreement and moderate to excellent between-days agreement. This implies that the measurements were consistent when taken by one examiner. Therefore, the null hypothesis was rejected. These results support the use of 2D and 3D motion analysis systems as reliable tools in examining lower-extremity biomechanics during functional tasks. Furthermore, such findings support the use of multidirectional SLL as a reliable and useful functional test when examining lower-extremity biomechanics. To the best of the researcher’s knowledge, this is the first study to report the 2D and 3D lower-extremity frontal-plane kinematics and kinetics during multidirectional SLL. Previous literature has reported the reliability of only one direction of SLL (mostly forward) using either 2D or 3D motion analysis, or either kinematics only or kinetics only. However, a comparison with the available literature can be made.

This study’s findings are consistent with previous studies. Many studies show an ICC ranging between 0.59 and 0.98 for both 3D and 2D motion analysis during different tasks (Munro et al., 2012a; Gao et al., 2012; Norris & Olson, 2011; Sled et al., 2010; Hollman et al., 2009; Miller & Callister, 2009; Ferber et al., 2002; Zeller et al., 2003; Ford et al., 2003; McLean et al., 2004a; Mclean et al., 2005)

With regard to 2D reliability, Munro et al. (2012a) reported good within-day ICC reliability ($\geq$ 0.59 - 0.88) and good to excellent between-day ICC reliability ($\geq$ 0.72 - 0.91). However, this study only examined FPPA during FSLL. The authors concluded that 2D analysis is a reliable measurement tool for lower-extremity dynamic FPPA. 2D FPPA has also been examined to predict or screen for knee injuries (Norris & Olson, 2011; Munro et al., 2012a; Mclean et al., 2005). During step down, 2D video analysis has shown excellent intra-rater reliability for knee valgus and HADD (Hollman et al., 2009). During other performance tests, moderate to high reliability for knee valgus has been reported (Miller & Callister, 2009). 2D sagittal-plane measurement has also shown excellent inter-rater and intra-rater reliability during mechanical lifting (Norris & Olson, 2011). The current findings are also consistent with a recent study by Belyea et al. (2015), who examined the reliability of the KinesioCapture app, a 2D motion-analysis application. ICCs ranging between 0.73–0.94) were reported. However, the task examined in the current study is more challenging as it shifts the load of the body onto one limb, while they examined a double-leg task. Moreover, they did not mention the time between test-retest sessions. This is an important point, as it should reflect the clinical setting and the time between test-retest sessions may affect the findings of any reliability study (Ross, 1997).

Comparing 3D findings, Alenezi et al. (2014) and Myer et al. (2015b) reported excellent within-day reliability for lower-limb kinematic and kinetic variables during SLL and single-leg cross-landing, respectively, ranging between 0.80 and 0.97. with CI ranging between 0.39 and 0.99, except knee-internal rotation angle and hip=adduction moment, which showed moderate to good reliability of 0.78 with CI = -0.33–0.93, and 0.63 with CI = -0.08–0.88. In a study that examined the reliability of single-leg drop jump and single-leg up-down tasks, Ortiz et al. (2007) compared between the reliability of these tasks when examining the averages of one, two, three, four and five trials. They found good reliability for both kinematic and kinetic variables, ICC $\geq$ 0.75 and ICC $\geq$ 0.86, respectively, when examining an average of five trials. The single and two-trial averages also showed good reliability for both kinematic and kinetic variables during an up-down task, but not a single-leg drop jump, with ICC $\geq$ 0.77 and ICC $\geq$ 0.86, respectively. Positively, this study reported SEMs and 95% CIs for all measurements. However, this study did not mention the time interval between test-retest sessions and only examined female subjects, which makes the findings inapplicable to male subjects.

Similar to the 3D findings of the current study, Whatman et al. (2011) found within-day ICC $\geq$ 0.92 and between-days ICC $\geq$ 0.80 for 3D hip and knee kinematics during small-knee bend, single-leg small-knee bend, step down and hop lunge.

In the current study, between-days agreement was lower than within-day agreement, which supports previous studies (Alenezi et al., 2014; Myer et al., 2015b). Regardless of the tasks examined, our findings also support the findings of other studies that evaluated 3D joint kinematics during other functional tests. For instance, Ford et al. (2007) reported a within-day ICC of 0.90 and a between-days ICC of 0.77 for most lower-limb kinematics during DVJ on a group of school football players (n = 11).

The difference between within-day and between-days ICC values might be due to lower errors in marker placement for within day compared to between days (Queen et al., 2006). Variations in subjects' performances and the difficulties of tasks may play an important role in this result. However, the CAST-model protocol was employed in this study in order to reduce re-application of markers error, as this model reported superiority over other models as it reduces skin-movement artefacts by attaching markers to the centre of segments. The difference between within-day and between-days reliability could have been minimised through eliminating marker placement by marking the skin, which it has been suggested increases the accuracy of marker placement (Ford et al., 2007). Moreover, the measurements during three visits were taken at different times of the day. This may have influenced the performances of participants. It might be better if the second and third visits were measured at the same time as the first one. However, this was beyond the researcher's control due to difficulties on making lab bookings and participants having spare time.

Furthermore, the current study included many tasks and repetitions. This could lead to systematic bias, which might be present because of fatigue or a learning effect. However, this was controlled by allowing practice trials for all subjects, randomization of the test order and allowing sufficient rest period between tests.

The current study has reported within-day and between-days SEMs for 2D and 3D lower-limb biomechanics during multidirectional SLL (Tables 4.6, 4.15–4.20). SEM is an important measurement as it makes prediction for any measurement and gives the range where the true value of any measurement is likely to lie (Denegard & Ball, 1993). Knowing such information about any measurement allows for accurate evaluation between tests changes and thus determines whether a change is real or due to measurement error (Munro et al., 2012a; Fletcher & Bandy, 2008; Domholdt, 2005). For instance, when a clinician evaluates 2D right-leg HADD angle pre-and-post intervention during LSLL, the assessor can be confident that the true scores lie within 1.65° (if measured on the same day) and within 2.05° (if measured on different days). Moreover, Portney and Watkins (1993) state that knowing the SEM makes the clinician 68% confident that the true value lies between +1 and – 1 standard deviation of the observed value. Most previous studies present SEM as absolute value. In the current study, both absolute and percentage SEMs are reported. However, it is difficult to decide whether the SEMs of the current findings are small or large because there is no existing standard scale that can be invoked in such decisions. In the literature, Dobson et al. (2017) suggest that acceptable SEMs for a chair-stand test, an 11-stair climb test and a 40 m, fast-paced walk test for patients with hip and knee osteoarthritis are < 10%. However, such a scale does not exist for biomechanical variables during other functional tasks. %SEMs were below than 10% of actual values in many variables. However, others were greater. This seems large when looking at %SEMs for some variables, but considering that the current study was dealing with variables with small absolute values, it may make them acceptable. For example, a %SEM of 26% looks relatively large, but when taken as an absolute SEM, which was 0.08 Nm/Kg, and comparing it with the actual value of knee-valgus moment (0.3 Nm/Kg), it might be small in reality. This is applicable to all 2D and 3D variables and for both within-day and between-days findings. To the best of the researcher's knowledge, no study has reported SEMs for 2D and 3D lower-extremity frontal plane biomechanics during multidirectional SLL. Yet, a comparison with the available literature can be conducted.

For kinematic variables, 2D measurements reported within-day SEMs ranging between .65° and 1.88° for both legs. The lowest SEM was reported for right-leg FPPA during LSLLP, while the highest was reported for right-leg HADD angle during MSLLP (Table 4.6). Between-days SEMs were reported range from 0.63° to 2.7° for both legs. The lowest between-days SEM was reported for right-leg FPPA during LSLLP, while the highest was reported for right-leg HADD angle during MSLLP (Table 4.6). Belyea et al. (2015) reported SEMs for 2D FPPA at IC and at maximum knee flexion during DVJ using a hand-held tablet. Comparing to the current study, Belyea et al. (2015) reported larger SEMs for 2D FPPA at both IC and maximum knee flexion (1.9° and 5.1°, respectively). The differences might be attributed to differences in the mean age of participants, 21 ± 1.4 years compared to 27.6 ± 4.4 in the current study. The nature of the task examined may affect the findings as well. They examined a double-leg task while the current study examined a single-leg task, which is highly demanding and more challenging because it increases the demand and load on the limb due to increasing the landing impact on the musculature as the base of support decreases (Pappas et al., 2007). A difference in the methods used to measure 2D FPPA between the current study and Belyea et al. (2015) study is that they used an uncommon procedure (hand-held tablet). The validity of this procedure needs more investigation. The methods in the current study are the most commonly used, and their reliability and validity are well established (Norris & Olson, 2011; Olson et al., 2011; Sigward et al., 2011; Mauntel et al., 2014; McLean et al., 2005; Nielsen & Daugaard, 2008; Sigward et al., 2008; Willson & Davis, 2008).

Munro et al. (2012a) reported higher SEMs than the current study (2.72 and 2.85 for men and women, respectively) during SLL. However, this was only for FPPA during forward SLL. Differences in the mean age of the participants may affect the results. The nature of the tasks examined may also affect the findings. Even though Munro et al. (2012a) examined SLL and the height of the platform was similar to the current study, the distance between the platform and the force platform is unknown, while it was 30cm away from the force platform in the current study.

For 3D variables, measurements reported within-day SEMs ranging between 0.01° and 3.4° for both legs. The lowest within-day SEM was reported for right-HADD moment and left GRF during MSLL. The highest was reported for left-knee flexion angle during MSLLP. Between-days SEMs were reported to range from 0.01° to 6.6° for both legs. The lowest between-days SEM was reported for left-leg GRF during MSLL, while the highest was reported for right-knee flexion angle during MSLLP (Tables 4.19 and 4.20). Alenezi et al. (2014) reported within-day and between-days SEMs very similar to the current study, ranging from 0.08–3.35) and 0.08–3.27, respectively. However, they only examined the right leg and subjects were asked to land with their arms across their chest, which does not represent the real situation of landing. In the current study, the highest SEM was reported for sagittal-plane movement across all tasks. This was expected, given the large ROM in sagittal-plane movement. However, it is generally lower than the SEMs reported by Ford et al. (2007) and Whatman et al. (2013) for sagittal-plane movement during a drop jump. However, the time between test-retest sessions in both studies does not replicate the actual clinical setting (7 and 10 weeks, respectively) and is longer than the current study, which may lead to larger SEMs in their studies.

The findings of the current study are important. Studies that examined SLL using 2D motion analysis mainly focused on FPPA. HADD has been suggested as being associated with knee injury and has important clinical implications in injury prevention and rehabilitation (Maykut et al., 2015). The current study adds to the literature, in that 2D HADD angle is also a reliable variable when examining lower-extremity biomechanics during an SLL task. Therefore, clinicians can confidently and reliably use 2D motion analysis to describe HADD angle during a clinical programme aiming to predict, prevent or rehabilitate hip injury, particularly in large-scale populations and/or in the absence of a 3D motion-analysis system.

Moreover, clinicians do not only use forward SLL, other directions are also commonly used. However, the biomechanical implications of these tasks are still unclear. The current study adds to the literature, in that other directions are also reliable and can be used to assess lower-extremity function and performance using both 2D and 3D motion analysis. Clinicians usually use the limb-symmetry index (LSI) to evaluate readiness to return to sport after injury. The injured leg should reach 80–90 per cent or above of the level of the uninjured leg of clinicians are to decide whether a patient should return to sport. This assumes that the unaffected leg is "normal" (Clark, 2001b). However, there is no evidence to support such an assumption. Our findings provide a better understanding of the 2D and 3D biomechanics of both lower extremities during these tasks, which will help in better understanding the potential mechanisms related to injury-risk factors as well as help practitioners take the right decisions on enough information. This may help in reducing the occurrence of re-injury. Knowing the SEM for any measurement is very important, particularly in clinical applications. Consequently, clinicians can now confidently evaluate the effect of any intervention aiming to change the biomechanics of the lower limbs using both 2D and 3D motion analysis during SLL tasks by applying the value of SEM to their measurement. Any change in measurements less than the SEM value means that the intervention does not have a significant effect.

The results of this study are subject to some limitations. Participants were healthy, moderately active and their ages were limited to 18–35 years old. Therefore, it is only applicable to the same population. Other populations, such as injured people or players of specific types of sports that include SLL tasks, such as footballers, need to be examined. The participants in this study were examined using standardised shoes and on a Mondo running surface. Although such a shoe was used to standardise the effect of shoe wear between subjects and such a surface can reflect the real situation of some sports such as running and volleyball, the interaction between shoe and surface may not reflect the actual interaction for some sports such as football or other sports that are played on grass. Hence, examining subjects wearing real sport shoes on a grass surface may push the literature forward. Finally, the results of this study are limited to one examiner. Between-rater reliability needs to be examined.

4.7 Conclusion

The results of this study led to acceptance of all the hypotheses and then provided important guidance and recommendations for clinicians when examining multidirectional SLL. It can be concluded that most of the 2D and 3D variables in a young healthy population are reliable when examining SLL tasks. Furthermore, multidirectional SLL is a reliable test to examine lower-extremity biomechanics. Such methods and tests should be employed to screen individual lower-extremity biomechanics and in injury-prevention studies. However, the relationship between the two motion-analysis techniques used in this study and whether the 2D can be a good alternative for 3D are still unclear and need to be examined.

5. Study three: Concurrent validity of two-dimensional analysis of lower-extremity frontal plane of movement during multidirectional single-leg landing

5.1 Study aims

1. To evaluate the validity of 2D video analysis of knee frontal-plane kinematics during multidirectional SLL when compared to 3D data.

2. To evaluate the validity of 2D video analysis of hip frontal-plane kinematics during multidirectional SLL when compared to 3D data.

5.2 ntroduction

The ability to screen injury-risk factors is a key role in the prevention of sport injuries as it helps in modifying modifiable risk factors (McLean et al., 2005). Some studies have shown reduced ACL injury through screening individuals with high-risk lower-extremity biomechanics, then undertaking appropriate training (Myer et al., 2005).

The gold standard for motion-screening is 3D motion analysis (McLean et al., 2005), as it provides accurate and reliable 3D lower-extremity measurements while performing different sport tasks (Gao et al. 2012; Sled, Khoja, Deluzio, Olney, & Culham, 2010; McLean et al. 2004a) and contributes effectively to screening and the rehabilitation of injuries related to these tasks. It can accurately describe both multiplane joint angles and moments during functional tasks. However, its application in a clinical setting or to a larger sample size is limited due to the high financial cost and the time-consuming nature of data collection (Willson & Davis, 2008; Nielsen & Daugaard 2008; Hewett et al. 2005; McLean et al. 2005). This suggests a need for simpler and clinically applicable alternatives. 2D motion analysis has become popular in clinical practice. It only requires a digital video camera and digitizing software. Stensrud et al. (2011) have reported that 2D motion analysis is universally available, reasonably cheap and typically portable. 2D motion analysis has been used to evaluate lower-extremity kinematics in healthy and injured populations (Herrington & Munro, 2010; Stenstrud et al., 2011; Herrington, 2011; Noyes et al., 2005; Willson & Davis, 2008). However, it is not without its flaws. For instance, it has questionable ability to capture complex and multiplanar dynamic movement. Such a limitation led many studies to question and examine the validity of 2D motion analysis during functional tasks (Mizner et al., 2012; Norris & Olson, 2011; Olson et al., 2011; McLean et al., 2005; Willson & Davis, 2008; Glass et al., 2008).

The findings of the afore-mentioned studies are conflicting with the correlation between 2D and 3D measurements ranging from 0.15 to 0.77. Discrepancies in findings may be due to 2D measurement methods and the tasks examined (Nagano, Sakagami, Ida, Akai, & Fukubayashi, 2008). Jones et al. (2014) attributed these conflicting findings, leading to differences between 2D and 3D motion analysis, to the fact that, in 2D, knee flexion can appear as a relatively knee-abducted position, particularly when the hip is internally rotated. This suggests that 2D validity, particularly in clinical use, is still unclear and needs more investigation. Moreover, most of the aforementioned studies examined bilateral tasks and mostly concentrated on the sagittal plane. Bilateral tasks are less challenging and may mask some important events that can occur during SLL which may more closely match the real situation of landing in sports. The majority of studies examining the frontal plane have mainly focused only on FPPA and have not assessed hip adduction.

Excessive movement within the frontal plane is important because it is considered a risk factor of knee injury and associated with non-contact ACL injury, particularly knee valgus and HADD (Paterno et al., 2010). Knee valgus collapse has been identified as a position of injury during dynamic movement (Krosshaug et al., 2007) and knee valgus and knee loading correlate with ACL injuries (Shin et al., 2011). Increased load within the frontal plane (abduction/ adduction load) has been found to increase ACL tension and apply high force to the ACL, thus increasing the risk of injury (Shultz et al., 2007). During landing, gender differences have been found only in the biomechanical variables of the frontal plane and time to peaks of these variables (Joseph et al., 2011; Kernozek et al., 2005). Such findings may explain the gender differences in ACL injury, and this suggests the importance of examining frontal-plane biomechanics during different landing tasks.

To the best of the researcher’s knowledge, no study has examined both hip and knee frontal-plane kinematics during a battery of single-leg tasks. Consequently, the aim of this study was to examine the validity of 2D motion analysis against 3D motion analysis when examining lower-extremity frontal-plane kinematics variables (FPPA and HADD angle) during multidirectional SLL.

5.3 Study hypothesis

Depending on the previous literature, these hypotheses were formulated:

Alternative hypotheses

H_1: 2D FPPA correlates with 3D knee-valgus angle during multidirectional SLL.

H_2: 2D HADD angle correlates with 3D HADD angle during multidirectional SLL.

Null hypotheses

$H0_1$: There is no relationship between 2D FPPA and 3D knee-valgus angle.

$H0_2$: There is no relationship between 2D HADD angle and 3D HADD angle.

5.4 Methods

5.4.1 Participants

Based on pilot work conducted on 12 subjects (the same subjects as in study 2), the lowest correlation ($r^2 = 0.22$) was used to calculate the power. Choosing the lowest r^2 made the researcher confident of recruiting the required sample for all variables. Therefore, using an r^2 value of 0.22, a sample-size calculation was performed using G power 2 statistical software, which showed that 34 subjects were required when power = 0.8 and the α level = 0.05 (Appendix V). Therefore, 34 healthy adults (19 male and 15 female), moderately active subjects, were voluntarily recruited from the staff and student populations of the University of Salford. Sample demographics are presented in Table 5.1.

5.4.2 Inclusion and exclusion criteria

The same criteria as for the reliability study were applied (see section 4.4.1)

5.4.3 Instrumentation, setup and study procedure

The same instrumentation, system setup and procedure as for study two were applied, apart from repeat visits (see sections 4.4.2–4.4.6).

Table 5.1: Sample demographics of validity study

		Number	Mean	SD	Minimum	Maximum
Age (years)	Male	19	28.6	4.5	20	35
	Female	15	26.8	2.9	24	31
	All	34	28	3.9	20	35
Height (m)	Male	19	1.7	0.04	1.68	1.79
	Female	15	1.64	0.04	1.59	1.7
	All	34	1.7	0.05	1.59	1.79
Body mass (kg)	Male	19	71	4.5	65	80
	Female	15	62.2	9.7	53	80
	All	34	67.7	7.9	53	80

SD = Standard deviation. M = Metres. Kg = Kilograms.

5.4.4 Statistical analysis

The variables examined in a validity study were tested for normality using a Shapiro Wilk's test. The association of 2D variables (FPPA and HADD) with corresponding 3D variables (knee-valgus angle and HADD angle) was examined in both legs using Pearson's product-moment correlation (*r*). This correlation evaluates the linear relationship between two random variables. Its value ranges from -1, which indicates negative correlation, through 0, which indicates no correlation, to +1, which indicates positive correlation (Zou, Tuncali, & Silverman, 2003). The classification of strength of correlation is small (0–0.3), moderate (0.3–0.5), strong (0.5–0.7) and very strong (0.7–1), as described by Hopkins, Marshall, Batterham, and Hanin, (2009). However, Pearson correlation evaluates how variables relate to each other. To evaluate how a 2D variable can explain and account for the variability of corresponding 3D variables, a linear regression analysis (r^2) was performed using 2D variables as independent (predictor) variables and 3D variables as dependent (predicted) variables. This can determine the nature of the correlation between variables, and the amount of variance of the predicted variables can be explained by the predictor variables. One

Sample T test was used to examine systematic difference. If there was no significant difference, Bland-Altman Plots were used to evaluate systematic bias.

5.5 Results

5.5.1 Test of normality

All 2D (FPPA and HADD angle) and 3D (knee valgus and HADD angle) variables reported a *P* value greater than 0.05, indicating normality. Appendix VI illustrates the results of normality tests for all the 2D and 3D variables of both legs during all tasks.

5.5.2 Descriptive characteristics

Descriptive characteristics (mean ± SD) for 2D and 3D variables for both legs in all tasks are presented in Table 5.2.

Table 5.2: Descriptive means (± SD) for 2D and 3D variables in all tasks

Variable	Tasks											
	FSLL		FSLLP		LSLL		LSLLP		MSLL		MSLLP	
	RT	LT	RT	LT	RT	LT	RT	LT	RT	LT	RT	LT
	2D variables mean (SD)											
FPPA (°)	-7.9 ± 4.8	-5.6 ± 4.2	-8.3 ± 5	-7.1 ± 5.9	-5.8 ± 3.6	-3.9 ± 2.2	-6.3 ±3.5	-4.4 ± 4.4	-9.7 ± 6.4	-6.9 ± 5.5	-12.4 ± 8.4	-9.3 ± 6.3
HADD (°)	7.6 ± 4.7	6.7 ± 4.5	7.3 ± 5.1	7.5 ± 5.7	3.6 ± 4.7	4.3 ± 2.2	4.8 ± 4.9	4.5 ±4.9	7.7 ±5	5.5 ± 5.7	9.6 ± 4.4	7.1 ± 5.5
	3D variables mean (SD)											
Knee valgus (°)	-1.3 ± 3.9	-.6 ± 3.9	-1.8 ± 4.2	-1.3 ± 3.7	-3.4 ± 3.3	-2.4 ± 3.5	-2.8 ± 3.6	-2.1 ± 3.7	-2 ± 5.1	-1 ± 3.5	-3.1 ± 5.5	-2.3 ± 4.2
HADD (°)	6.5 ± 4.9	5.8 ± 5.4	7.6 ± 6.3	6.6 ± 5.6	3.8 ± 6.2	5.1 ± 4.9	4.6 ± 6.4	5.3 ± 5.8	7.1 ± 5.6	6.7 ± 6.6	8.8 ± 5.3	7.2 ± 6.2

[a] All values are in degrees, FSLL = forward single-leg landing, FSLLP = forward single-leg landing off a platform, LSLL = lateral single-leg landing, LSLLP = lateral single-leg landing off a platform, MSLL = medial single-leg landing, MSLLP = medial single-leg landing off a platform, HADD = hip adduction angle, RT = right leg, LT = left leg, 2D = two-dimensional, 3D = three-dimensional, SD = standard deviation.

5.5.3 Validity

Table 5.3 shows Pearson correlation (r) (P value) and linear-regression analysis (r^2) results for 2D variables (FPPA and HADD angle) with corresponding 3D variables (knee valgus and HADD angle) for both legs during all tasks. Appendix VII illustrates scatter plots for the correlation between theses variables.

2D FPPA reported an association with 3D knee-valgus angle ranging from $r = 0.17$ to $r = 0.42$. The largest correlation (moderate) was noted between these variables in FSLL, LSLL and MSLL (Table 5.3). However, the smallest correlation ($r = 0.17$) was reported for the left leg during FSLLP and for the right leg during LSLLP. Interestingly, a small negative correlation existed between 2D FPPA and 3D knee-valgus angle in right legs during MSLLP.

Linear regression analysis reported r^2 values ranging between 0.03 and 0.17 for the right leg and between 0.03 and 0.16 for the left leg, indicating that 2D FPPA might be, at best, a moderate predictor, as it can explain up to 17 per cent of 3D knee-valgus angle but only for tasks performed without a platform. So, it might be difficult to explain 3D knee kinematics using 2D motion analysis during other tasks (Table 5.3).

One Sample T test revealed that there was a significant difference between all 2D FPPA and 3D knee valgus angles for both legs in all tasks (Appendix VIII).

2D HADD angle in both legs reported a strong and significant correlation with 3D HADD angle, ranging from $r = 0.70$ to $r = 0.90$ in all tasks, apart from the right leg during MSLLP, which reported only a small association ($r = 0.27$).

Linear regression analysis (r^2) revealed that 49–81 per cent of 3D HADD angle can be explained by 2D measurement (Table 5.3), which means that 2D HADD angle is a relatively good predictor of 3D HADD angle.

One Sample T test revealed that there was no significant difference between all 2D and 3D HADD angles for both legs in all tasks (Appendix VIII). Therefore, Bland-Altman plots were conducted for all these variables. The slope of regression in Bland-Altman plots indicated no systematic bias between 2D and 3D HADD angles for both legs in all tasks (Appendix VIII).

Table 5.3: Pearson correlation (r) (P values), and linear regression analysis (r²) for 2D variables with 3D variables for both legs during all tasks

Variable	FSLL				LSLL				MSLL			
	RT		LT		RT		LT		RT		LT	
	***r* (*p*)**	***r*²**	***r* (*p*)**	***r*²**	***r* (*p*)**	***r*²**	***r* (*p*)**	***r*²**	***r* (*p*)**	***r*²**	***r* (*p*)**	***r*²**
	3D knee-valgus angle											
FPPA	0.42* (0.014)	0.174	0.35* (0.043)	(0.12)	0.28 (0.11)	0.08	0.40* (0.02)	0.16	0.20 (0.24)	0.04	0.37* (0.03)	0.14
	3D HADD angle											
2D HADD	0.79** (<0.001)	0.62	0.70** (<0.001)	0.49	0.81** (<0.001)	0.66	0.72** (<0.001)	0.52	0.90** (<0.001)	0.81	0.88** (<0.001)	0.77
	FSLLP				**LSLLP**				**MSLLP**			
	RT		**LT**		**RT**		**LT**		**RT**		**LT**	
	3D knee-valgus angle											
FPPA	0.26 (0.13)	0.07	0.17 (0.33)	0.03	0.18 (0.31)	0.032	0.26 (0.14)	0.07	-0.02 (0.92)	0.0004	0.29 (0.096)	0.08
	3D HADD angle											
2D HADD	0.85** (<0.001)	0.72	0.85** (<0.001)	0.72	0.79** (<0.001)	0.62	0.88** (<0.001)	0.77	0.27 (0.13)	0.073	0.84** (<0.001)	0.71

2D = two-dimensional, 3D = three-dimensional, FSLL = forward single-leg landing, FSLLP = forward single-leg landing off a platform, LSLL = lateral single-leg landing, LSLLP = lateral single-leg landing off a platform, MSLL = medial single-leg landing, MSLLP = medial single leg-landing off a platform, FPPA, frontal plane projection angle, HADD = hip adduction angle. * = correlation is significant at the 0.05 level, ** = correlation is significant at the 0.01 level.

5.6 Discussion

In the last decade, analysing the biomechanics of lower extremities has become common clinical practice. However, considering the fact that 3D motion analysis is not usually available in clinical settings due to its high cost and the need for a large space and a professional operator, 2D motion analysis is usually used as an alternative and clinicians should have sufficient knowledge of how 2D motion analysis works so as to have an alternative to 3D. Unfortunately, the validity of 2D and the association between 2D and 3D lower-extremity biomechanics, particularly frontal-plane biomechanics, are still not well understood (Sorenson et al., 2015), especially during SLL. Accordingly, the current study was conducted to increase the knowledge about video-based motion analysis by examining the validity and correlation between 2D and 3D lower-extremity frontal-plane movement during multidirectional SLL. This is imperative for practitioners who intend to examine a large population. To the best of the researcher's knowledge, this is the first study to examine such a relationship during this variety of tasks. Other tasks have been examined such as DVJ (Belyea et al., 2015; Ortiz et al., 2016) and single-leg drop jump (Sorenson et al., 2015). Most of the previous literature has only assessed 2D FPPA to quantify 3D knee-valgus angle and only examined one leg (dominant) (Sigward et al., 2008; McLean et al., 2005). The current study examined 2D FPPA and HADD angle to quantify the corresponding 3D variables in both legs, which may help practitioners to accurately compare between legs.

The findings of the present study partially support the first hypothesis, as they indicate that 2D FPPA, at best, moderately correlates with 3D knee-valgus angle during forward, lateral and medial SLL (table 5.3). Linear regression analysis indicates that 2D FPPA can, at best, explain up to 17 per cent of 3D knee-valgus angle, but only for tasks performed without a platform. However, it might be difficult to explain 3D knee kinematics using 2D motion analysis during other tasks, as it reported very low r^2 values (Table 5.3).

Such findings are comparable with some of the literature, but not all. Slightly better correlation between 2D knee FPPA and 3D knee valgus was found during side step ($r^2 = 0.25$), side jump ($r^2 = 0.36$) (McLean et al., 2005), SLS ($r = 0.31$) (Schurr et al., 2017) and 5-repetition vertical jump ($r^2 = 0.34$) (Nagano et al., 2008). The slightly better results might be due to a number of reasons. First, there are differences between the method used in the current study and in McLean et al's. (2005) study. The joint centres in the current study were determined using markers during data collection, while they were determined manually during the digitisation process in McLean et al.'s (2005) study. Manual estimation of joint centres has been shown to be less reliable, which may have introduced bias into McLean et al.'s (2005) study. Second is the populations examined. For example, McLean et al. (2005) examined basketball players with playing experience of more than ten years. It is well known that this sport is very demanding and involves many single-leg manoeuvres, which means that the participants may adapt to perform the task better than those in the current study, and potentially in a more consistent manner. Moreover, they examined the dominant leg only, while the current study examined both legs. The dominant leg has been found to offer more postural support and stability (Decker et al., 2003). However, the similar study by McLean et al. (2005) found similar results to the present study during shuttle running (r^2 = 0.13).

Like the present study, Willson and Davis (2008) examined the utility of 2D FPPA in female subjects with and without PFPS. They found that 2D FPPA did not significantly correlate with 3D knee-valgus angle during SLS (r = 0.21). Such results make the current study's findings acceptable, as they examined less dynamic tasks which can be performed with more stability than those examined in the present study. However, Sorenson et al. (2015) recently reported a correlation between 2D FPPA and 3D knee valgus during single-leg drop landing, but less than that reported in the current study (r^2 = 0.06). This correlation increased to (r^2 = 0.72) at IC. It seems that the study of Sorenson et al. (2015) is more comparable to the present study as they examined healthy subjects with nearly the same averages of age, height and body mass. The task they examined is very like the FSLLP examined in the current study. However, they only examined female subjects, while both genders were examined in the current study, which makes the result generalizable for both genders. A good correlation was reported only at IC, where the leg is in a more erect position (less hip and knee flexion), which minimized out-of-plane error, while the measurements in the current study were taken with knee flexion. Out-of-plane error was found to increase when knee flexion exceeded 40° (Cheng & Pearcy 1999).

In contrast to the current study, Belyea et al. (2015) examined the correlation between 2D FPPA and 3D knee-valgus angle at maximum knee flexion during DVJ. Positively, a good number of participants (n = 22) and both genders were included. Significant correlation between the aforementioned variables was noted (r = 0.48). However, no significant differences between them were noted at maximum knee flexion. A handheld tablet to capture 2D video was used in Belyea et al.'s (2015) study. This may have affected the orientation of the tablet when collecting the data and affected the angle of the tablet relative to the plane of movement, which may result in parallax error, which, in turn, could affect the results. Parallax error can occur when the subject is viewed away from the optical axis of the camera (Kirtley, 2006). Hence, it might be better if they used a tripod to hold the tablet. Moreover, the between-subject variability might be less than it is in the current study, as they examined double-leg tasks which give additional support and stability. This might increase the correlation obtained in Belyea et al.'s (2015) study. The between-subjects variability in the present study may result from arm swing, as no instruction was given with regard to arm movement. However, the justification for this was to better reflect the real situation of landing.

It is clear that the findings regarding the validity of 2D FPPA against 3D knee valgus are conflicting and might be inversely correlated with the difficulty of the task. For instance, when performing a double-leg task, the two legs provide more base support and more stability than a one-leg task, which is expected to offer better measurement for the frontal plane of movement.

In the current study, it was observed that the participants struggled to quickly stabilise the lower limb when SLL, which resulted in movement of the knee from side to side. This may have led to different times when the peak of 2D FPPA and 3D knee valgus occurred during landing cycle. This was expected to affect the accuracy of the measurements. Such an observation may explain other studies that found a good correlation at IC (Sorenson et al., 2015), as measurements were taken with the knee nearly extended and before subjects started struggling with their balance. However, the occurrence of injury was suggested to be in a position of no return, which includes knee flexion, so measurements should be taken in a knee-flexion position.

Moreover, FPPA is not a single movement but rather a combination of movements, which include rotation. 2D measures movement in a constant line of the frontal plane, which does not take into account rotational movement. This may affect the ability of 2D FPPA to predict 3D knee valgus.

In the current study, there were differences in data-collection frame rates between 2D (sampled at 30 Hz) compared to the faster 3D sampling frame (100 Hz). This may have led to losing some full 2D pictures and to sampling rate error. Such limitations may explain the lack of significant correlation between 2D FPPA and 3D knee valgus (Maykut et al., 2015). Greater correlation may be gained with a higher 2D sampling rate. Nevertheless, using a higher sampling rate was not applicable in the current study, as discussed previously in section 4.4.4.2. Noteworthily, SLL is a dynamic task and evaluating it from 2D still images may also have led to errors.

2D FPPA measurements also overestimated values, compared to 3D. A possible explanation is the influence of sagittal-plane movement, as knee flexion can appear as knee abduction when the hip is internally rotated (Jones et al., 2014). This could have influenced our findings. Also, the correlation between time to balance and frontal-plane biomechanics was not measured in the present study, but it needs to be examined in future studies. Such an observation may encourage the researcher to find more accurate methods that can estimate the time of peak 2D FPPA and correspond to the same time as the peak 3D knee valgus angle.

By comparing the nature of the tasks examined in the current study with those examined in other studies, and reporting better correlation between 2D FPPA and 3D knee valgus, the current study's tasks are much more dynamic and involve greater force and testing of balance. As the tasks in the current study have not been examined well before, many concerns about them are still unresolved, e.g. the relationship between muscle force and balance with these tasks and how they differ in other tasks. Therefore, such relationships should be considered in future research.

Regardless of this, these findings might be clinically useful, especially when examining FSLL, LSLL and MSLL. It is a good indication if a quick, simple and reliable tool (2D) can account for 17 per cent of the variance of an expensive and time-consuming tool (3D), and if it can be employed to help in improving rehabilitation programmes. For example, during a rehabilitation programme, by aiming to reduce knee-valgus angle where there is no access to 3D motion analysis, therapists can, at least partially, realise the use of 3D motion analysis and still know 17 per cent of what is happening to their measurements by using the change in 2D measurements. Thus, some time and effort for the patient and therapist could be saved, with the ability to follow the outcomes of the intervention still the same. Moreover, further analysis was conducted to examine whether a correlation exists between 2D FPPA and 2D HADD angle. Moderate to strong correlation was reported between these variables in both legs during most tasks (Appendix IX). Linear regression (r^2) revealed that 2D HADD angle can explain up to 64 per cent of the variability of 2D FPPA (Appendix IX). This suggest that 2D measurement of HADD is clinically beneficial as it can explain the variability in FPPA. Examining HADD angle using 2D seems to be more important than FPPA when screening people, because it might be a significant source of changes in FPPA. Since HADD is a component of knee valgus, its increase may suggest an increase in knee valgus. This suggests that 2D motion analysis may be an applicable surrogate for 3D motion analysis, particularly for hip movement during landing

With regard to the validity of 2D HADD angle compared to 3D HADD angle, 2D HADD angle in both legs has strong positive correlation with 3D HADD angle, ranging between r = 0.70 and r =

0.90 in all tasks, apart from the right leg during MSLLP, which reported only a small association (r = 0.27). Linear regression analysis (r^2) revealed that 49–81 per cent of 3D HADD angle can be explained by 2D measurement (Table 5.3), which means that 2D HADD angle is a good predictor of 3D HADD angle, which led us to reject the second null hypothesis. Clinically, 2D kinematic measurement seems to be useful when assessing ACL injury risk. Compared to those who remained intact, individuals who later had an ACL injury were found to report greater 3D knee-valgus angle (Hewett et al. 2005). Since HADD is a component of knee valgus, its increase may suggest an increase in knee valgus. This suggests that 2D motion analysis may be an applicable surrogate for 3D motion analysis, particularly for hip movement during landing.

Most of the studies that have examined the relationship between 2D and 3D frontal-plane biomechanics focused only on the knee. To the best of the researcher's knowledge, only two studies have examined the relationship between 2D and 3D hip kinematics during treadmill running and single-leg drop jump. Sorenson et al. (2015) reported a similar correlation between 2D HADD angle (defined as hip FPPA) and 3D HADD angle (defined as frontal plane hip position) at IC during single-leg drop jump (r = 0.72) with 52 per cent of the variability of 2D HADD being explained by the variability in 3D HADD (r^2 = 0.52). This correlation increased to r = 0.84, with almost 70 per cent of the 2D hip FPPA being explained by the variability in the 3D hip frontal-plane position at maximum excursion. However, only female participants and one direction of landing were examined in the study by Sorenson et al. (2015).

Maykut et al. (2015) examined the correlation between 2D and 3D HADD angle during running. Moderate correlation between these variables was found in the left and right legs (r = 0.539, r^2 = 0.291; r = 0.623, r^2 =0.388, respectively). Although running is a more stable task than SLL, the lower correlation reported by Maykut et al. (2015) may be attributed to the possible existence of fatigue, as they analysed an average of five trials.

The strong correlation reported between 2D and 3D HADD angle in the present study may be attributed to several reasons. One is the nature of the variable itself, as it may reduce within-subject variability. HADD movement does not include rotation, which means that motion occurs in a constant line of the frontal plane, which allows 2D to measure movement accurately with no overestimated scores. This, contrary to FPPA, might have led to better estimation of the time when 2D and 3D peak HADD occur during the landing cycle. Moreover, 2D calibration was done carefully in the current study. The actual size of any object captured using a 2D camera is unknown unless calibration is done. Calibration was not included in any of the aforementioned studies, which may result in perspective or out-of-plane errors. Such errors occur when the subject moves outside the calibrated plane (toward a changed subject). To minimize such errors, the line of sight should align with the centre of motion (Kirtley, 2006). Out-of-plane errors may also occur when the subject moves outside the calibrated plane, which makes measurement to an assumed size incorrect (Payton, 2008).

The results of the current study are important. Studies that examined the validity of 2D motion analysis compared to 3D mainly focused on FPPA, and only during limited tasks. HADD has been suggested as being associated with knee injury. The current study suggests that 2D motion analysis can be a good and valid alternative to 3D when measuring HADD angle during single-leg tasks, such as those included in the current study. Some clinical advantages might be gained from simple 2D motion analysis during multidirectional SLL. Compared to 3D, using 2D can help the practitioner to screen and predict a large number of those who are at risk of knee injury. Although it might be less useful when predicting 3D knee valgus using 2D FPPA, 3D knee valgus can still be predicted using 2D HADD angle, as it is a main component of knee valgus and there is a direct correlation (McLean et al., 2004a). Consequently, when 2D HADD angle increases, a prediction of increased 3D knee valgus can be assumed.

5.7 Conclusion

This study forms part of an increasing body of evidence exploring the relationship between 2D and 3D for measuring knee and hip angles. The results of this study suggest that 2D motion analysis might be an applicable alternative method when measuring knee and hip angles, particularly in the field or in a clinic that does not have access to a 3D motion system. However, caution should be taken when using 2D analysis to predict 3D knee valgus angles, as it shows lower validity. Nevertheless, the clinical utility of such findings needs to be examined.

6. Study four: Intertask correlation for both 2D and 3D variables during multidirectional SLL

6.1 Study aims

1. To examine whether the 2D biomechanical characteristics in multidirectional SLL are related.

2. To examine whether there is a relationship between 3D biomechanical variables during multidirectional SLL.

6.2 Introduction

In recent decades, screening individuals to predict the risk of future injury and improve performance has become common practice, not only in professional sport but also at other levels of sports (Mottram & Comerford, 2008). Functional tests are the way most recommended to screen the lower extremities to evaluate quality of movement (Whatman et al., 2011). Functional tests are also frequently used to evaluate alterations to lower-extremity biomechanics, as they have been suggested as being associate with injury and performance. However, evidence for the validity and ability of functional tests to predict injury or performance is still unclear (Whatman et al., 2011). Regardless of this shortcoming, functional tests have superiority over traditional assessment methods, such as special orthopaedic tests, which may no longer considered adequate because they examine isolated muscles and/or joints alone, while functional tests can evaluate multiple joints and muscles within the context of athletes' or patients' function (Kivlan & Martin, 2012; Mottram & Comerford, 2008). Clinicians use such tests to make decisions about choosing exercises and to assess the progress of a patient during any rehabilitation programme.

Understanding the mechanism of knee injury is important for its treatment and prevention. Some studies have described the mechanism of ACL injury and concluded that knee-valgus collapse with the knee slightly flexed in combination with tibial external or internal rotation is the main mechanism of injury (Olsen et al., 2004). Such a result was obtained from analysing game videotapes. However, it seems difficult to understand the actual mechanism from analysing videotapes, as the injury occurs rapidly during games and practice (Nagano, Ida, Akai, & Fukubayashi, 2009). This makes the determination of exactly when injury occurred difficult. To gain a better understanding of the mechanism of injury, researchers analyse the biomechanics of the lower extremities during tasks that pose a high injury risk to the knee, these are commonly seen on the sports field, using motion capture in a laboratory environment.

Knee injuries are mostly non-contact in nature and usually occur during the landing phase of any sport manoeuvre (Krosshaug et al., 2007; Boden et al., 2000; Agel et al., 2005). At the time of injury, increased relative knee extension and valgus have been demonstrated by individuals (Krosshaug et al., 2007; Boden et al., 2000). Some factors associated with ACL loading, such as increased anterior shear force, knee abduction, knee abduction moment and decreased knee flexion (Markolf et al., 1995), have also been associated with injury.

Several studies have examined the correlation between biomechanics characteristics during functional tasks, such as side jump, 45° cutting and shuttle run (McLean et al., 2005), bilateral and unilateral landing (Pappas et al., 2007), stepping down and DVJ (Earl et al., 2007), 45° and 90° cutting (Imwalle et al., 2009), SLL, stepping and drop jump (Harty et al., 2011), jogging, single small-knee bending, double small-knee bending, lunge, hop and step down (Whatman et al., 2011), DVJ and 35° cutting (Kristianslund & Krosshaug, 2013), double small-knee bending and drop jump (Whatman et al., 2013), SLL, 90° and 180° cutting (Jones et al., 2014), SLL, SLS, DLL and double-leg squat (DLS) (Donohue et al., 2015), SLS, SLL and drop jump (Munro, Herrington, & Comfort, 2017).

All the aforementioned studies examined only female subjects, except Earl et al. (2007), Pappas et al. (2007) and Donohue et al. (2015), who included both genders. A small sample size was examined in all of the above studies, except those of Kristianslund and Krosshaug (2013), Donohue et al. (2015) and Munro et al. (2017). The former study examined the correlation between cutting and drop jump in 120 participants and reported a weak correlation for knee-valgus moment and a stronger correlation for knee-valgus angle. However, these findings were statistically insignificant. Donohue et al.'s (2015) study investigated the correlation and differences between SLL, SLS, DLL, and DLS in 34 female recreational athletes and found a correlation ($r \geq 0.5$) for maximum knee and hip flexion between both landings and squatting tasks. A correlation was also reported for maximum knee abduction, HADD angle and maximum knee-abduction moment between the two landings and between SLS and both landings ($r \geq 0.54$). Munro et al. (2017) recently examined 88 female football and basketball players to investigate the correlation between SLS, SLL and drop jump with regard to 2D FPPA. Significant correlations were found between tasks. However, the results of Kristianslund and Krosshaug (2013), Donohue et al. (2015) and Munro et al. (2017) may not be generalised to other populations as they only examined female handball, football and basketball players. The coefficient of determination (r^2) is a useful measurement when conducting a correlation study. It gives the proportion of variance of one variable that is predictable from the other (Jones et al., 2014). However, none of the aforementioned studies considered a calculation of r^2, except Jones et al. (2014), who reported that 40 per cent of knee-valgus angle during cutting can be explained by knee-valgus angle during SLL. This percentage reduces to 21% for knee-valgus moment. There is less generalisability for these findings as they only examined female footballers.

It is clear from the literature that the correlation between different tasks has been examined. However, most of the studies examined the correlation between double- and single-leg tasks and there are many major sporting tasks that have not been covered, particularly multi-directional single-leg tasks, which are important as they are where injuries most often occur.

Multidirectional SLL is a common task performed in many sports, such as tennis, squash and volleyball, and it is commonly associated with ACL injury. It is also used as a screening test to determine a return to sport (Xergia et al., 2013) as it gives information about neuromuscular deficits (Paterno et al., 2010). However, no investigation to date has examined the relationship between multidirectional SLL using either 2D or 3D motion analysis. Such data may provide a better understanding of the biomechanical factors associated with ACL injury which, in turn, could facilitate screening people at risk of ACL injury and their rehabilitation. Also, it is important to understand whether these functional tests are biomechanically similar or different. Therefore, the aim of this study was to examine intertask correlation of the knee and hip joints during multidirectional SLL using both 2D and 3D motion analysis systems. Comparisons of biomechanics among athletic tasks can explain the characteristics of these tasks and help in the identification of which tasks pose a risk of injury. This, in turn, helps in the prevention and treatment of injuries. A better understanding of between-tasks performance may provide insights into the consistency of biomechanical patterns employed by individuals during sporting tasks.

6.3 Study hypotheses

Alternative hypotheses

H_1: Correlation will be found between the examined tasks for 2D variables (FPPA and HADD angle).

H_2: Correlation will be found between the examined tasks for 3D variables (knee-valgus angle, HADD angle, knee-valgus moment, HADD moment and knee-extensor moment).

Null hypotheses

$H0_1$: Correlation will not be found between the examined tasks for 2D variables (FPPA and HADD angle).

$H0_2$: Correlation will not be found between the examined tasks for 3D variables (knee-valgus angle, HADD angle, knee-valgus moment, HADD moment and knee-extensor moment).

6.4 Methods

6.4.1 Participants

The same participants as in study three (see section 5.4.1). Sample demographics were presented previously in Table 5.1.

6.4.2 Inclusion and exclusion criteria

The same criteria as for the reliability study were applied (see section 4.4.1)

6.4.3 Instrumentation and setup

The same instrument and system setup as in study two (see sections 4.4.2–4.4.4)

6.4.4 Study procedure and data processing

The same procedure and data processing as in study two were applied (see sections 4.4.5–4.4.6)

6.4.5 Statistical analysis

All statistical analyses were performed using the Statistical Package for the Social Sciences (version 21, IBM SPSS Statistics). Descriptive statistics (mean and standard deviations) present the data descriptively. The normality of data was examined using a Shapiro Wilk's test.

Each 2D and 3D variable was analysed separately. In parametric data, between-tasks correlation in 2D variables (FPPA and HADD angle) and 3D variables (knee-valgus angle, HADD angle, knee-valgus moment, HADD moment and knee-extensor moment) was evaluated using Pearson product-moment correlation. In non-parametric data, correlation was assessed using Spearman's rank correlation (ρ). However, Pearson correlation evaluates how variables relate to each other. To evaluate how each variable can explain and account for variability of the other, linear regression analysis (r^2) was performed for parametric data. For normally distributed variables, a repeated-measures ANOVA with Bonferroni post hoc analysis was used to determine whether there were any significant between-tasks differences in all the variables. Nonparametric variables were examined using a Friedman test. The significance level was set at $P < 0.05$.

Categorisation of the strength of correlation was small (0–0.3), moderate (0.3–0.5), strong (0.5–0.7) and very strong (0.7–1), as described by Hopkins et al. (2009).

6.5 Results

6.5.1 Test of normality

Most of the 2D and 3D variables reported a *P* value greater than 0.05, confirming normality of the data, apart from right- and left-knee-valgus moment during FSLLP, right-HADD moment during MSLL, and right-knee-extensor moment during FSLL. Appendix X illustrates the results of normality tests for all 2D and 3D variables for both legs during all tasks.

6.5.2 Descriptive characteristics

Descriptive characteristics (man ± SD) for the 2D and 3D variables for both legs in all tasks are presented in Tables 6.1 and 6.2, respectively.

2D FPPA and 2D HADD angle seems to be greater in MSLL with and without a platform, while lower 2D FPPA was reported during LSLL with and without a platform. Females showed greater scores for both 2D variables in all tasks in both legs (Figs 6.1 & 6.2).

3D knee-valgus angle reported greater values in LSLL and MSLLP, particularly for female subjects (figure 6.3). As Figure 6.4 illustrates, FSLLP, MSLL and MSLLP reported 3D HADD angles greater than in other tasks for both legs and both genders. Generally, kinetic data during FSLLP and MSLLP were greater than in other tasks (Figs 6.5–6.7)

Table 6.1: Descriptive (mean ± SD) for the 2D variables in each task[a]

	Tasks											
Variable	FSLL		FSLLP		LSLL		LSLLP		MSLL		MSLLP	
	RT	LT	RT	LT	RT	LT	RT	LT	RT	LT	RT	LT
FPPA (°)	-7.9 ± 4.8	-5.6 ± 4.2	-8.3 ± 5	-7.1 ± 5.9	-5.8 ± 3.6	-3.9 ± 2.2	-6.3 ±3.5	-4.4 ± 4.4	-9.7 ± 6.4	-6.9 ± 5.5	-12.4 ± 8.4	-9.3 ± 6.3
HADD (°)	7.6 ± 4.7	6.7 ± 4.5	7.3 ± 5.1	7.5 ± 5.7	3.6 ± 4.7	4.3 ± 2.2	4.8 ± 4.9	4.5 ±4.9	7.7 ±5	5.5 ± 5.7	9.6 ± 4.4	7.1 ± 5.5

[a] = All values are in degree. FSLL = Forward single-leg landing, FSLLP = Forward single-leg landing off a platform, LSLL = Lateral single-leg landing, LSLLP = Lateral single-leg landing off a platform, MSLL = Medial single-leg landing, MSLLP = Medial single-leg landing off a platform, FPPA = Frontal plane projection angle, HADD = Hip adduction. Negative value of FPPA means knee move to valgus, RT = Right, LT = Left.

Table 6.2: Descriptive (mean ± SD) for 3D variables in each task[a]

variables	Tasks											
	FSLL		FSLLP		LSLL		LSLLP		MSLL		MSLLP	
	RT	LT	RT	LT	RT	LT	RT	LT	RT	LT	RT	LT
	Angles[a]											
Knee valgus (°)	-1.3 ± 3.9	-0.6 ± 3.9	-1.8 ± 4.2	-1.3 ± 3.7	-3.4 ± 3.3	-2.4 ± 3.5	-2.8 ± 3.6	-2.1 ± 3.7	-2 ± 5.1	-1 ± 3.5	-3.1 ± 5.5	-2.3 ± 4.2
HADD (°)	6.5 ± 4.9	5.8 ± 5.4	7.6 ± 6.3	6.6 ± 5.6	3.8 ± 6.2	5.1 ± 4.9	4.6 ± 6.4	5.3 ± 5.8	7.1 ± 5.6	6.7 ± 6.6	8.8 ± 5.3	7.2 ± 6.2
	Moments[b]											
Knee valgus	0.13 ± 0.16	0.1 ± 0.1	0.31 ± 0.24	0.2 ± 0.2	0.25 ± 0.12	0.13 ± 0.1	0.37 ± 0.23	0.14 ± 0.15	0.25 ± 0.23	0.1 ± 0.1	0.53 ± 0.34	0.24 ± 0.21
HADD	-1.4 ± 0.4	-1.6 ± 0.3	-1.6 ± 0.4	-1.9 ± 0.32	-1.4 ± 0.3	-1.8 ± 0.3	-1.6 ± 0.44	-1.9 ± 0.3	-1.3 ± 0.3	-1.6 ± 0.3	-1.6 ± 0.6	-1.9 ± 0.34
Knee extensor	2.6 ± 0.5	2.6 ± 0.51	2.95 ± 0.54	2.9 ± 0.51	2.3 ± 0.5	2.2 ± 0.43	2.7 ± 0.43	2.6 ± 0.5	2.3 ± 0.43	2.2 ± 0.6	2.9 ± 0.44	2.8 ± 0.7

[a] All values are in degrees. [b] All values are in Nm/kg. FSLL = Forward single-leg landing, FSLLP = Forward single-leg landing off a platform, LSLL = Lateral single-leg landing, LSLLP = Lateral single-leg landing off a platform, MSLL = Medial single-leg Landing, MSLLP = Medial single-leg landing off a platform, HADD = Hip adduction. A negative value for knee valgus means the knee moves to valgus, RT= right, LT = left.

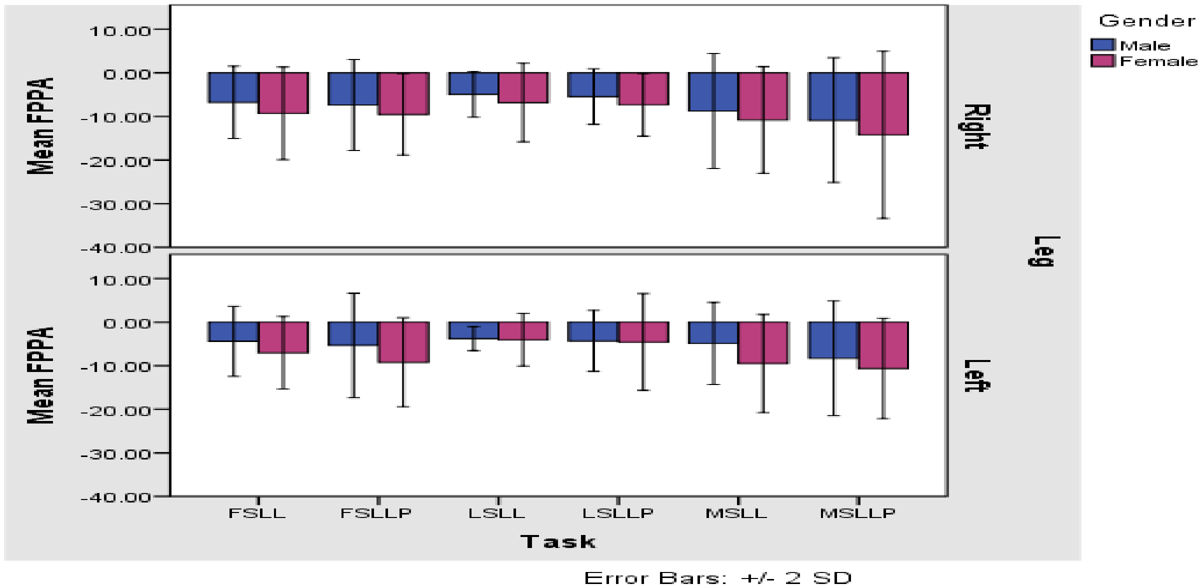


Figure 6.1: 2D FPPA during all tasks

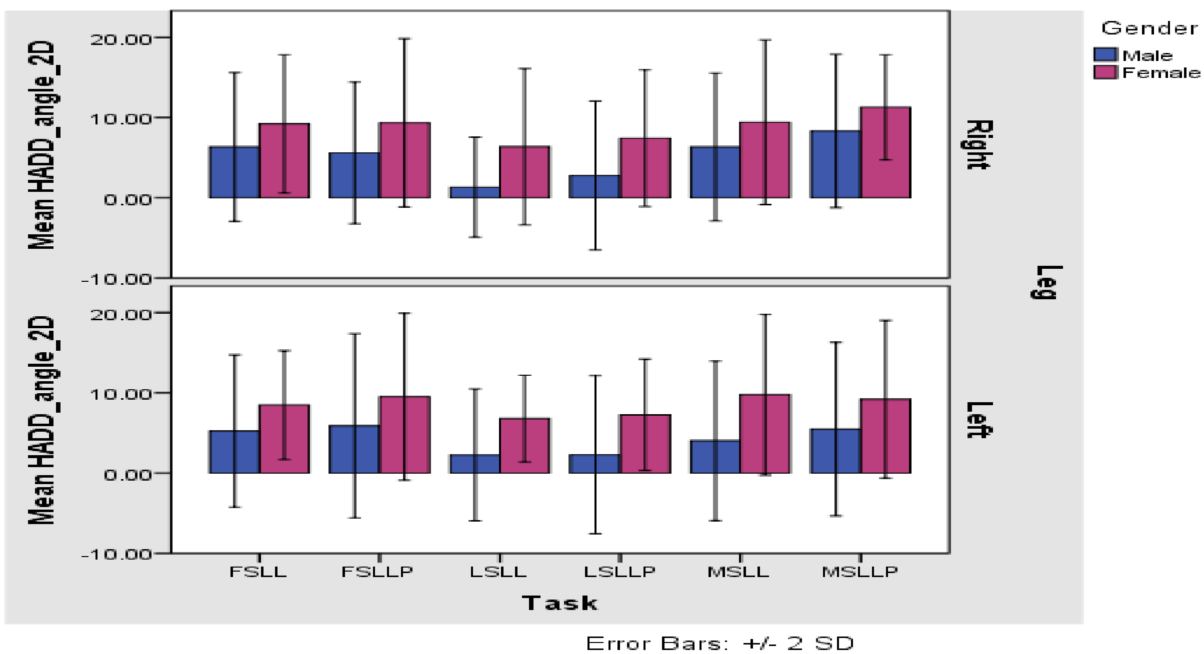


Figure 6.2: 2D hip adduction angle during all tasks

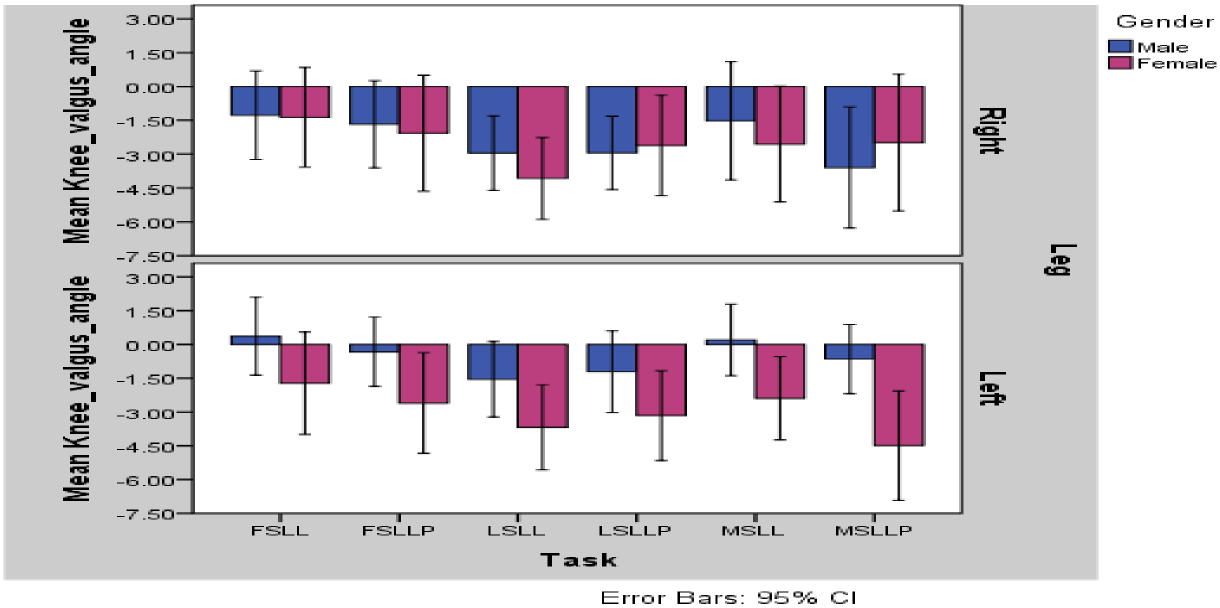


Figure 6.3: 3D knee valgus angle during all tasks

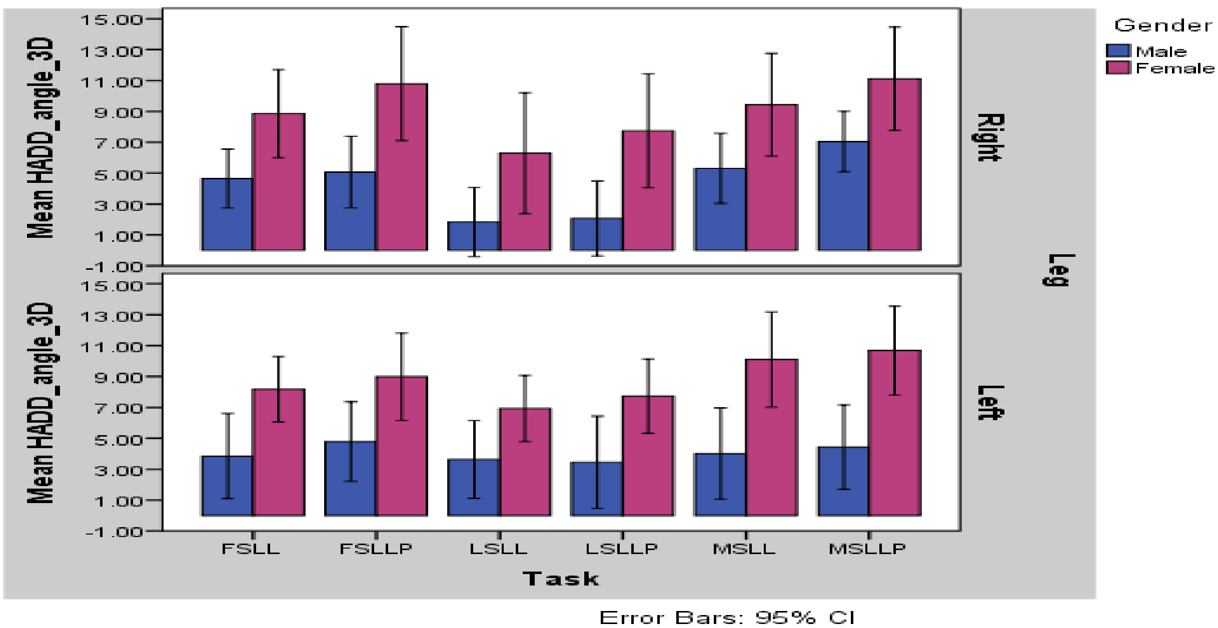


Figure 6.4: 3D HADD angle during all tasks

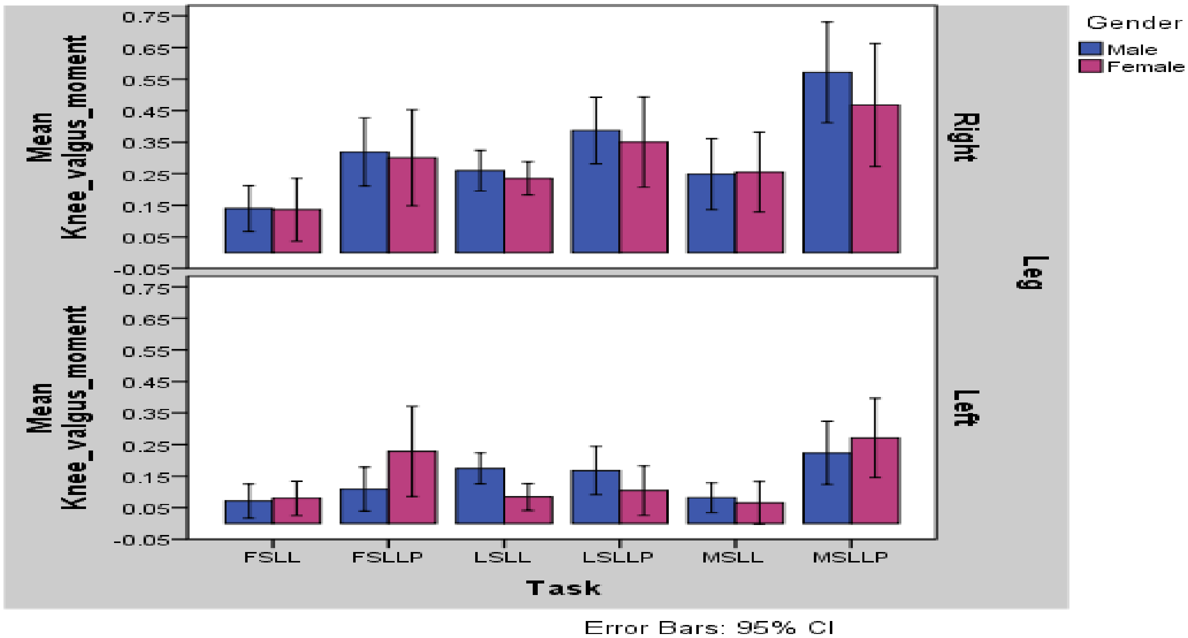


Figure 6.5: Knee valgus moment during all task

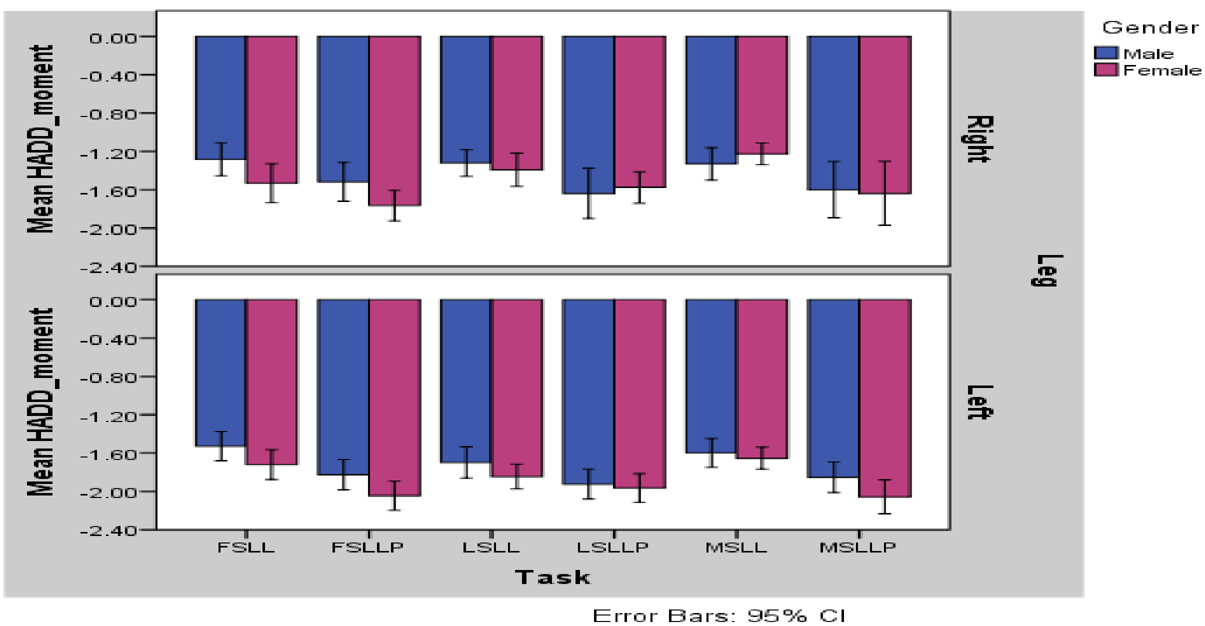


Figure 6.6: HADD moment during all tasks

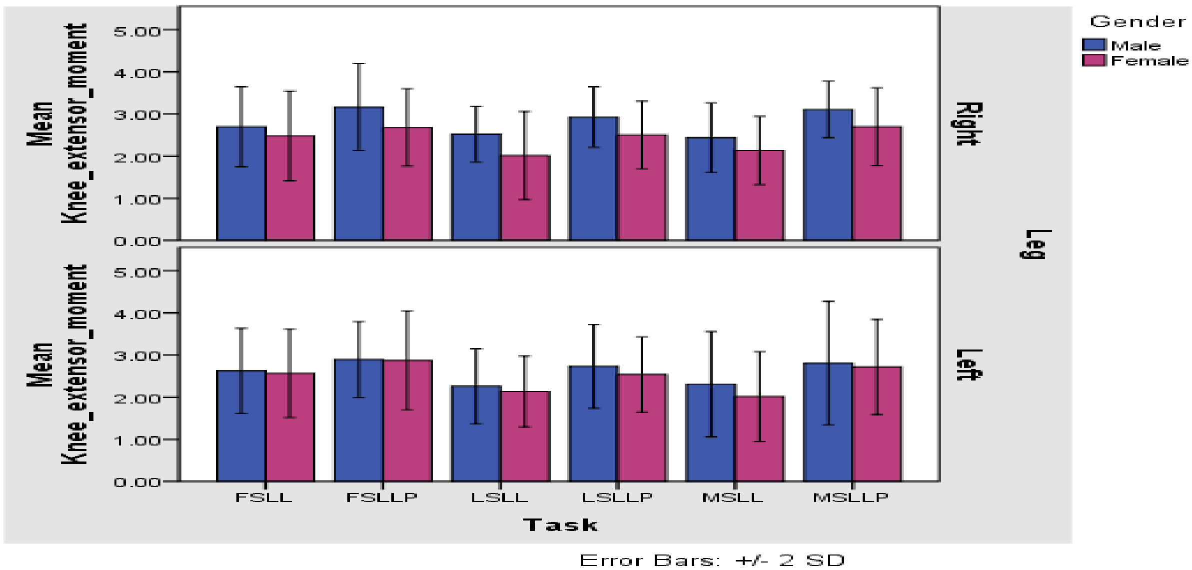


Figure 6.7: Knee extensor moment during all tasks

6.5.3 2D variables

6.5.3.1 FPPA

Tables 6.3 and 6.4 illustrate the correlation of 2D FPPA between tested tasks in both legs. Right-leg 2D FPPA during FSLL showed very strong and significant correlation with 2D FPPA during all other tasks [0.60 (r^2 = 0.35) – 0.76 (r^2 = 0.85)], apart from LSLL which reported moderate but significant correlation [0.44 (r^2 = 0.20)]. Therefore, the null hypothesis was rejected.

A very strong and significant relationship was noted between 2D FPPA during FSLLP and 2D FPPA during LSLL, LSLLP, MSLL and MSLLP (Table 6.3). The relationship between 2D FPPA during LSLL and during LSLLP was also very strong and significant (Table 6.3). Moderate but significant correlation was reported between LSLLP and MSLL. MSLL reported very strong and significant correlation with MSLLP. Other between tasks reported small to moderate correlation for FPPA (Table 6.3).

Repeated measures revealed that there were some between-tasks significant differences in right-leg FPPA ($F_{(1, 33)} = 135.368$; $P < 0.001$). Differences were found between FSLL and MSLLP, FSLLP and LSLL, FSLLP and MSLLP, LSLL and MSLL, LSLL and MSLLP, LSLLP and MSLL, LSLLP and MSLLP (Appendix XI).

Table 6.3: Between-tasks correlation (r) (P value) and linear regression analysis (r^2) for 2D FPPA (right leg)

Tasks	Tasks *r* (*P* value) (r^2)					
	FSLL	FSLLP	LSLL	LSLLP	MSLL	MSLLP
FSLL		**0.76** (< 0.001) (0.85)**	**0.44**(0.010) (0.20)**	**0.61**(< 0.001) (0.37)**	**0.68** (< 0.001) (0.46)**	**0.60**(< 0.001) (0.35)**
FSLLP			**0.57**(< 0.001) (0.32)**	**0.59**(< 0.001) (0.35)**	**0.57**(< 0.001) (0.32)**	**0.55**(= 0.001) (0.30)**
LSLL				**0.68**(< 0.001) (0.46)**	**0.**33 (= 0.055) (0.11)	0.28 (0.11) (0.08)
LSLLP					**0.40* (= 0.019) (0.16)**	0.28 (= 0.10) (0.08)
MSLL						**0.77**(< 0.001) (0.60)**
MSLLP						

FSLL = Forward SLL, FSLLP = Forward SLL off a platform, LSLL = Lateral SLL, LSLLP = Lateral SLL off a platform, MSLL = Medial SLL, MSLLP = Medial SLL off a platform. * Correlation is significant at the 0.05 level, ** Correlation is significant at the 0.01 level.

In the left leg, the correlation between 2D FPPA during FSLL and all other tasks ranged between ($r = 0.58$ ($r^2 = 0.34$) and $r = 0.77$ ($r^2 = 0.59$)), indicating a significant and very strong relationship (Table 6.4).

All other between-tasks correlations were significant (moderate to very strong) (Table 6.4).

Repeated measures revealed that there were some between-tasks significant differences in left-leg FPPA ($F_{(1, 33)} = 81.925$; $P < 0.001$). Differences were found between FSLL and MSLLP, FSLLP and LSLL, FSLLP and MSLLP, LSLL and MSLL, LSLL and MSLLP, LSLLP and MSLLP, MSLL and MSLLP (Appendix XI).

Table 6.4: Between-tasks correlation (r) (P value) and linear regression analysis (r²) for 2D FPPA (left leg)

Tasks	FSLL	FSLLP	LSLL	LSLLP	MSLL	MSLLP
FSLL		**0.70**(< 0.001) (0.48)**	**0.58**(< 0.001) (0.34)**	**0.68**(< 0.001) (0.46)**	**0.71**(< 0.001) (0.50)**	**0.77**(< 0.001) (0.59)**
FSLLP			**0.48** (= 0.004) (0.23)**	**0.40*(= 0.018) (0.16)**	**0.65**(< 0.001) (0.43)**	**0.79**(< 0.001) (0.62)**
LSLL				**0.47**(= 0.005) (0.22)**	**0.40*(= 0.020) (0.16)**	**0.53**(= 0.001) (0.28)**
LSLLP					**0.56**(= 0.001) (0.32)**	**0.59**(< 0.001) (0.35)**
MSLL						**0.74**(< 0.001) (0.55)**
MSLLP						

Column header: Tasks, *r* (*P* value) (r^2)

FSLL = Forward SLL, FSLLP = Forward SLL off a platform, LSLL = Lateral SLL, LSLLP = Lateral SLL off a platform, MSLL = Medial SLL, MSLLP = Medial SLL off a platform. * Correlation is significant at the 0.05 level, ** Correlation is significant at the 0.01 level.

6.5.3.2 HADD angle

The correlation of 2D HADD angle between tested tasks in both legs is presented in Tables 6.5 and 6.6. 2D HADD angle mostly reported very strong and significant correlation between tasks in both legs. Consequently, the null hypothesis was rejected.

For the right leg, repeated measures revealed that there were some between-tasks significant differences ($F_{(1, 33)} = 93.581$; $P < 0.001$). Differences were found between FSLL and LSLL, FSLL and LSLLP, FSLLP and LSLL, FSLLP and LSLLP, FSLLP and MSLLP, LSLL and MSLL, LSLL and MSLLP, LSLLP and MSLL, LSLL and MSLLP (Appendix XI).

Table 6.5: Between-tasks correlation (r) (P value) and linear regression analysis (r²) for 2D HADD angle (right leg)

Tasks	FSLL	FSLLP	LSLL	LSLLP	MSLL	MSLLP
FSLL		**0.73**(< 0.001) (0.53)**	**0.62**(< 0.001) (0.38)**	**0.69**(< 0.001) (0.48)**	**0.70**(< 0.001) (0.49)**	**0.64**(< 0.001) (0.41)**
FSLLP			**0.66**(< 0.001) (0.44)**	**0.70**(< 0.001) (0.50)**	**0.74**(< 0.001) (0.54)**	**0.75**(< 0.001) (0.56)**
LSLL				**0.62**(< 0.001) (0.38)**	**0.65**(< 0.001) (0.42)**	**0.53**(=0.001) (0.28)**
LSLLP					**0.65**(< 0.001) (0.42)**	**0.60**(< 0.001) (0.35)**
MSLL						**0.65**(< 0.001) (0.43)**
MSLLP						

Column header: Tasks, *r* (*P* value) (r^2)

FSLL = Forward SLL, FSLLP = Forward SLL off a platform, LSLL = Lateral SLL, LSLLP = Lateral SLL off a platform, MSLL = Medial SLL, MSLLP = Medial SLL off a platform. * Correlation is significant at the 0.05 level, ** Correlation is significant at the 0.01 level.

Left leg showed a significant difference ($F_{(1, 33)} = 64.035$; $P < 0.001$). Differences were found between FSLL and LSLL, FSLL and LSLLP, FSLLP and LSLL, FSLLP and LSLLP, LSLL and MSLL, LSLL and MSLLP, LSLLP and MSLLP (Appendix XI).

Table 6.6: Between-tasks correlation (r) (P value) and linear regression analysis (r^2) for 2D HADD angle (left leg)

Tasks	Tasks *r* (*P* value) (r^2)					
	FSLL	FSLLP	LSLL	LSLLP	MSLL	MSLLP
FSLL		**0.90**(< 0.001) (0.82)**	**0.69**(< 0.001) (0.48)**	**0.66**(< 0.001) (0.44)**	**0.65**(< 0.001) (0.43)**	**0.76**(< 0.001) (0.58)**
FSLLP			**0.64**(< 0.001) (0.41)**	**0.63**(< 0.001) (0.40)**	**0.67**(< 0.001) (0.45)**	**0.78**(< 0.001) (0.61)**
LSLL				**0.85**(< 0.001) (0.72)**	**0.74**(< 0.001) (0.55)**	**0.66**(< 0.001) (0.43)**
LSLLP					**0.73**(< 0.001) (0.53)**	**0.67**(< 0.001) (0.45)**
MSLL						**0.74**(< 0.001) (0.55)**
MSLLP						

FSLL = Forward SLL, FSLLP = Forward SLL off a platform, LSLL= Lateral SLL, LSLLP = Lateral SLL off a platform, MSLL = Medial SLL, MSLLP = Medial SLL off a platform. * Correlation is significant at the 0.05 level, ** Correlation is significant at the 0.01 level.

6.5.4 3D variables

6.5.4.1 Kinematics

All 3D kinematic variables [knee valgus (Tables 6.7 & 6.8) and HADD angles (Tables 6.9 & 6.10)] in all tasks and both legs showed very strong and significant correlation, ranging between [$r = 0.71$ ($r^2 = 0.50$)] and [$r = 0.89$ ($r^2 = 0.79$)]. Therefore, the null hypothesis was rejected.

For the right leg, repeated measures revealed that 3D knee-valgus angle showed between-tasks significant differences ($F_{(1, 33)} = 12.748$; $P = 0.001$). Differences were noted between FSLL and LSLL, FSLL and LSLLP, FSLL and MSLLP, FSLLP and LSLL, LSLL and MSLL (Appendix XI).

Repeated measures revealed that 3D left-knee valgus angle showed between-tasks significant differences ($F_{(1, 33)} = 7.315$; $P = 0.011$). Differences were noted between FSLL and LSLL, FSLL and LSLLP, FSLL and MSLLP, FSLLP and LSLL, LSLL and MSLL, MSLL and MSLLP (Appendix XI).

For the 3D HADD angle in the right leg, there were some between-tasks significant differences ($F_{(1, 33)}$ = 49.278; P < 0.001), namely between FSLL and LSLL, FSLL and LSLLP, FSLL and MSLLP, FSLLP and LSLL, FSLLP and LSLLP, LSLL and MSLL, LSLL and MSLLP, LSLLP and MSLL, LSLLP and MSLLP (Appendix XI).

For the left leg, repeated measures showed a significant between-tasks difference in the 3D HADD angle ($F_{(1, 33)}$ = 45.642; P < 0.001). Differences were only detected between LSLL and MSLLP (Appendix XI).

Table 6.7: Between-tasks correlation (r) (P value) and linear regression analysis (r²) for 3D knee valgus angle (right leg)

Task	Tasks *r* (*P* value) (*r*²)					
	FSLL	FSLLP	LSLL	LSLLP	MSLL	MSLLP
FSLL		**0.80**(< 0.001) (0.63)**	**0.86**(< 0.001) (0.73)**	**0.77**(< 0.001) (0.59)**	**0.84**(< 0.001) (0.71)**	**0.80**(< 0.001) (0.65)**
FSLLP			**0.76**(< 0.001) (0.58)**	**0.87**(< 0.001) (0.75)**	**0.72**(< 0.001) (0.51)**	**0.87**(< 0.001) (0.75)**
LSLL				**0.71**(< 0.001) (0.50)**	**0.88**(< 0.001) (0.77)**	**0.84**(< 0.001) (0.71)**
LSLLP					**0.70**(< 0.001) (0.49)**	**0.82**(< 0.001) (0.68)**
MSLL						**0.87**(< 0.001) (0.76)**
MSLLP						

FSLL = Forward SLL, FSLLP = Forward SLL off a platform, LSLL = Lateral SLL, LSLLP = Lateral SLL off a platform, MSLL = Medial SLL, MSLLP = Medial SLL off a platform. * Correlation is significant at the 0.05 level, ** Correlation is significant at the 0.01 level.

Table 6.8: Between-tasks correlation (r) (P value) and linear regression analysis (r²) for 3D knee valgus angle (left leg)

Tasks	Tasks *r* (*P* value) (*r*²)					
	FSLL	FSLLP	LSLL	LSLLP	MSLL	MSLLP
FSLL		**0.87**(< 0.001) (0.76)**	**0.90**(< 0.001) (0.81)**	**0.84**(< 0.001) (0.71)**	**0.89**(< 0.001) (0.78)**	**0.74**(< 0.001) (0.55)**
FSLLP			**0.86**(< 0.001) (0.74)**	**0.79**(< 0.001) (0.63)**	**0.85**(< 0.001) (0.72)**	**0.79**(< 0.001) (0.63)**
LSLL				**0.82**(< 0.001) (0.66)**	**0.81**(< 0.001) (0.66)**	**0.77**(< 0.001) (0.59)**
LSLLP					**0.82**(< 0.001) (0.67)**	**0.82**(< 0.001) (0.67)**
MSLL						**0.89**(< 0.001) (0.80)**
MSLLP						

FSLL = Forward SLL, FSLLP = Forward SLL off a platform, LSLL= Lateral SLL, LSLLP = Lateral SLL off a platform, MSLL = Medial SLL, MSLLP = Medial SLL off a platform. * Correlation is significant at the 0.05 level, ** Correlation is significant at the 0.01 level.

Table 6.9: Between-tasks correlation (r) (P value) and linear regression analysis (r^2) for 3D HADD angle (right leg)

Tasks	Tasks r (P value) (r^2)					
	FSLL	FSLLP	LSLL	LSLLP	MSLL	MSLLP
FSLL		**0.88**(< 0.001) (0.77)**	**0.86**(< 0.001) (0.74)**	**0.87**(< 0.001) (0.75)**	**0.88**(< 0.001) (0.78)**	**0.72**(< 0.001) (0.51)**
FSLLP			**0.84**(< 0.001) (0.69)**	**0.85**(< 0.001) (0.72)**	**0.84**(< 0.001) (0.71)**	**0.74**(< 0.001) (0.55)**
LSLL				**0.76**(< 0.001) (0.57)**	**0.85**(< 0.001) (0.72)**	**0.72**(< 0.001) (0.52)**
LSLLP					**0.81**(< 0.001) (0.65)**	**0.84**(< 0.001) (0.71)**
MSLL						**0.74**(< 0.001) (0.54)**
MSLLP						

FSLL = Forward SLL, FSLLP = Forward SLL off a platform, LSLL = Lateral SLL, LSLLP = Lateral SLL off a platform, MSLL = Medial SLL, MSLLP = Medial SLL off a platform. * Correlation is significant at the 0.05 level, ** Correlation is significant at the 0.01 level.

Table 6.10: Between-tasks correlation (r) (P value) and linear regression analysis (r^2) for 3D HADD angle (left leg)

Tasks	Tasks r (P value) (r^2)					
	FSLL	FSLLP	LSLL	LSLLP	MSLL	MSLLP
FSLL		**0.81**(< 0.001) (0.65)**	**0.89**(< 0.001) (0.78)**	**0.82**(< 0.001) (0.67)**	**0.85**(< 0.001) (0.72)**	**0.81**(< 0.001) (0.65)**
FSLLP			**0.85**(< 0.001) (0.72)**	**0.79**(< 0.001) (0.62)**	**0.75**(< 0.001) (0.55)**	**0.86**(< 0.001) (0.74)**
LSLL				**0.89**(< 0.001) (0.79)**	**0.82**(< 0.001) (0.67)**	**0.81**(< 0.001) (0.66)**
LSLLP					**0.73**(< 0.001) (0.54)**	**0.82**(< 0.001) (0.67)**
MSLL						**0.80**(< 0.001) (0.64)**
MSLLP						

FSLL = Forward SLL, FSLLP = Forward SLL off a platform, LSLL = Lateral SLL, LSLLP = Lateral SLL off a platform, MSLL = Medial SLL, MSLLP = Medial SLL off a platform. * Correlation is significant at the 0.05 level, ** Correlation is significant at the 0.01 level.

6.5.4.2 Kinetics

6.5.4.2.1 Knee-valgus moment

For the right leg, significant moderate to strong correlations were found between FSLL and all other tasks, apart from LSLL which showed a small correlation. FSLLP also reported moderate correlation with all other tasks. Strong correlation was found between MSLL and MSLLP (r = 0.73 (r^2 = 0.53) (Table 6.11).

A non-parametric test of differences among tasks rendered a Chi-square value of 49.159, which was significant (P < 0.001) (Fig. 6.8) (Appendix XI).

For the left leg, only moderate correlation, at the best, was reported between some tasks (Table 6.12).

A non-parametric test of left-leg differences among tasks rendered a Chi-square value of 25.342, which was significant (P < 0.001) (Fig. 6.9) (Appendix XI).

Table 6.11: Between-tasks correlation (r) (P value) and linear regression analysis (r²) for knee valgus moment (right leg)

Tasks	Tasks *r* (*P* value) (r^2) or (ρ) (*P* value)					
	FSLL	FSLLP	LSLL	LSLLP	MSLL	MSLLP
FSLL		**ρ = 0.41* (0.017)**	0.28(0.0115) (0.08)	**0.58**(< 0.001) (0.34)**	**0.66**(< 0.001) (0.44)**	**0.56**(0.001) (0.31)**
FSLLP			ρ = 0.33(0.055)	ρ = 0.33(0.054)	ρ = 0.31(0.07)	**ρ = 0.48**(0.004)**
LSLL				0.11 (0.549) (0.011)	0.26(0.141) (0.07)	0.29 (0.093) (0.09)
LSLLP					**0.53**(0.001) (0.29)**	**0.67**(< 0.001) (0.45)**
MSLL						**0.73**(< 0.001) (0.53)**
MSLLP						

FSLL = Forward SLL, FSLLP = Forward SLL off a platform, LSLL= Lateral SLL, LSLLP = Lateral SLL off a platform, MSLL = Medial SLL, MSLLP = Medial SLL off a platform. * Correlation is significant at the 0.05 level, ** Correlation is significant at the 0.01 level.

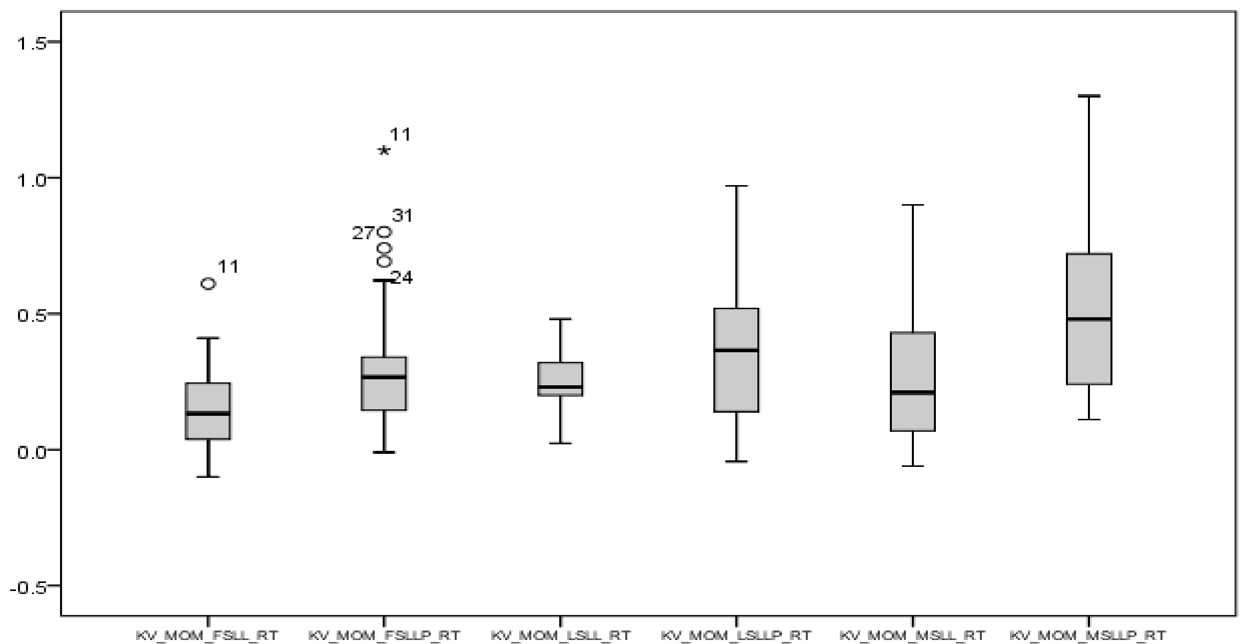


Figure 6.8: Boxplots for the difference in right leg knee valgus moment among tasks

Table 6.12: Between-tasks correlation (r) (P value) and linear regression analysis (r²) for knee-valgus moment (left leg)

Task	Tasks *r* (*P* value) (*r²*) or (ρ) (*P* value)					
	FSLL	FSLLP	LSLL	LSLLP	MSLL	MSLLP
FSLL		ρ = 0.17 (0.324)	0.030 (0.866) (0.001)	**0.47**(0.005) (0.22)**	0.17 (0.337) (0.029)	**0.49**(0.003) (0.24)**
FSLLP			ρ = -0.19 (0.284)	ρ = -0.056 (0.758)	ρ = 0.082 (0.645)	**ρ* = 0.38 (0.026)**
LSLL				**0.44**(0.009) (0.20)**	-0.092 (0.605) (0.008)	0.009 (0.961) (0.001)
LSLLP					0.16 (0.358) (0.03)	0.22 (0.211) (0.05)
MSLL						**0.45** (0.007) (0.20)**
MSLLP						

FSLL = Forward SLL, FSLLP = Forward SLL off a platform, LSLL = Lateral SLL, LSLLP = Lateral SLL off a platform, MSLL = Medial SLL, MSLLP = Medial SLL off a platform. * Correlation is significant at the 0.05 level, ** Correlation is significant at the 0.01 level.

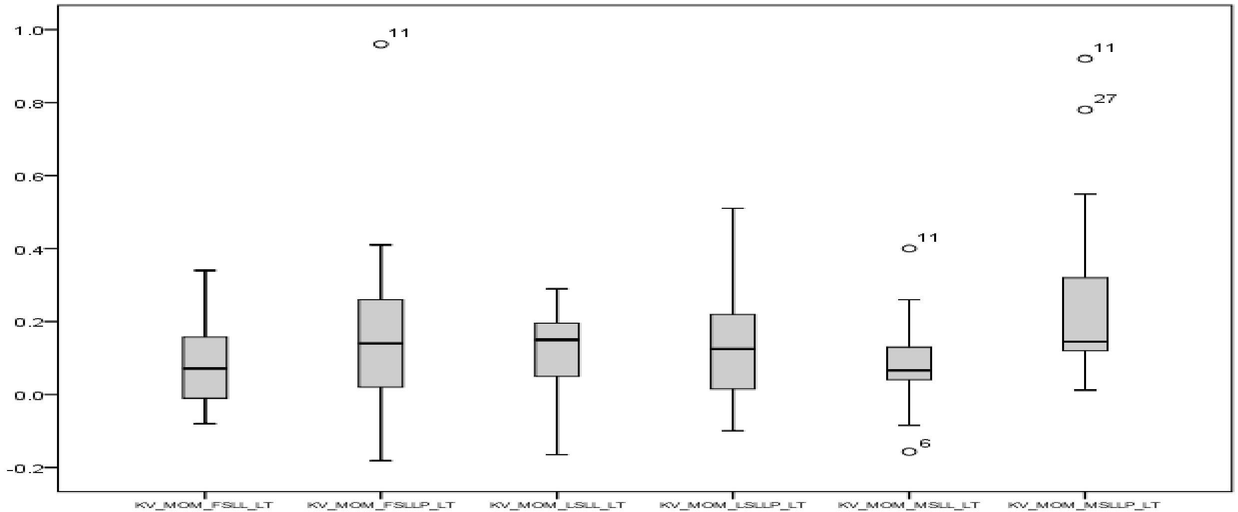


Figure 6.9: Boxplots for the differences of left leg knee valgus moment among tasks

6.5.4.2.2 <u>HADD moment</u>

HADD moment showed significant moderate to very strong correlation between all tasks in the right leg, ranging from [r = 0.53, (r^2 = 0.28)] to [r = 0.79, (r^2 = 0.62)] (Table 6.13).

A non-parametric repeated measure of right-leg differences among tasks rendered a Chi-square value of 31.425, which was significant (P < 0.001) (Fig. 6.10) (Appendix XI).

However, small to strong correlation was found between tasks in the left leg. The strongest relationship was reported between FSLL and FSLLP [(r = 0.60, (r^2 = 0.37)], while the smallest were reported between FSLL and LSLL, FSLLP and MSLLP, MSLL and MSLLP [(r = 0.30, (r^2 = 0.09)] (Table 6.14).

Repeated measures revealed that left-leg HADD moment showed some between-tasks differences ($F_{(1, 33)}$ = 2300.634; P < 0.001), specifically between FSLL and FSLLP, FSLL and LSLLP, FSLL and MSLLP, FSLLP and MSLL, LSLL and LSLLP, LSLLP and MSLL, MSLL and MSLLP (Appendix XI).

Table 6.13: Between-tasks correlation (r) (P value) and linear regression analysis (r^2) for HADD moment (right leg)

Task	Tasks *r* (*P* value) (r^2) or (ρ) (*P* value)					
	FSLL	FSLLP	LSLL	LSLLP	MSLL	MSLLP
FSLL		**0.68**(< 0.001) (0.47)**	**0.73**(< 0.001) (0.53)**	**0.53**(0.001) (0.28)**	**ρ** = 0.57 (< 0.001)**	**0.63**(< 0.001) (0.39)**
FSLLP			**0.67**(< 0.001) (0.45)**	**0.66**(< 0.001) (0.44)**	**ρ** = 0.67 (< 0.001)**	**0.78**(< 0.001) (0.61)**
LSLL				**0.56**(0.001) (0.31)**	**ρ** = 0.74 (< 0.001)**	**0.79**(< 0.001) (0.62)**
LSLLP					**ρ** = 0.66 (< 0.001)**	**0.73**(< 0.001) (0.53)**
MSLL						**ρ** = 0.71 (< 0.001)**
MSLLP						

FSLL = Forward SLL, FSLLP = Forward SLL off a platform, LSLL = Lateral SLL, LSLLP = Lateral SLL off a platform, MSLL = Medial SLL, MSLLP = Medial SLL off a platform. * Correlation is significant at the 0.05 level, ** Correlation is significant at the 0.01 level.

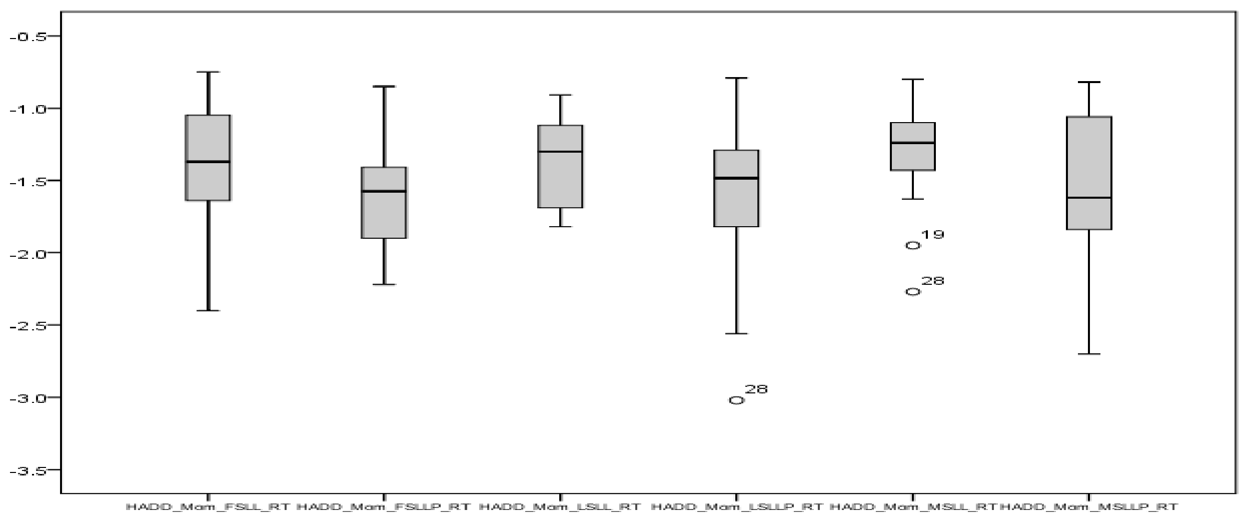


Figure 6.10: Boxplots for the differences in right leg HADD moment among tasks

Table 6.14: Between-tasks correlation (r) (P value) and linear regression analysis (r^2) for HADD moment (left leg)

Task	Tasks r (P value) (r^2)					
	FSLL	FSLLP	LSLL	LSLLP	MSLL	MSLLP
FSLL		**0.60**(< 0.001) (0.37)**	0.30 (0.086) (0.09)	**0.42*(0.013) (0.18)**	**0.49**(0.003) (0.24)**	**0.34*(0.049) (0.12)**
FSLLP			0.31 (0.071) (0.10)	**0.48**(0.004) (0.23)**	**0.41*(0.018) (0.16)**	0.30 (0.084) (0.09)
LSLL				**0.56**(0.001) (0.31)**	**0.50**(0.003) (0.25)**	0.33 (0.055) (0.11)
LSLLP					**0.44**(0.008) (0.20)**	**0.45**(0.008) (0.20)**
MSLL						0.30 (0.094) (0.09)
MSLLP						

FSLL = Forward SLL, FSLLP = Forward SLL off a platform, LSLL = Lateral SLL, LSLLP = Lateral SLL off a platform, MSLL = Medial SLL, MSLLP = Medial SLL off a platform. * Correlation is significant at the 0.05 level, ** Correlation is significant at the 0.01 level.

6.5.4.2.3 Knee-extensor moment

Between-tasks correlation for knee-extensor moment in both legs is illustrated in Table 6.15 and 6.16. Moderate to very strong correlation was reported between all tasks in both legs, ranging from [ρ = 0.38 to (r = 0.83, r^2 = 0.68)].

A non-parametric repeated measure of right-leg knee-extensor differences among tasks rendered a Chi-square value of 96.138, which was significant ($P < 0.001$) (Fig. 6.11) (Appendix XI).

Repeated measures revealed that left-knee extensor moment showed some between-tasks differences ($F_{(1, 33)} = 1066.204$; $P < 0.001$). Differences were reported between FSLL and FSLLP, FSLL and LSLL, FSLL and MSLL, FSLLP and LSLL, FSLLP and MSLL, LSLL and LSLLP, LSLL and MSLLP, LSLLP and MSLL, MSLL and MSLLP (Appendix XI).

Table 6.15: Between-tasks correlation (r) (P value, and linear regression analysis (r²) for knee-extensor moment (right leg)

Task	Tasks *r* (*P* value) (r^2) or (ρ) (*P* value)					
	FSLL	FSLLP	LSLL	LSLLP	MSLL	MSLLP
FSLL		**ρ* = 0.38 (0.028)**	**ρ** = 0.53 (0.001)**	**ρ** = 0.55 (0.001)**	**ρ** = 0.61 (< 0.001)**	**ρ* = 0.39 (0.022)**
FSLLP			**0.75** (< 0.001) (0.56)**	**0.83** (< 0.001) (0.68)**	**0.64** (< 0.001) (0.41)**	**0.78** (< 0.001) (0.60)**
LSLL				**0.79** (< 0.001) (0.63)**	**0.68** (< 0.001) (0.46)**	**0.76** (< 0.001) (0.58)**
LSLLP					**0.66** (< 0.001) (0.44)**	**0.82** (< 0.001) (0.67)**
MSLL						**0.60** (< 0.001) (0.36)**
MSLLP						

FSLL = Forward SLL, FSLLP = Forward SLL off a platform, LSLL = Lateral SLL, LSLLP = Lateral SLL off a platform, MSLL = Medial SLL, MSLLP = Medial SLL off a platform. * Correlation is significant at the 0.05 level, ** Correlation is significant at the 0.01 level.

Table 6.16: Between-tasks correlation (r) (P value) and linear regression analysis (r²) for knee-extensor moment (left leg)

Task	Tasks *r* (*P* value) (r^2)					
	FSLL	FSLLP	LSLL	LSLLP	MSLL	MSLLP
FSLL		**0.66** (< 0.001) (0.44)**	**0.77** (< 0.001) (0.60)**	**0.67** (< 0.001) (0.45)**	**0.81** (< 0.001) (0.66)**	**0.62** (< 0.001) (0.39)**
FSLLP			**0.57** (< 0.001) (0.33)**	**0.49** (0.003) (0.24)**	**0.67** (< 0.001) (0.44)**	**0.47** (0.005) (0.22)**
LSLL				**0.74** (< 0.001) (0.54)**	**0.84** (< 0.001) (0.70)**	**0.75** (< 0.001) (0.56)**
LSLLP					**0.71** (< 0.001) (0.51)**	**0.77** (< 0.001) (0.59)**
MSLL						**0.72** (< 0.001) (0.51)**
MSLLP						

FSLL = Forward SLL, FSLLP = Forward SLL off a platform, LSLL = Lateral SLL, LSLLP = Lateral SLL off a platform, MSLL = Medial SLL, MSLLP = Medial SLL off a platform. * Correlation is significant at the 0.05 level, ** Correlation is significant at the 0.01 level.

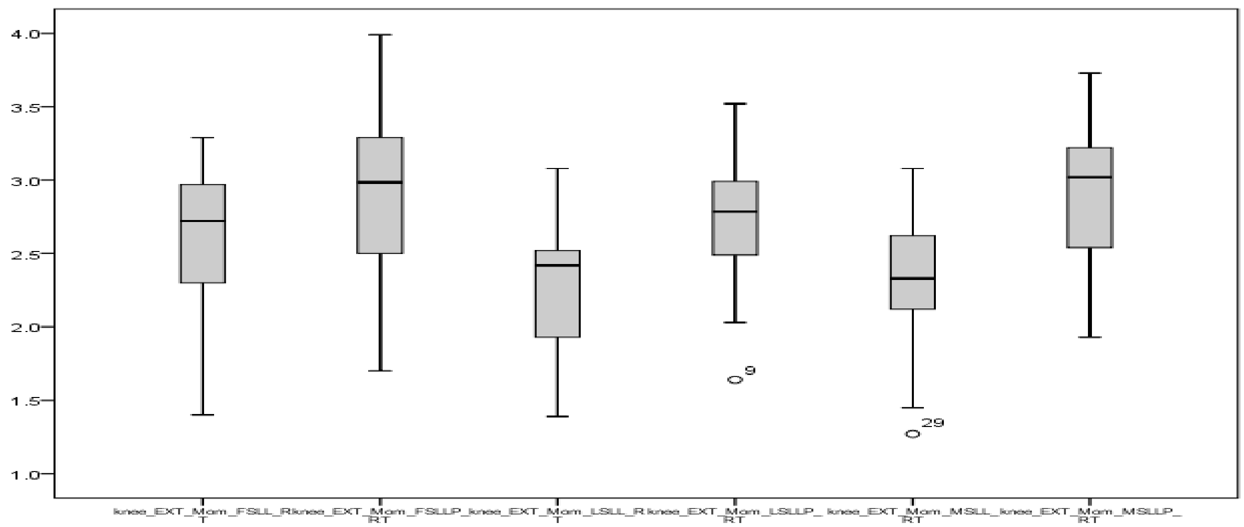


Figure 6.11: Boxplots for the differences of right leg knee extensor moment among tasks

6.6 Discussion

The use of functional tasks is a common tool to examine lower-extremity biomechanics and how those can be used in both the prevention and prediction of injuries. Examining the differences and similarities between these different functional tasks became the focus of many researchers, as the identification of these may help in understanding how body segments behave during different tasks, how they associate with injury, and then determine which tasks can be used to predict injury or used in certain stages of rehabilitation programmes according to their difficulty.

SLL in different directions and from different heights is commonly used as a screening tool and a rehabilitation exercise. It is, however, unknown whether SLL in one direction is biomechanically similar to SLL in a different direction. This is a very important point as it answers several questions: do clinicians actually need to use all these tests or can one of them reflect or predict others? If not, which tests are biomechanically the most demanding and which are less demanding? Knowing this will allow employment of the right tests in the appropriate stages during the progression of a therapeutic programme. Therefore, this study was conducted to examine if there is a relationship between the characteristics of biomechanical variables during multidirectional SLL using both 2D and 3D motion analysis systems.

Most of the studies that have compared the biomechanical characteristics of different functional tasks compared either double-leg tasks with single-leg tasks, or double-leg tasks with double-leg tasks. Given the different nature of single- and double-leg tasks, it seems difficult to make comparisons with the current study. Any single-leg task, such as single-leg drop landing and SLL, has been described as “more challenging” because the load of the body is shifted onto one leg. Such a task has shown greater frontal plane ROM, angles, moments, GRF and energy dissipation, and decreased knee-flexion angle, compared to double-leg landing. A single-leg task is mostly performed in a high-speed manner. The decreased base of support involved in such task makes it more dangerous, as the demands of impact absorption increase on the one leg’s muscles (Yeow, Lee, & Goh, 2011; Pappas et al., 2007; Ford et al., 2005). It also matches the real situation of injury, which mostly occurs during the deceleration phase of landing on one leg (Olsen et al., 2004; Boden et al., 2000; Kirkendall & Garrett 2000). Therefore, it is important to compare between the different single-leg tasks that are commonly seen in sport and usually used as screening or rehabilitation tools.

A limited number of studies have compared between two or more single-leg tasks. For example, Ortiz et al. (2011) compared between side hopping and crossover hopping. McLean et al. (2005) compared between side step and side jump, Whatman et al. (2011) examined single-knee bending, small single-knee bend, hop lunge and step down.

Moderate to strong correlation was reported for some of the variables in the aforementioned studies. However, McLean et al. (2005) and Whatman et al. (2011) only examined kinematics. Whatman et al. (2011) examined the dominant leg in most of the tasks they investigated, while it is better to examine both legs as injury can occur in both and performance might be different between legs. Moreover, the nature of single-leg squatting and side stepping examined in both studies might be not be comparable to SLL, particularly with regard to the speed at which they were being performed and how they decelerated, which makes comparison with the current study difficult, as the nature of all tasks in the present study was that they were to be performed in a high-speed manner and with quick deceleration.

The findings of the current study are comparable the previous work. For example, the value of knee-valgus angle during MSLLP is similar to that found by Marshal et al. (2015) during hurdle hops. HADD angle during FSLL and MSLLP are nearly the same as those found during SLL (Pappas et al., 2007), crossover hops and side hops (Ortiz et al., 2011; Alenizi et al., 2014), respectively. Similar, HADD moment during most of the tasks is consistent with the findings of Alenizi et al. (2014). However, knee-extensor moment has not been examined before in single-leg tasks, which makes comparison difficult.

6.6.1 2D variables

This study found that the values of FPPA and HADD were greatest during FSLLP, MSLL and MSLLP, and lowest during LSLL and LSLLP for both legs, indicating that FSLLP, MSLL and MSLLP might be more challenging and demanding than other tasks, because FPPA and HADD angle were greater with these tasks, suggesting greater ACL loading (Imwalle et al., 2009). Such findings are unsurprising, given how these tasks were performed. During lateral landing, the foot was placed away from the midline of the body, which resulted in increased hip abduction and decreased hip adduction (Dempsey et al., 2009). HADD is considered to be a component of knee-valgus position, and if this decrease, it leads to a decrease in valgus angle (Mascal et al., 2003). The opposite would occur during medial landing.

The results also show that 2D FPPA and HADD angle during FSLL had very strong correlation with all other tasks, and 2D FPPA and HADD angle in any task performed without a platform mostly correlated significantly with the same task performed with a platform in both legs, with up to 20–85 per cent of the variance being explained by FSLL (Table 6.3–6.6.6). This suggests that lower-limb kinematics during SLL tasks may not be affected by the direction and height of landing (30 cm), as the same changes that occur during FSLL may occur during other tasks. However, this may not apply to heights other than 30 cm. Atkin et al. (2014) examined the relationship between SLS and SLL. Despite the differences in the tasks examined, they found that 2D FPPA during SLS correlated moderately with 2D FPPA during SLL (r = 0.35). This correlation became strong for females when participants were analysed by gender (r = 0.87). Although both tasks are performed on one leg, SLS is different from the tasks examined in the current study, as it is not performed in a high-speed manner, leading to subjects controlling the movement in a way that does not match what actually happens during sport activity. Moreover, Atkin et al. (2014) only examined eight women. Therefore, stronger correlation may not represent that of a larger population. Munro et al. (2017) found that 2D FPPA significantly correlated during SLS, SLL and drop jump. However, none of the aforementioned studies examined HADD angle.

The present study's findings suggest that individuals who exhibit larger FPPA or HADD angle in any of the without-platform tasks are likely to exhibit similarly when the tasks are performed with a platform. They also suggest that if SLL is performed from a 30 cm height, it does not make a big difference with regard to the 2D lower-extremity angle. Therefore, when examining subjects to assess the angle of lower extremities using 2D, FSLL would be enough to give a good picture of movement around the knee joint. Doing other tasks might be unnecessary, particularly when taking into account the time required for the subject and therapist, as FSLL can explain up to 80 per cent of other tasks' performances. This is helpful in saving time for the clinician and patients.

The lack of significant correlation between, for example, LSLL and MSLL might be due to the 2D camera position, which was placed perpendicular to the frontal plane of motion during the calibration process. The subject was in a static position. When the subject moved laterally, the camera angle of vision remained still, which may affect the results. Additionally, the load on the knee and hip joints during LSLL might be more than the load during other directions of landing due to trunk movement towards the side of the leg that was being landed on during LSLL. Medial and lateral movement of the trunk has been reported to influence frontal-plane moment (Powers, 2010).

6.6.2 3D variables

6.6.2.1 Kinematics

The results of the current study reveal that inter-tasks have very strong and significant relationships with all 3D kinematic variables [knee valgus (Tables 6.7 & 6.8) and HADD angles (Tables 6.9 & 6.10)] in both legs, with 50–77 per cent of the variance being explained by FSLL. Although the relationship between different single-leg tasks has not been examined before, such findings are consistent with the findings of previous studies. Harty et al. (2011) found that knee-valgus angle correlated significantly during step down, SLL and DVJ (r = 0.72 - 0.76). However, they only examined female athletes. Whatman et al. (2011) reported that hip and knee kinematics during SLS, DLS, lunge, hop-lunge and step-down were found to have moderate to strong correlation with jogging. Recently, Donohue et al. (2015) examined the kinematics and kinetics of 34 subjects, half of them females, to explore the correlation between 3D variables during SLL, DLL, SLS and DLS. They found that hip- and knee-abduction angles significantly related between SLL and DLL, and between SLS and two types of landing ($r \geq 0.54$). The good reliability that was found for these variables in study two of the present project (Tables 4.9–4.14) may play an important role in this strong correlation. The findings of the present study suggest that individuals are likely to show similar profiles for injury risk when screened using the examined functional tasks, which show similar knee and hip kinematics as well. Therefore, any one of the examined tasks might be enough when clinicians aim to assess the kinematics of the hip or knee using a 3D motion analysis system. This can save time for both clinicians and subjects, which in turn allows more subjects to be screened. Linking this finding with the findings of Harty et al. (2011) and McLean et al. (2005), there is growing evidence to suggest that individuals who show greater dynamic knee valgus may do so across a wide range of other tasks. However, the existence of such inter-task correlation in real situations of competition or practice is still unknown and should be considered in future research.

6.6.2.2 Kinetics

6.6.2.2.1 Knee-valgus moment

Knee-valgus moment has been suggested as predicting ACL injury (Hewett et al., 2005). It is also reported to be a component of knee-joint loading, which is sensitive to neuromuscular control variation (McLean et al., 2004). Markolf et al. (1995) also suggest that knee-valgus moment increases the load on the ACL and the risk of injury, particularly when combined with anterior tibial force in a flexion position. The present study's results reveal that some significant moderate to strong correlation was found in knee-valgus moment between some of the tasks (Tables 6.11 & 6.12) with 54 per cent of the variance being explained (at best). However, there was no correlation between most of the tasks. This suggests that a clinician should employ all these tasks in any rehabilitation or screening programme when the target is knee-valgus moment. The lack of significant correlation for this variable could be attributed to several reasons. Given the examined population and the tasks examined, it can be seen that the tasks are highly demanding, because the load of the body is managed by only one leg's musculature, while the participants were moderately active and the types of sport they participated in were not controlled. The sports they practised may not include tasks such as those examined in the current study, which may affect the findings. More importantly, trunk movement was uncontrolled, thus matching the real situation of landing as the individual lands without any instruction. However, it may have led to a lack of significant between-tasks correlations because of inconsistent of relative trunk position. When individuals performed lateral or medial landing, the trunk was observed to move in the direction of the leg that was being landed on, which, in turn, may increase the load on the leg and lead to change in moments. Greater correlation might be obtained if the sport type and trunk motion were controlled. This may explain the small negative correlation found for some tasks in both legs (Tables 6.11 & 6.12). Going back to Tables 4.9–4.14, it is also clear that the repeatability of knee-valgus moment during most of the tasks was moderate. This may also affect the results, as a lack of performance consistency during examined tasks may limit the ability to find correlation between variables during the performance of these tasks.

Such findings illustrate different demands on the knee when there is a change in the direction of landing, which may partially explain the increase of ACL injury in soccer, as it involves many tasks that include different directions (Jones et al., 2014; Faude, Junge, Kindermann, & Dvorak, 2005). However, strong correlation was reported between some tasks e.g. FSLL correlated significantly with FSLLP [(r = 0.49, (r^2 = 0.25)]. This was expected, as both tasks were in a forward direction, where the effect of trunk motion might be minimal.

The results from previous literature are conflicting. For instance, Jones et al. (2014) found no correlation between knee-valgus moment with SLL or pivoting; Donohue et al. (2015) found that knee-valgus moment correlates between single- and double-leg landing, and between SLS and single- and double-leg landings. Harty et al. (2011) found a moderate relationship between knee-valgus moment during step down, SLL and DVJ. The differences between studies might be due to the sample populations, difficulty of tasks and level of subject performance or experience to practise the examined tasks. For example, Harty et al. (2011) examined 37 female athletes (age 19.5 ± 1.2 years, mass 74.6 ± 7.8 kg, and height 1.73 ± 0.09 m) who were participating in sports that include jumping, cutting or pivoting and had an average of 10.4 ± 3.1 years of experience in their respective sports. This may be an advantage when compared to the participants of the current study who were moderately active but with no specific type of sport (age 28. ± 3.9 years, mass 67.7 ± 7.9 kg). However, the height was nearly the same (1.7 ± 0.05).

6.6.2.2.2 HADD moment

HADD moment has been reported previously as being one of the most dangerous biomechanical deviations. Its increase leads to difficulty in resisting adduction, which in turn leads to dynamic-valgus collapse in the knee, which increases the risk of ACL injury (Imwalle et al., 2009). Several researchers have also linked HADD moment to ACL injury and PFPS during different sporting tasks (Souza & Powers, 2009; Hewett et al., 2005). Most of the studies that have compared tasks focused only on knee-valgus moment (Kristianslund & Krosshaug 2013; Donohue et al., 2015). To the best of the researcher's knowledge, only one study has examined the correlation between SLL and cutting with regard to HADD moment (Jones et al., 2014). The results of the current study indicate that HADD moment showed significant strong correlation between all tasks in the right leg (Table 6.13). This implies that one task can explain 39–62 per cent of the HADD moment occurring during others. Although such tasks have not been examined before, the results support the findings of previous studies that examined the correlation between different tasks. Jones et al. (2014) reported moderate correlation between SLL and cutting (r = 0.46), though it was statistically significant. The authors concluded that females who exhibit poor SLL biomechanics will exhibit the same during other changing-direction tasks but SLL cannot replace other tasks. The greater correlation in the current study might be due to the nature of the examined tasks, as all of them were single-leg tasks that can be performed with similar speed and power, while Jones et al.'s (2014) study compared tasks which might generally be performed in a different manner.

With regard to the left leg, significant moderate to strong correlation was observed between some of the examined tasks (table 6.14) with, at best, 37 per cent of FSLLP variance being explained by FSLL. The other tasks showed no significant correlation (Table 6.14), indicating limb asymmetry in hip-adduction moment. The disparity between legs may be attributed to several reasons. In the reliability study of this project (Tables 4.9–4.14), the ICC for HADD moment in most of the tasks was moderate, particularly between days and for the left leg, while it was greater for the right leg. This may influence correlation in the left leg. Moreover, the dominant leg may also influence the findings as it has been found to offer more postural support and stability (Decker et al., 2003) and to have significantly larger hip-abductor muscle strength (81 ± 23.7 Nm) than the non-dominant leg (76 ± 9.9 Nm) (Jacobs, Uhl, Seeley, Sterling, & Goodrich, 2005). Consequently, examining the limb symmetry of HADD moment should be considered in future work.

6.6.2.2.3 Knee-extensor moment

Compared to healthy controls, ACL female patients have demonstrated larger knee-extensor moment during side-to-side hopping (Ortiz et al., 2011), suggesting an association between injury and knee-extensor moment. In vivo and in vitro studies suggest that the quadriceps muscle can generate enough anterior tibial translational to injure the ACL (Griffin et al., 2006; Boden et al., 2000; DeMorat et al., 2004). High-knee extensor moments, accompanied by small-knee flexion, are important mechanisms of non-contact ACL strain (Boden et al., 2000; Olsen et al., 2004). However, how knee-extensor moment changes with knee flexion during sudden deceleration tasks such as landing is still unknown (Podraza, & White, 2010). Moreover, the correlation of knee-extensor moment with different functional tasks has not been studied extensively, particularly during single-leg tasks. To the best of the author's knowledge, only one study has examined differences in knee-extensor moment between single- and double-leg stop-jump tasks (Wang, 2011).

The result indicate that knee-extensor moment in FSLL shows significant moderate to very strong correlation with all tasks in both legs, with 31–70 per cent of variance in one variable being explained by the other variable. This indicates that individuals perform these tasks with similar patterns of extensor moment. Therefore, when examining knee-extensor moment, FSLL would be suitable to explore the moment occurring during the majority of other tasks. For instance, if a patient is involved in a rehabilitation programme aiming to alter knee-extensor moment or knee-extensor strength, the changes in these parameters during FSLL will be the same as during other tasks. Consequently, there is no need to examine all the tasks, which can help save time in an assessment session for both clinician and patient. Wang (2011) examined the differences in knee-extensor moment between single- and double-leg stop-jump tasks and found that knee-extensor moment was significantly greater during a single-leg stop-jump task. However, the relationship between these tasks was not examined in Wang's (2011) study.

This study is not without its limitations. This study only included moderately active participants. It is still unknown if some correlation might exist in an elite athletic or injured population;

therefore, application of the findings to other populations should be pursued with caution. Also, the examination was conducted in a laboratory where the movement may be unique, so it seems unlikely that such unique movement tasks may differ during practice or competition, which means such correlation between tasks may no longer apply. While the current study represents the most used clinical tasks, it does not include all varieties of tasks. Last, investigation of all potential factors of why there is correlation between tasks was beyond the scope of this study. It is possible that factors other than biomechanical alignment, such as muscle strength and knee laxity, could be important to consider. So, future studies to validate these results by understanding the underlying neuromuscular factors that cause similarities in performance between tasks should be conducted.

6.7 Conclusion

The present study found significant (moderate to very strong) relationships in lower-extremity 2D and 3D biomechanical variables between FSLL, FSLLP, LSLL, LSLLP, MSLL and MSLLP, confirming the study hypothesis. The findings suggest that moderately active people who exhibit poor FSLL biomechanics may exhibit the same during other SLL direction tasks. This provides additional support for the use of landing as a rehabilitation exercise and a screening test for injury, particularly when the injury mechanism is mainly during unilateral loading tasks. However, it should be addressed that when examining knee-valgus moment, individuals may exhibit different profiles and thus other directions, such as MSLL, should be employed.

7. Chapter 7: Summary, conclusion, suggestions for future work and clinical implications

7.1.1 Summary

Lower-limb biomechanics during functional tasks has been examined in different studies. Some of those studies used bilateral tests, which prevent~~s~~ comparison between the sound and affected legs. Such comparison was possible with tests that require only one leg to be completed, as the sound leg can be used as a control while quantifying the function of the affected leg. Landing in sport mostly occurs unilaterally, which makes up about 70 per cent of non-contact ACL injuries. The studies that examined single-leg tasks focused mainly on the sagittal plane, while the frontal plane of movement is important because the movement of position suggested as being associated with non-contact ACL injury mainly occurs in the frontal plane, such as knee valgus and hip adduction. Thus, examining the biomechanics of the frontal plane during a unilateral task is important to understanding how individual joint biomechanics respond to meet sport demands.

An SLL test is a functional performance test, it is commonly used in both research and clinical practice to evaluate the dynamic stability of the lower extremities, particularly the knee joint (Dos Reis et al., 2015). It is also an important screening tool that can be used to identify those who are at risk of lower-extremity injury and to evaluate the progress of a rehabilitation regime for individuals with ACL injury or PFPS. A number of studies have investigated the biomechanics of SLL. However, these studies mainly examined SLL in one direction only (forward) while sport demands involve multidirectional landing; consequently, multidirectional SLL needs to be examined. Both 2D and 3D motion analysis are important as each one has its uses and advantages. Therefore, both methods should be considered when examining lower-limb biomechanics to understand how reliable these methods are and how 2D close is to the gold standard of 3D, which may fill the gap between clinical and research environments.

Comparison of biomechanics among athletic tasks can explain the characteristics of these tasks and help in the identification of those tasks which pose a risk of injury. This, in turn, could help in the prevention and treatment of injury. A few studies have compared tasks in terms of biomechanical characteristics, particularly frontal-plane kinematic and kinetics. Most of these studies examined a small sample of females only and did not calculate a coefficient of determination (r^2), which is important when conducting a correlation study. Moreover, most of the studies examined the correlation between double- and single-leg tasks. The possibility of a relationship between a single leg-task and a double-leg task might be limited due to the difference in nature of the tasks. There are many tasks that are seen in the sporting environment and used as screening tools and rehabilitation exercises. Many of these tasks have not been covered in the literature, particularly multi-directional single-leg landing tasks which have been linked to non-contact knee injury. Therefore, examining the correlation between different types of single-leg tasks should be done in order to better understand the causes and contributing factors that may lead to lower-extremity injury, and to understand how individuals use joint biomechanics to meet the demands of these sport tasks.

Given the limitations of the available literature, the aims of this thesis were:

1. To systematically review the available literature investigating the biomechanics of the lower-extremity frontal plane of motion during multidirectional SLL.

2. To examine the reliability of using 2D motion analysis system to measure lower extremity kinematics during multidirectional SLL.

3. To examine the reliability of using 3D motion analysis system to measure lower extremity kinematics during multidirectional SLL.

4. To examine the validity of 2D motion analysis in measuring lower extremity frontal plane kinematics during multidirectional SLL in comparison to findings from 3D motion analysis system.

5. To examine the relationships between biomechanical characteristics during multidirectional SLL tasks using both 2D and 3D motion analysis.

7.1.2 Conclusion

The first aim of this thesis was addressed by conducting a systematic review of the available literature investigating the frontal-plane biomechanics of the lower extremities during multidirectional SLL. It was found that only SLL in a forward direction was tested in the majority of the literature, using 3D motion analysis only, indicating the importance of examining other directions of SLL using both 2D and 3D motion-analysis systems.

Regarding the second and third aims, within-day and between-days reliability and establishing SEMs for lower-extremity biomechanical variables using 2D and 3D motion analysis during multidirectional SLL were examined. Generally, within-day ICCs were greater than between-days ICCs. Yet, the majority of 2D and 3D variables showed good to excellent reliability (ICCs 0.61–0.98) with SEMs (0.63°–6.6°) of kinematic variables and (0.01–0.34 Nm/Kg) kinetic variables, indicating that 2D and 3D variables during multidirectional SLL are reliable and reproducible within and between days when examined by the same rater and can be used with confidence when measuring lower-extremity biomechanics following the measurement instructions explained in section 4.4. Clinicians can also can employee SEM values to accurately evaluate whether changes in biomechanics are actual changes in performance or not, as well as whether differences between legs or individuals are greater than measurement errors. However, knee-valgus moment and hip-adduction moment were less reliable among all the tasks. Possible reasons for this might be between-subject variability, the presence of fatigue due to the numbers of tasks and trials being performed, and trunk movement, which may place high stress on the muscles and joints and shift the centre of mass to the landing limb, which may influence the moment.

The fourth aim was to investigate the correlation between 2D and 3D motion-analysis techniques when measuring the lower-extremity frontal plane of movement during multidirectional SLL. It was found that 2D FPPA, at best, moderately correlated with 3D knee valgus in FSLL, LSLL and MSLL, with 17 per cent of 3D variance being explained by 2D, while 2D HADD angle showed a strong and significant correlation with 3D HADD angle (ranging between $r = 0.70$ and $r = 0.90$) and 49–81 per cent of 3D variance being explained in all tasks, apart from the right leg during MSLLP, which reported only a small association ($r = 0.27$), indicating the possible clinical use of 2D motion analysis, which suggests that 2D is a cost-effective alternative to 3D when measuring hip angles. When measuring knee angles, it can be used as an acceptable proxy for an expensive 3D motion-analysis system. Yet, it should be used with some caution when measuring knee angles. In addition to its reliability, the ability of 2D HADD to detect 3D HADD without a highly equipped laboratory can help practitioners to identify at-risk athletes. So, it is highly recommended for injury-risk assessment.

The lower correlation found for knee angles might be due to the fact that knee-valgus angle is a combination of movements that include rotational movement. This rotational movement cannot be calculated via 2D motion capture. Moreover, movement of the knee from side to side may result in different times when peak 2D FPPA and 3D knee valgus occur during the landing cycle. Furthermore, the accuracy of estimation of the time when peak 2D FPPA corresponds to peak 3D knee valgus angle might be insufficient.

Regarding the last aim, which was to investigate the relationship between selected biomechanical variables among a battery of SLL tasks using both 2D and 3D motion analysis techniques. the findings of this study reveal that 2D and 3D variables, apart from knee=valgus moment, report significant moderate to very strong correlation among all examined tasks, suggesting that moderately active people who exhibit poor FSLL biomechanics are likely to perform the same during all other directions of SLL. From this it can be concluded that one of these tasks can reflect the others, suggesting that one of these tasks might be enough when intending to measure lower-limb biomechanics and a clinician does not need to test all these tasks. The use of different directions of SLL would probably not add that much additional benefit to give extra information about a biomechanical profile of the lower limbs. Therefore, using one direction of SLL along with other tests may show different biomechanical characteristics (such as strength) and might be more feasible to implement in any rehabilitation programme or screening protocol. However, when a clinician intends to evaluate knee-valgus moment, other SLL directions should be employed as they can demonstrate different profiles. Some reasons that might be behind the lack of significance in knee-valgus moment correlations, such as trunk movement, were not controlled for in this study. Additionally, the lack of performance consistency found in the reliability study for this variable during the examined tasks may limit the ability to find correlation between variables during performance of these tasks.

7.1.3 Suggestions for future research

While working on this project and based on the findings, several questions have arisen which might be considered in future research. From the reliability study (chapter 4), it is recommended that biomechanical variables during multidirectional SLL should be considered reliable in future work. However, this applies to one rater and it is not known if similar reliability would be obtained by different evaluators, so between-raters reliability may be considered in the future as this better represents a clinical situation where patients may be examined by different therapists. Future studies examining other populations, such as athletes, including different sports, different levels of participation and injured populations, are also needed to understand the biomechanical differences between them. Also, future work could look at the variability between individual performances and its influence on biomechanical characteristics. Examining participants in an actual sporting background is also recommended. Moreover, the findings of the reliability study suggest the importance of using the CAST model to measure lower-extremity biomechanics in future research. However, studies using full-body 3D models are also needed to investigate trunk-position changes as the CAST model does not take these into account.

Considering the correlation findings between 2D and 3D motion-analysis systems (chapter 5), future research on larger populations and during different sporting tasks is needed. The focus of the current study was on the frontal plane of movement because the suggested position of injury mainly occurs in the frontal plane. However, including the sagittal plane in future studies may add important information to the literature. To expand the generalizability of using 2D motion analysis in clinics, populations commonly seen in clinics should be examined in future studies, such as lower-extremity injured individuals.

Although the findings of the inter-tasks correlation study (chapter 6) could be taken as reference values for these tasks, as they have not been examined before, it should be noted that these are not normative data. A future large-scale study is recommended to look at normal values for the variables in these tasks. Moreover, this study is a correlation study, it does not establish any cause-and-effect relationship. Future research could examine the effects of injury-prevention programmes to see whether changes to some of these tasks could be transferred to other tasks or not. The SEMs reported in the reliability study (chapter 4) would allow researchers to determine whether changes in biomechanical characteristics are due to intervention or measurement errors. Also, a large prospective study applying these measurements and tracking injuries is recommended to identify those who are at risk of knee injury during these tasks. Moreover, only healthy, moderately active individuals were included in this study, further studies could explore these relationships in either injured or uninjured athlete populations. Finally, the findings of this project do not consider fatigue situations that may occur during actual practice or competition. It is well known that injury may occur during any time of play, particularly when fatigue is present. Therefore, future work looking at the performance of these tasks pre- and post-fatigue is recommended.

7.1.4 Clinical implications

The findings of this project have several clinical implications. Multidirectional SLL can be used reliably in research and clinical fields using 3D and 2D motion analysis, respectively, to assess the biomechanical characteristics of the lower extremities. In clinics where there is no access to a 3D motion analysis system, 2D can be a suitable alternative to evaluate lower-extremity biomechanics, particularly the hip joint, during multidirectional SLL. The strong correlation observed between tasks suggests that individuals may demonstrate similar profiles of injury risk when screened using multidirectional SLL. This suggests that using FSLL may represent other directions, which eliminate the need for using different directions of SLL. Also, individuals with poor alignment during FSLL are likely to show poor alignment in other tasks. Consequently, employing other tasks that have been utilized in previous studies, such as double-leg landing (Hewett et al., 2005) along with FSLL, may be more beneficial than employing only different directions of SLL.

8. References

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9. Appendices

Appendix I: Data extraction tool

Data extraction tool adapted from JBI-SUMARI

Author

Journal

Year

Reviewer

Method

Setting

Participants

Number of participants

Group A

Group B

Interventions A

Intervention B

Appendix II: Ethical Approval

University of Salford
MANCHESTER

Research, Innovation and Academic Engagement Ethical Approval Panel

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6 August 2015

Dear Ahmed,

RE: ETHICS APPLICATION HSCR 15-49 – Within-day and between-days reliability and validity of lower limb kinetic using two dimensional (2D) and three dimensional (3D) movement analysis systems during multidirectional hops

Based on the information you provided, I am pleased to inform you that application HSCR15-49 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

Appendix III: Activity level questioner

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

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

1. **<u>Personal information</u>**

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

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

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

2. **<u>Additional information</u>**

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 ( )

Please note: if you answer YES to any of the following questions, you will be asked to Provide a letter from your GP before being allowed to participate in

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

Please note: if you answered YES to any of the above questions, you will be asked to

Provide a letter from your GP before being allowed to participate in

If this questionnaire was not completed and countersigned immediately prior to the test, the subject must complete this section.

I certify that none of the above information has changed since I completed this questionnaire.

Signed: …….……………………………………… Date: …………………...

Appendix IV: Test of normality for the reliability study

	Shapiro-Wilk		
	Statistic	df	Sig.
2D_FPPA_FSLL_RT_1	.945	12	.564
2D_FPPA_FSLL_RT_2	.987	12	.998
2D_FPPA_FSLL_RT_3	.962	12	.807
2D_FPPA_FSLL_LT_1	.917	12	.265
2D_FPPA_FSLL_LT_2	.857	12	.045
2D_FPPA_FSLL_LT_3	.903	12	.174
2D_FPPA_FSLLP_RT_1	.898	12	.149
2D_FPPA_FSLLP_RT_2	.873	12	.072
2D_FPPA_FSLLP_RT_3	.846	12	.033
2D_FPPA_FSLLP_LT_1	.960	12	.780
2D_FPPA_FSLLP_LT_2	.921	12	.295
2D_FPPA_FSLLP_LT_3	.921	12	.292
2D_FPPA_LSLL_RT_1	.870	12	.065
2D_FPPA_LSLL_RT_2	.709	12	.001
2D_FPPA_LSLL_RT_3	.796	12	.008
2D_FPPA_LSLL_LT_1	.966	12	.868
2D_FPPA_LSLL_LT_2	.928	12	.360
2D_FPPA_LSLL_LT_3	.987	12	.999
2D_FPPA_LSLLP_RT_1	.912	12	.224
2D_FPPA_LSLLP_RT_2	.880	12	.087

2D_FPPA_LSLLP_RT_3	.927	12	.346
2D_FPPA_LSLLP_LT_1	.857	12	.045
2D_FPPA_LSLLP_LT_2	.931	12	.390
2D_FPPA_LSLLP_LT_3	.948	12	.604
2D_FPPA_MSLL_RT_1	.985	12	.996
2D_FPPA_MSLL_RT_2	.956	12	.719
2D_FPPA_MSLL_RT_3	.951	12	.646
2D_FPPA_MSLL_LT_1	.884	12	.098
2D_FPPA_MSLL_LT_2	.960	12	.778
2D_FPPA_MSLL_LT_3	.948	12	.601
2D_FPPA_MSLLP_RT_1	.915	12	.245
2D_FPPA_MSLLP_RT_2	.964	12	.837
2D_FPPA_MSLLP_RT_3	.967	12	.878
2D_FPPA_MSLLP_LT_1	.934	12	.422
2D_FPPA_MSLLP_LT_2	.936	12	.454
2D_FPPA_MSLLP_LT_3	.958	12	.755
2D_HADD_FSLL_RT_1	.961	12	.796
2D_HADD_FSLL_RT_2	.934	12	.427
2D_HADD_FSLL_RT_3	.913	12	.233
2D_HADD_FSLL_LT_1	.975	12	.953
2D_HADD_FSLL_LT_2	.922	12	.306
2D_HADD_FSLL_LT_3	.979	12	.980
2D_HADD_FSLL_RT_1	.973	12	.936
2D_HADD_FSLLP_RT_2	.988	12	.999
2D_HADD_FSLLP_RT_3	.955	12	.716

2D_HADD_FSLLP_LT_1	.956	12	.722
2D_HADD_FSLLP_LT_2	.963	12	.822
2D_HADD_FSLLP_LT_3	.937	12	.457
2D_HADD_LSLL_RT_1	.914	12	.238
2D_HADD_LSLL_RT_2	.849	12	.036
2D_HADD_LSLL_RT_3	.929	12	.374
2D_HADD_LSLL_LT_1	.955	12	.707
2D_HADD_LSLL_LT_2	.963	12	.819
2D_HADD_LSLL_LT_3	.937	12	.464
2D_HADD_LSLLP_RT_1	.917	12	.265
2D_HADD_LSLLP_RT_2	.893	12	.128
2D_HADD_LSLLP_RT_3	.955	12	.705
2D_HADD_LSLLP_LT_1	.959	12	.772
2D_HADD_LSLLP_LT_2	.892	12	.124
2D_HADD_LSLLP_LT_3	.770	12	.004
2D_HADD_MSLL_RT_1	.934	12	.426
2D_HADD_MSLL_RT_2	.942	12	.530
2D_HADD_MSLL_RT_3	.937	12	.463
2D_HADD_MSLL_LT_1	.917	12	.264
2D_HADD_MSLL_LT_2	.968	12	.894
2D_HADD_MSLL_LT_3	.799	12	.009
2D_HADD_MSLLP_RT_1	.917	12	.259
2D_HADD_MSLLP_RT_2	.938	12	.470
2D_HADD_MSLLP_RT_3	.905	12	.185
2D_HADD_MSLLP_LT_1	.887	12	.108

2D_HADD_MSLLP_LT_2	.890	12	.118
2D_HADD_MSLLP_LT_3	.898	12	.150
KV_FSLL_RT_1	.898	12	.149
KV_FSLL_RT_2	.907	12	.197
KV_FSLL_RT_3	.917	12	.264
KV_FSLL_LT_1	.942	12	.522
KV_FSLL_LT_2	.952	12	.671
KV_FSLL_LT_3	.973	12	.939
KV_FSLLP_RT_1	.906	12	.189
KV_FSLLP_RT_2	.961	12	.796
KV_FSLLP_RT_3	.955	12	.704
KV_FSLLP_LT_1	.917	12	.265
KV_FSLLP_LT_2	.962	12	.818
KV_FSLLP_LT_3	.893	12	.127
KV_LSLL_RT_1	.970	12	.909
KV_LSLL_RT_2	.884	12	.099
KV_LSLL_RT_3	.962	12	.813
KV_LSLL_LT_1	.948	12	.604
KV_LSLL_LT_2	.932	12	.396
KV_LSLL_LT_3	.921	12	.299
KV_LSLLP_RT_1	.949	12	.626
KV_LSLLP_RT_2	.961	12	.794
KV_LSLLP_RT_3	.984	12	.995
KV_LSLLP_LT_1	.934	12	.424
KV_LSLLP_LT_2	.958	12	.748

KV_LSLLP_LT_3	.899	12	.152
KV_MSLL_RT_1	.917	12	.262
KV_MSLL_RT_2	.961	12	.799
KV_MSLL_RT_3	.924	12	.325
KV_MSLL_LT_1	.900	12	.160
KV_MSLL_LT_2	.960	12	.784
KV_MSLL_LT_3	.947	12	.593
KV_MSLLP_RT_1	.953	12	.683
KV_MSLLP_RT_2	.954	12	.696
KV_MSLLP_RT_3	.846	12	.033
KV_MSLLP_LT _1	.942	12	.525
KV_MSLLP_LT _2	.925	12	.326
KV_MSLLP_LT_3	.955	12	.713
3DHADD_FSLL_RT_1	.939	12	.486
3DHADD_FSLL_RT_2	.952	12	.671
3DHADD_FSLL_RT_3	.930	12	.381
3DHADD_FSLL_LT_1	.951	12	.658
3DHADD_FSLL_LT_2	.930	12	.375
3DHADD_FSLL_LT_3	.948	12	.609
3DHADD_FSLLP_RT_1	.982	12	.990
3DHADD_FSLLP_RT_2	.956	12	.720
3DHADD_FSLLP_RT_3	.979	12	.980
3DHADD_FSLLP_LT_1	.980	12	.983
3DHADD_FSLLP_LT_2	.967	12	.880
3DHADD_FSLLP_LT_3	.984	12	.994

3DHADD_LSLL_RT_1	.920	12	.286
3DHADD_LSLL_RT _2	.961	12	.803
3DHADD_LSLL_RT _3	.936	12	.443
3DHADD_LSLL_LT_1	.914	12	.240
3DHADD_LSLL_LT_2	.906	12	.187
3DHADD_LSLL_LT_3	.922	12	.299
3DHADD_LSLLP_RT_1	.872	12	.070
3DHADD_LSLLP_RT_2	.958	12	.751
3DHADD_LSLLP_RT_3	.961	12	.799
3DHADD_LSLLP_LT_1	.909	12	.204
3DHADD_LSLLP_LT_2	.951	12	.649
3DHADD_LSLLP_LT_3	.948	12	.611
3DHADD_MSLL_RT_1	.953	12	.687
3DHADD_MSLL_RT_2	.929	12	.374
3DHADD_MSLL_RT_3	.885	12	.102
3DHADD_MSLL_LT_1	.856	12	.044
3DHADD_MSLL_LT_2	.907	12	.197
3DHADD_MSLL_LT_3	.922	12	.305
3DHADD_MSLLP_RT_1	.986	12	.997
3DHADD_MSLLP_RT_2	.957	12	.747
3DHADD_MSLLP_RT_3	.930	12	.377
3DHADD_MSLLP_LT_1	.956	12	.719
3DHADD_MSLLP_LT_2	.958	12	.749
3DHADD_MSLLP_LT_3	.928	12	.361
HADD_Mom_FSLL_RT_1	.923	12	.311

HADD_Mom_FSLL_RT_2	.941	12	.510
HADD_Mom_FSLL_RT_3	.888	12	.111
HADD_Mom_FSLL_LT_1	.970	12	.914
HADD_Mom_FSLL_LT_2	.914	12	.237
HADD_Mom_FSLL_LT_3	.957	12	.736
HADD_Mom_LSLL_RT_1	.957	12	.736
HADD_Mom_LSLL_RT_2	.891	12	.121
HADD_Mom_LSLL_RT_3	.921	12	.298
HADD_Mom_LSLL_LT_1	.844	12	.031
HADD_Mom_LSLL_LT_2	.949	12	.623
HADD_Mom_LSLL_LT_3	.945	12	.560
HADD_Mom_MSLL_RT_1	.896	12	.139
HADD_Mom_MSLL_RT_2	.937	12	.457
HADD_Mom_MSLL_RT_3	.965	12	.849
HADD_Mom_MSLL_LT_1	.960	12	.783
HADD_Mom_MSLL_LT_2	.931	12	.390
HADD_Mom_MSLL_LT_3	.883	12	.096
HADD_Mom_FSLLP_RT_1	.943	12	.534
HADD_Mom_FSLLP_RT_2	.975	12	.952
HADD_Mom_FSLLP_RT_3	.841	12	.029
HADD_Mom_FSLLP_LT_1	.929	12	.372
HADD_Mom_FSLLP_LT_2	.916	12	.252
HADD_Mom_FSLLP_LT_3	.939	12	.491
HADD_Mom_LSLLP_RT_1	.912	12	.228
HADD_Mom_LSLLP_RT_2	.892	12	.124

HADD_Mom_LSLLP_RT_3	.891	12	.120
HADD_Mom_LSLLP_LT_1	.945	12	.566
HADD_Mom_LSLLP_LT_2	.974	12	.947
HADD_Mom_LSLLP_LT_3	.911	12	.222
HADD_Mom_MSLLP_RT_1	.945	12	.569
HADD_Mom_MSLLP_RT_2	.888	12	.112
HADD_Mom_MSLLP_RT_3	.656	12	.000
HADD_Mom_MSLLP_LT_1	.951	12	.649
HADD_Mom_MSLLP_LT_2	.938	12	.469
HADD_Mom_MSLLP_LT_3	.913	12	.234
Int_Rot_angle_FSLL_RT_1	.896	12	.141
Int_Rot_angle_FSLL_RT_2	.952	12	.660
Int_Rot_angle_FSLL_RT_3	.945	12	.565
Int_Rot_angle_FSLL_LT_1	.849	12	.035
Int_Rot_angle_FSLL_LT_2	.922	12	.305
Int_Rot_angle_FSLL_LT_3	.897	12	.144
Int_Rot_angle_FSLLP_RT_1	.937	12	.464
Int_Rot_angle_FSLLP_RT_2	.919	12	.282
Int_Rot_angle_FSLLP_RT_3	.929	12	.373
Int_Rot_angle_FSLLP_LT_1	.881	12	.089
Int_Rot_angle_FSLLP_LT_2	.893	12	.129
Int_Rot_angle_FSLLP_LT_3	.958	12	.755
Int_Rot_angle_LSLL_RT_1	.963	12	.819
Int_Rot_angle_LSLL_RT_2	.969	12	.901
Int_Rot_angle_LSLL_RT_3	.911	12	.221

Int_Rot_angle_LSLL_LT_1	.911	12	.218
Int_Rot_angle_LSLL_LT_2	.940	12	.493
Int_Rot_angle_LSLL_LT_3	.961	12	.794
Int_Rot_angle_LSLLP_RT_1	.870	12	.065
Int_Rot_angle_LSLLP_RT_2	.889	12	.113
Int_Rot_angle_LSLLP_RT_3	.870	12	.065
Int_Rot_angle_LSLLP_LT_1	.955	12	.711
Int_Rot_angle_LSLLP_LT_2	.957	12	.742
Int_Rot_angle_LSLLP_LT_3	.991	12	1.000
Int_Rot_angle_MSLL_RT_1	.874	12	.074
Int_Rot_angle_MSLL_RT_2	.899	12	.156
Int_Rot_angle_MSLL_RT_3	.912	12	.229
Int_Rot_angle_MSLL_LT_1	.818	12	.015
Int_Rot_angle_MSLL_LT_2	.858	12	.046
Int_Rot_angle_MSLL_LT_3	.870	12	.066
Int_Rot_angle_MSLLP_RT_1	.960	12	.779
Int_Rot_angle_MSLLP_RT_2	.941	12	.508
Int_Rot_angle_MSLLP_RT_3	.908	12	.204
Int_Rot_angle_MSLLP_LT_1	.934	12	.429
Int_Rot_angle_MSLLP_LT_2	.922	12	.306
Int_Rot_angle_MSLLP_LT_3	.917	12	.259
Int_Rot_Mom_FSLL_RT_1	.902	12	.167
Int_Rot_Mom_FSLL_RT_2	.978	12	.976
Int_Rot_Mom_FSLL_RT_3	.922	12	.307
Int_Rot_Mom_FSLL_LT_1	.963	12	.822

Int_Rot_Mom_FSLL_LT_2	.940	12	.493
Int_Rot_Mom_FSLL_LT_3	.902	12	.169
Int_Rot_Mom_FSLLP_RT_1	.874	12	.073
Int_Rot_Mom_FSLLP_RT_2	.927	12	.351
Int_Rot_Mom_FSLLP_RT_3	.861	12	.051
Int_Rot_Mom_FSLLP_LT_1	.957	12	.734
Int_Rot_Mom_FSLLP_LT_2	.959	12	.768
Int_Rot_Mom_FSLLP_LT_3	.950	12	.640
Int_Rot_Mom_LSLL_RT_1	.942	12	.527
Int_Rot_Mom_LSLL_RT_2	.920	12	.286
Int_Rot_Mom_LSLL_RT_3	.880	12	.087
Int_Rot_Mom_LSLL_LT_1	.888	12	.112
Int_Rot_Mom_LSLL_LT_2	.936	12	.453
Int_Rot_Mom_LSLL_LT_3	.930	12	.383
Int_Rot_Mom_LSLLP_RT_1	.943	12	.544
Int_Rot_Mom_LSLLP_RT_2	.961	12	.797
Int_Rot_Mom_LSLLP_RT_3	.949	12	.618
Int_Rot_Mom_LSLLP_LT_1	.923	12	.315
Int_Rot_Mom_LSLLP_LT_2	.983	12	.992
Int_Rot_Mom_LSLLP_LT_3	.955	12	.716
Int_Rot_Mom_MSLL_RT_1	.930	12	.385
Int_Rot_Mom_MSLL_RT_2	.924	12	.325
Int_Rot_Mom_MSLL_RT_3	.911	12	.220
Int_Rot_Mom_MSLL_LT_1	.960	12	.777
Int_Rot_Mom_MSLL_LT_2	.942	12	.521

Int_Rot_Mom_MSLL_LT_3	.933	12	.412
Int_Rot_Mom_MSLLP_RT_1	.921	12	.292
Int_Rot_Mom_MSLLP_RT_2	.955	12	.704
Int_Rot_Mom_MSLLP_RT_3	.882	12	.093
Int_Rot_Mom_MSLLP_LT_1	.928	12	.360
Int_Rot_Mom_MSLLP_LT_2	.961	12	.796
Int_Rot_Mom_MSLLP_LT_3	.961	12	.805
Knee_FLX_Angle_FSLL_RT_1	.956	12	.722
Knee_FLX_Angle_FSLL_RT_2	.889	12	.115
Knee_FLX_Angle_FSLL_RT_3	.960	12	.777
Knee_FLX_Angle_FSLL_LT_1	.943	12	.536
Knee_FLX_Angle_FSLL_LT_2	.942	12	.519
Knee_FLX_Angle_FSLL_LT_3	.951	12	.650
Knee_FLX_Angle_FSLLP_RT_1	.943	12	.540
Knee_FLX_Angle_FSLLP_RT_2	.912	12	.226
Knee_FLX_Angle_FSLLP_RT_3	.901	12	.161
Knee_FLX_Angle_FSLLP_LT_1	.868	12	.062
Knee_FLX_Angle_FSLLP_LT_2	.903	12	.175
Knee_FLX_Angle_FSLLP_LT_3	.940	12	.495
Knee_FLX_Angle_LSLL_RT_1	.953	12	.688
Knee_FLX_Angle_LSLL_RT_2	.912	12	.224
Knee_FLX_Angle_LSLL_RT_3	.951	12	.655
Knee_FLX_Angle_LSLL_LT_1	.937	12	.460
Knee_FLX_Angle_LSLL_LT_2	.927	12	.346
Knee_FLX_Angle_LSLL_LT_3	.975	12	.954

Knee_FLX_Angle_LSLLP_RT_1	.890	12	.116
Knee_FLX_Angle_LSLLP_RT_2	.909	12	.208
Knee_FLX_Angle_LSLLP_RT_3	.859	12	.048
Knee_FLX_Angle_LSLLP_LT_1	.933	12	.410
Knee_FLX_Angle_LSLLP_LT_2	.962	12	.816
Knee_FLX_Angle_LSLLP_LT_3	.939	12	.480
Knee_FLX_Angle_MSLL_RT_1	.941	12	.505
Knee_FLX_Angle_MSLL_RT_2	.977	12	.970
Knee_FLX_Angle_MSLL_RT_3	.961	12	.800
Knee_FLX_Angle_MSLL_LT_1	.936	12	.450
Knee_FLX_Angle_MSLL_LT_2	.950	12	.636
Knee_FLX_Angle_MSLL_LT_3	.934	12	.425
Knee_FLX_Angle_MSLLP_RT_1	.897	12	.146
Knee_FLX_Angle_MSLLP_RT_2	.964	12	.840
Knee_FLX_Angle_MSLLP_RT_3	.808	12	.012
Knee_FLX_Angle_MSLLP_LT_1	.972	12	.933
Knee_FLX_Angle_MSLLP_LT_2	.960	12	.789
Knee_FLX_Angle_MSLLP_LT_3	.939	12	.485
knee_EXT_Mom_FSLL_RT_1	.845	12	.032
knee_EXT_Mom_FSLL_RT_2	.906	12	.189
knee_EXT_Mom_FSLL_RT_3	.965	12	.857
knee_EXT_Mom_FSLL_LT_1	.929	12	.368
knee_EXT_Mom_FSLL_LT_2	.955	12	.707
knee_EXT_Mom_FSLL_LT_3	.926	12	.336
knee_EXT_Mom_FSLLP_RT_1	.953	12	.675

knee_EXT_Mom_FSLLP_RT_2	.929	12	.372
knee_EXT_Mom_FSLLP_RT_3	.956	12	.728
knee_EXT_Mom_FSLLP_LT_1	.966	12	.869
knee_EXT_Mom_FSLLP_LT_2	.923	12	.310
knee_EXT_Mom_FSLLP_LT_3	.874	12	.073
knee_EXT_Mom_LSLL_RT_1	.912	12	.226
knee_EXT_Mom_LSLL_RT_2	.930	12	.383
knee_EXT_Mom_LSLL_RT_3	.918	12	.274
knee_EXT_Mom_LSLL_LT_1	.950	12	.640
knee_EXT_Mom_LSLL_LT_2	.954	12	.699
knee_EXT_Mom_LSLL_LT_3	.937	12	.459
knee_EXT_Mom_LSLLP_RT_1	.945	12	.571
knee_EXT_Mom_LSLLP_RT_2	.931	12	.394
knee_EXT_Mom_LSLLP_RT_3	.986	12	.998
knee_EXT_Mom_LSLLP_LT_1	.923	12	.310
knee_EXT_Mom_LSLLP_LT_2	.894	12	.134
knee_EXT_Mom_LSLLP_LT_3	.937	12	.466
knee_EXT_Mom_MSLL_RT_1	.974	12	.946
knee_EXT_Mom_MSLL_RT_2	.969	12	.901
knee_EXT_Mom_MSLL_RT_3	.911	12	.222
knee_EXT_Mom_MSLL_LT_1	.940	12	.504
knee_EXT_Mom_MSLL_LT_2	.959	12	.774
knee_EXT_Mom_MSLL_LT_3	.950	12	.644
knee_EXT_Mom_MSLLP_RT_1	.937	12	.461
knee_EXT_Mom_MSLLP_RT_2	.933	12	.413

knee_EXT_Mom_MSLLP_RT_3	.917	12	.265
knee_EXT_Mom_MSLLP_LT_1	.979	12	.981
knee_EXT_Mom_MSLLP_LT_2	.952	12	.667
knee_EXT_Mom_MSLLP_LT_3	.962	12	.816
KV_MOM_FSLL_RT_1	.933	12	.410
KV_MOM_FSLL_RT_2	.902	12	.167
KV_MOM_FSLL_RT_3	.913	12	.232
KV_MOM_FSLL_LT_1	.962	12	.808
KV_MOM_FSLL_LT_2	.942	12	.531
KV_MOM_FSLL_LT_3	.907	12	.198
KV_MOM_FSLLP_RT_1	.902	12	.168
KV_MOM_FSLLP_RT_2	.904	12	.177
KV_MOM_FSLLP_RT_3	.947	12	.588
KV_MOM_FSLLP_LT_1	.965	12	.855
KV_MOM_FSLLP_LT_2	.974	12	.948
KV_MOM_FSLLP_LT_3	.936	12	.448
KV_MOM_LSLL_RT_1	.955	12	.714
KV_MOM_LSLL_RT_2	.823	12	.017
KV_MOM_LSLL_RT_3	.941	12	.505
KV_MOM_LSLL_LT_1	.954	12	.701
KV_MOM_LSLL_LT_2	.866	12	.058
KV_MOM_LSLL_LT_3	.905	12	.183
KV_MOM_LSLLP_RT_1	.912	12	.226
KV_MOM_LSLLP_RT_2	.955	12	.715
KV_MOM_LSLLP_RT_3	.934	12	.426

KV_MOM_LSLLP_LT_1	.965	12	.858
KV_MOM_LSLLP_LT_2	.904	12	.178
KV_MOM_LSLLP_LT_3	.952	12	.660
KV_MOM_MSLL_RT_1	.951	12	.646
KV_MOM_MSLL_RT_2	.949	12	.628
KV_MOM_MSLL_RT_3	.896	12	.143
KV_MOM_MSLL_LT_1	.926	12	.341
KV_MOM_MSLL_LT_2	.839	12	.027
KV_MOM_MSLL_LT_3	.894	12	.132
KV_MOM_MSLLP_RT_1	.847	12	.034
KV_MOM_MSLLP_RT_2	.923	12	.314
KV_MOM_MSLLP_RT_3	.933	12	.412
KV_MOM_MSLLP_LT_1	.938	12	.477
KV_MOM_MSLLP_LT_2	.908	12	.199
KV_MOM_MSLLP_LT_3	.840	12	.028
GRF_FSLL_RT_1	.912	12	.228
GRF_FSLL_RT_2	.932	12	.402
GRF_FSLL_RT_3	.933	12	.408
GRF_FSLL_LT_1	.840	12	.028
GRF_FSLL_LT_2	.945	12	.570
GRF_FSLL_LT_3	.924	12	.321
GRF_FSLLP_RT_1	.980	12	.982
GRF_FSLLP_RT_2	.914	12	.242
GRF_FSLLP_RT_3	.961	12	.791
GRF_FSLLP_LT_1	.917	12	.259

GRF_FSLLP_LT_2	.848	12	.035
GRF_FSLLP_LT_3	.883	12	.095
GRF_LSLL_RT_1	.955	12	.713
GRF_LSLL_RT_2	.951	12	.653
GRF_LSLL_RT_3	.960	12	.780
GRF_LSLL_LT_1	.927	12	.354
GRF_LSLL_LT_2	.938	12	.474
GRF_LSLL_LT_3	.899	12	.155
GRF_LSLLP_RT_1	.936	12	.445
GRF_LSLLP_RT_2	.952	12	.669
GRF_LSLLP_RT_3	.881	12	.089
GRF_LSLLP_LT_1	.917	12	.262
GRF_LSLLP_LT_2	.842	12	.029
GRF_LSLLP_LT_3	.913	12	.235
GRF_MSLL_RT_1	.927	12	.350
GRF_MSLL_RT_2	.969	12	.896
GRF_MSLL_RT_3	.876	12	.077
GRF_MSLL_LT_1	.967	12	.876
GRF_MSLL_LT_2	.962	12	.805
GRF_MSLL_LT_3	.885	12	.101
GRF_MSLLP_RT_1	.942	12	.524
GRF_MSLLP_RT_2	.987	12	.998
GRF_MSLLP_RT_3	.914	12	.241
GRF_MSLLP_LT_2	.970	12	.916
GRF_MSLLP_LT_3	.937	12	.459

Appendix V: The output of G power 2 statistical software for power calculation

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

Appendix VI: Result of normality test for validity study.

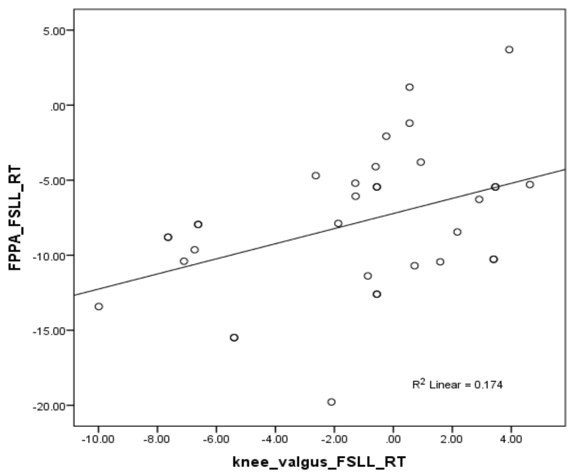
Test of normality for 2D FPPA, 2D HADD, 3D knee valgus and 3D HADD for both legs during all tasks							
2D Variables	Shapiro-Wilk			3D variables	Shapiro-Wilk		
	Statistic	df	Sig.		Statistic	df	Sig.
2D_FPPA_FSLL_RT	.985	34	.901	3D_KV_FSLL_RT	.930	34	.131
2D_FPPA_FSLL_LT	.976	34	.635	3D_KV_FSLL_LT	.973	34	.548
2D_FPPA_FSLLP_RT	.955	34	.172	3D_KV_FSLLP_RT	.968	34	.419
2D_FPPA_FSLLP_LT	.975	34	.610	3D_KV_FSLLP_LT	.954	34	.160
2D_FPPA_LSLL_RT	.840	34	.132	3D_KV_LSLL_RT	.973	34	.549
2D_FPPA_LSLL_LT	.885	34	.112	3D_KV_LSLL_LT	.955	34	.177
2D_FPPA_LSLLP_RT	.929	34	.129	3D_KV_LSLLP_RT	.976	34	.639
2D_FPPA_LSLLP_LT	.937	34	.050	3D_KV_LSLLP_LT	.955	34	.174
2D_FPPA_MSLL_RT	.985	34	.911	3D_KV_MSLL_RT	.959	34	.221
2D_FPPA_MSLL_LT	.949	34	.112	3D_KV_MSLL_LT	.978	34	.697
2D_FPPA_MSLLP_RT	.955	34	.175	3D_KV_MSLLP_RT	.962	34	.282
2D_FPPA_MSLLP_LT	.967	34	.382	3D_KV_MSLLP_LT	.978	34	.695
2D_HADD_FSLL_RT	.968	34	.403	3D_HADD_FSLL_RT	.960	34	.241
2D_HADD_FSLL_LT	.980	34	.785	3D_HADD_FSLL_LT	.934	34	.142
2D_HADD_FSLLP_RT	.905	34	.116	3D_HADD_FSLLP_RT	.958	34	.206
2D_HADD_FSLLP_LT	.968	34	.415	3D_HADD_FSLLP_LT	.983	34	.862
2D_HADD_LSLL_RT	.938	34	.054	3D_HADD_LSLL_RT	.930	34	.130
2D = two-dimensional, 3D = three dimensional, FPPA = frontal plane projection angle, HADD = hip adduction angle, KV = knee valgus angle, FSLL = forward single leg landing, FSLLP = forward single leg landing off a platform, LSLL, lateral single leg landing, LSLLP = lateral single leg landing off a platform, MSLL = medial single leg landing, MSLLP = medial single leg landing off a platform. RT = right leg, LT = left leg.							

Continue test of normality for 2D FPPA, 2D HADD, 3D knee valgus and 3D HADD for both legs during all tasks							
2D Variables	Shapiro-Wilk			3D variables	Shapiro-Wilk		
	Statistic	df	Sig.		Statistic	df	Sig.
2D_HADD_LSLL_LT	.952	34	.137	3D_HADD_LSLL_LT	.979	34	.734
2D_HADD_LSLLP_RT	.978	34	.718	3D_HADD_LSLLP_RT	.960	34	.249
2D_HADD_LSLLP_LT	.983	34	.869	3D_HADD_LSLLP_LT	.968	34	.406
2D_HADD_MSLL_RT	.929	34	.128	3D_HADD_MSLL_RT	.951	34	.134
2D_HADD_MSLL_LT	.955	34	.179	3D_HADD_MSLL_LT	.931	34	.133
2D_HADD_MSLLP_RT	.967	34	.375	3D_HADD_MSLLP_RT	.965	34	.333
2D_HADD_MSLLP_LT	.973	34	.561	3D_HADD_MSLLP_LT	.974	34	.571

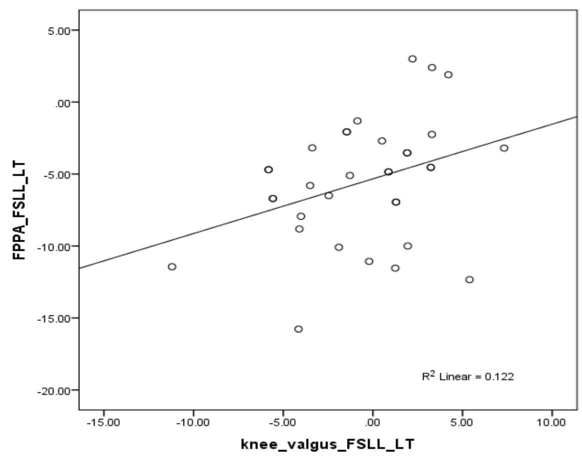
2D = two dimensional, 3D = three dimensional, FPPA = frontal plane projection angle, HADD = hip adduction angle, KV = knee valgus angle, FSLL = forward single leg landing, FSLLP = forward single leg landing off a platform, LSLL, lateral single leg landing, LSLLP = lateral single leg landing off a platform, MSLL = medial single leg landing, MSLLP = medial single leg landing off platform. RT = right leg, LT = left leg.

Appendix VII: Scatter plot illustrating the correlation between 2D FPPA and 3D knee valgus and between 2D HADD angle and 3D HADD angle in all tasks for both limbs

The correlation between 2D FPPA and 3D knee valgus in all tasks for both limbs

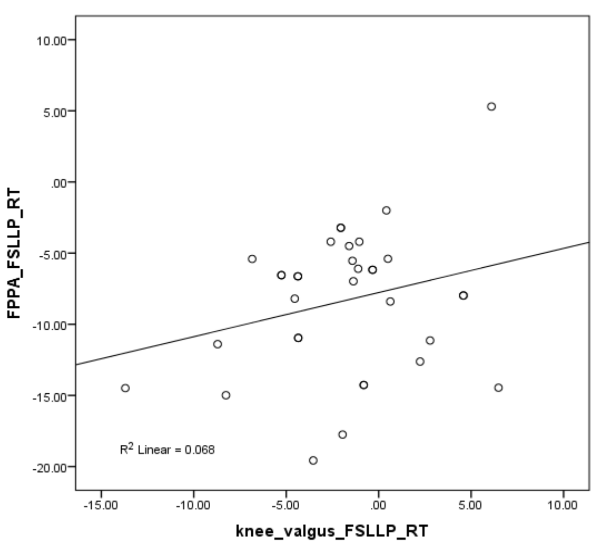


Right leg (r = 0.42, P = 0.014)

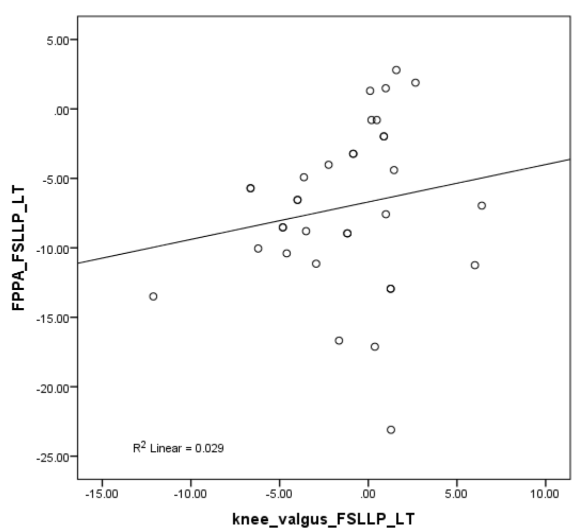


Left leg (r = 0.35, P= 0.043)

forward SLL

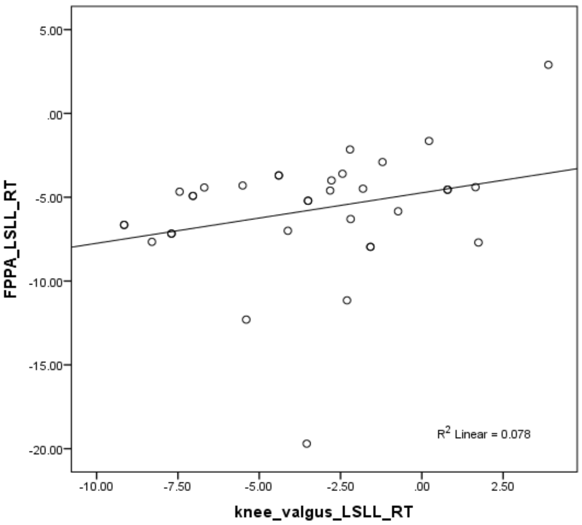


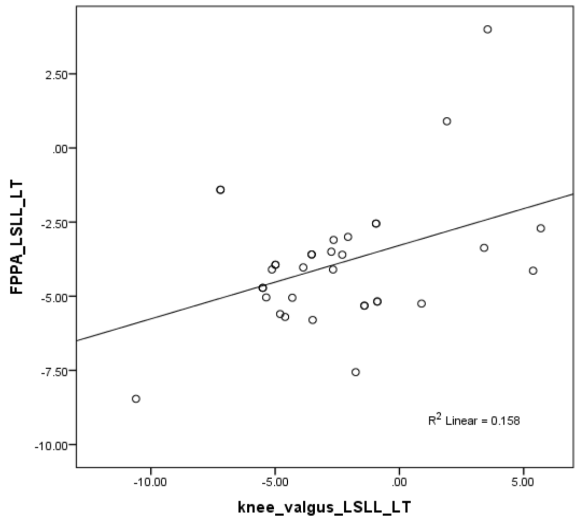
Right leg (r = 0.26, P = 0.13)



Left leg (r = 0.17, P = 0.33)

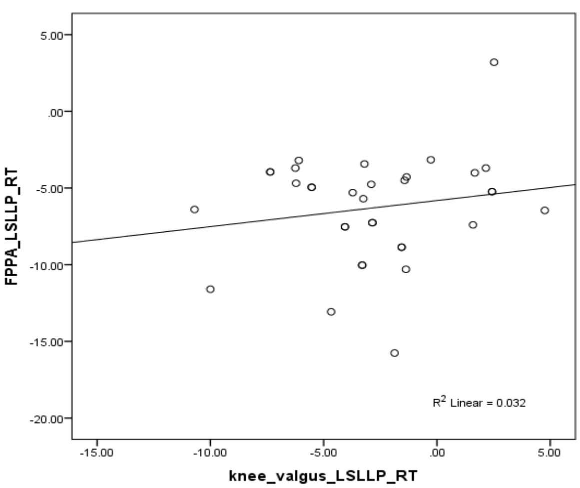
Forward SLL off a platform

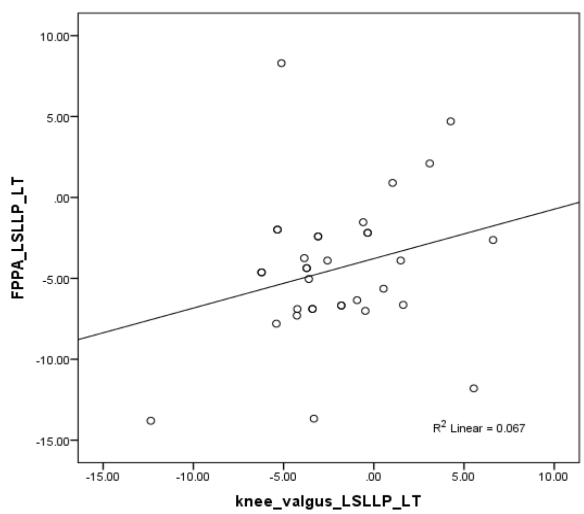




Right leg (r = 0.28, P = 0.11)   Left leg (r = 0.40, P = 0.02)

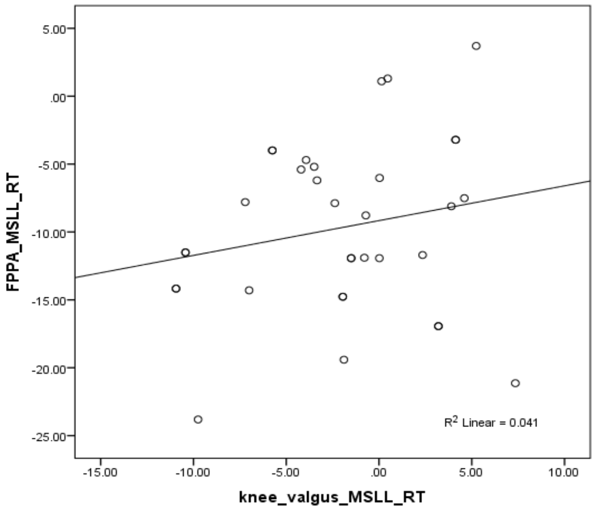
Lateral SLL

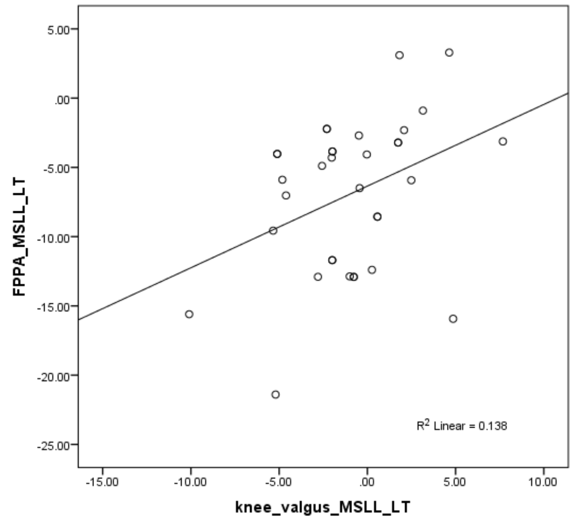




Right leg (r = 0.18, P = 0.31)   Left leg (r = 0.26, P = 0.14)

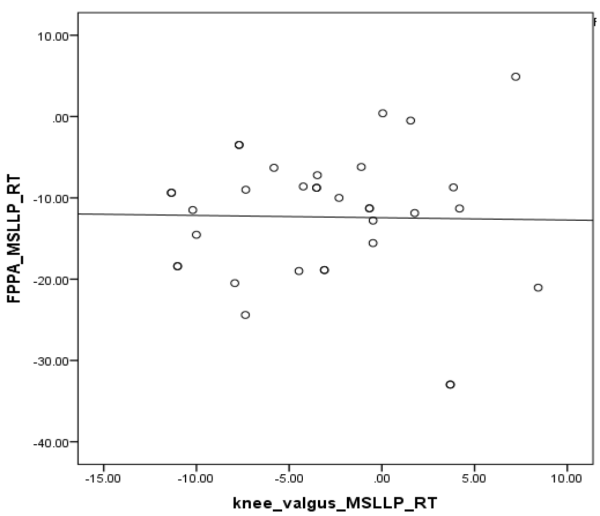
Lateral SLL off a platform

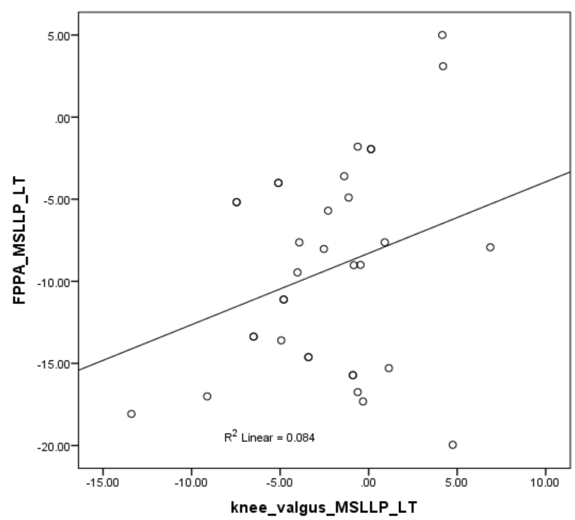




Right leg (r = 0.20, P = 0.24)   Left leg (r = 0.37, P = 0.03)

Medial SLL

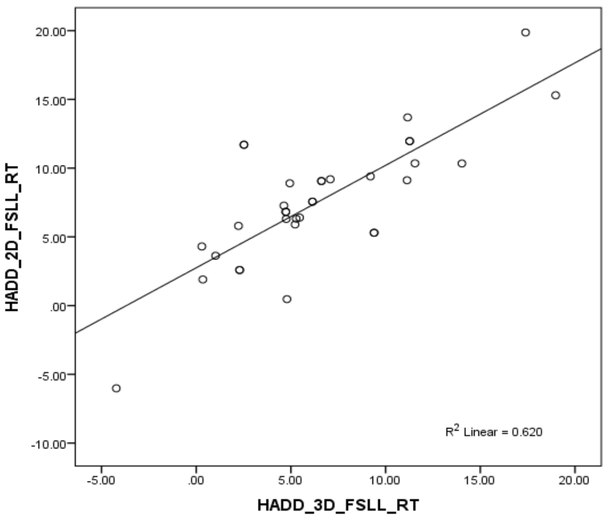




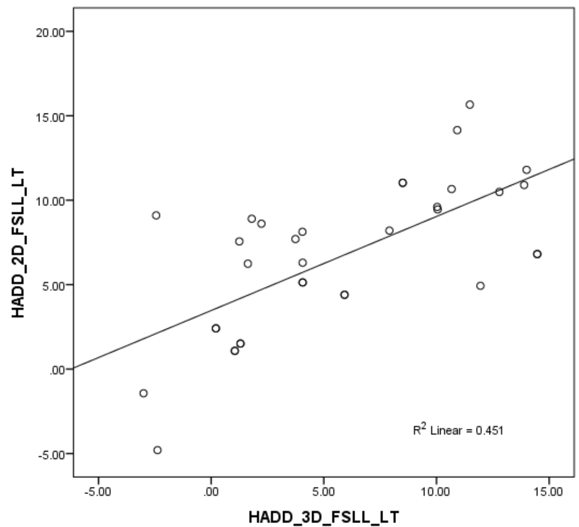
Right leg (r = -0.02, P = 0.92)   Left leg (r = 0.29, P = 0.096)

Medial SLL off a platform

Scatter plot illustrating the correlation between 2D HADD angle and 3D HADD angle in all tasks for both limbs.

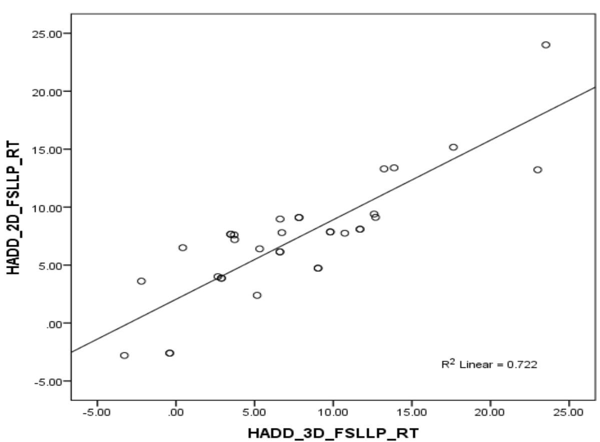


Right leg (r = 0.79, P < 0.001)

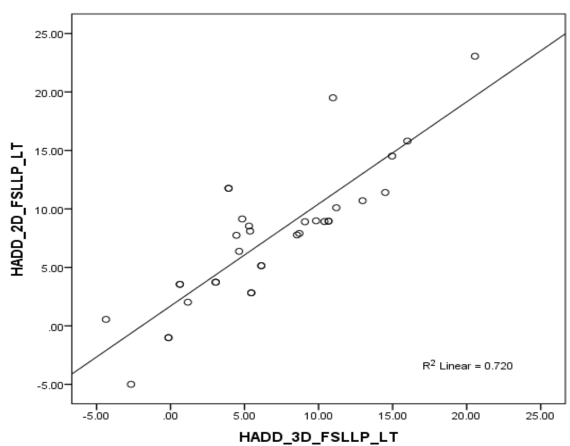


Left leg (r = 0.70, P < 0.001)

Forward SLL

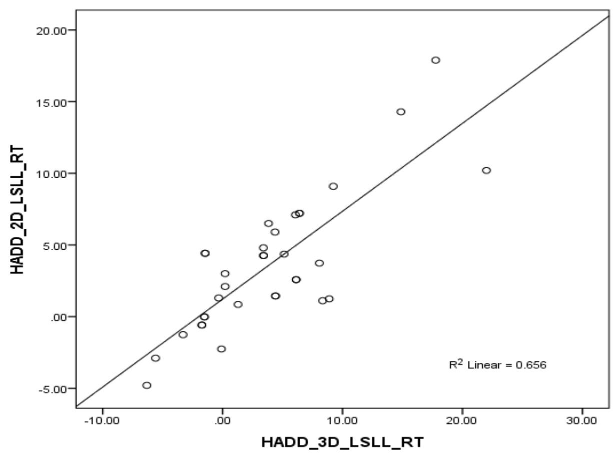


Right leg (r = 0.85, P < 0.001)

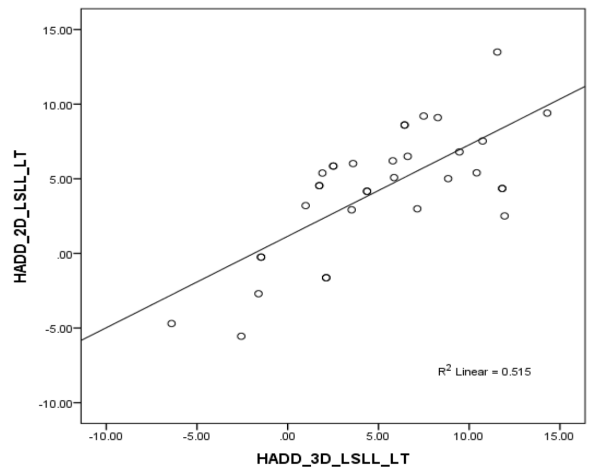


Left leg (r = 0.70, P < 0.001)

Forward SLL off a platform

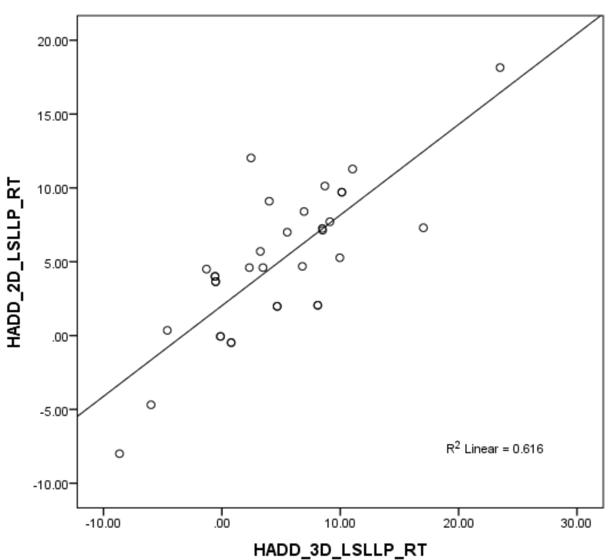


Right leg (r = 0.81, P < 0.001)

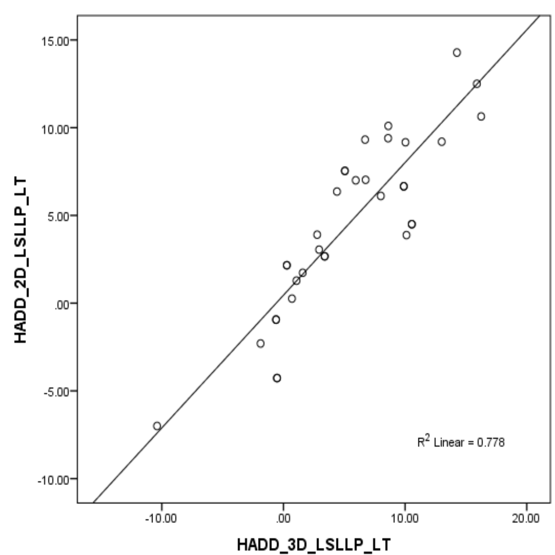


Left leg (r = 0.72, P < 0.001)

Lateral SLL

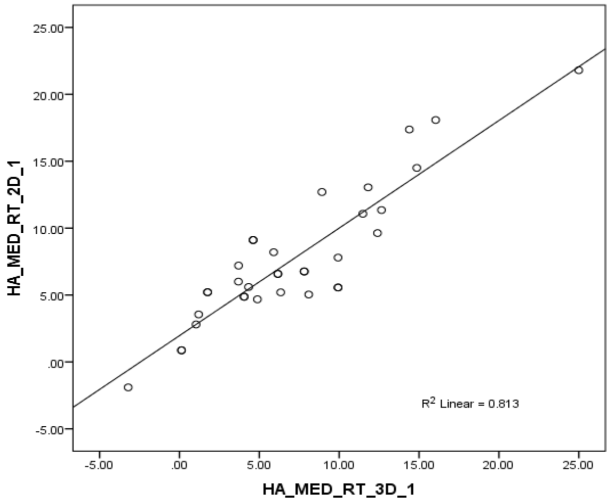


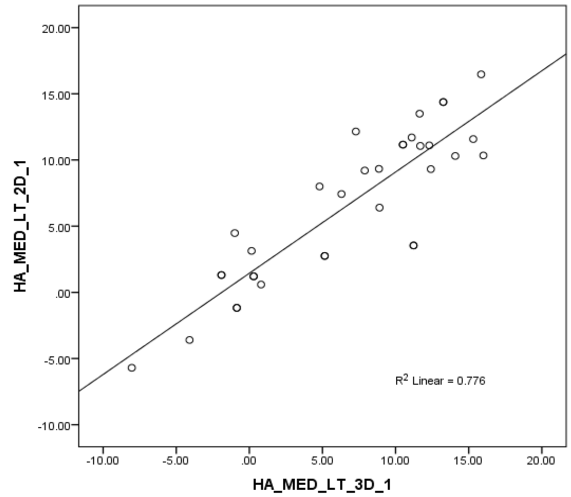
Right leg (r = 0.79, P < 0.001)



Left leg (r = 0.88, P < 0.001)

Lateral SLL off a platform

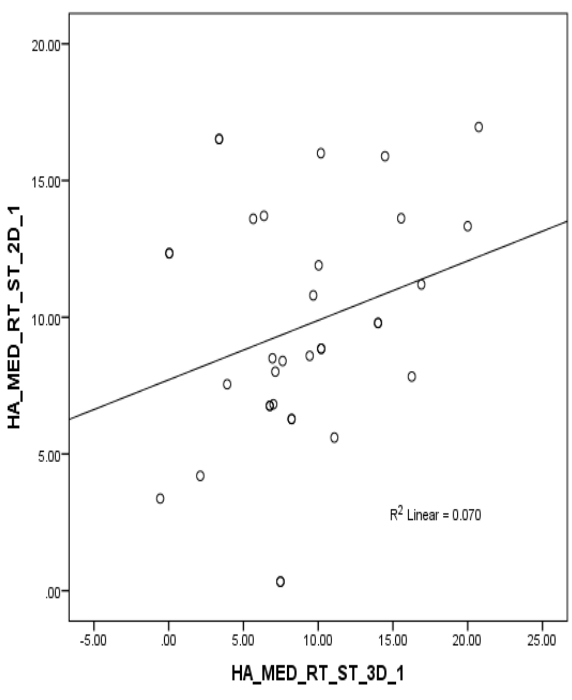


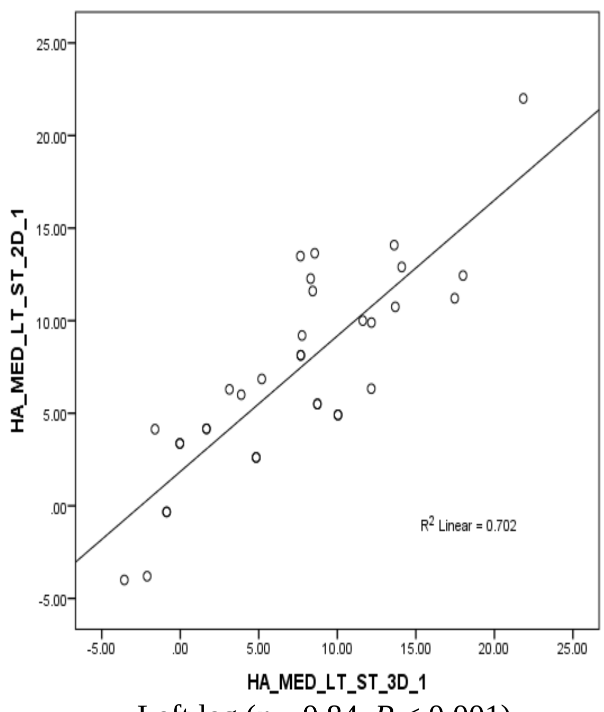


Right leg (r = 0.90, P <

Left leg (r = 0.88, P < 0.001)

Medial SLL





Right leg (r = 0.27, P = 0.13)

Left leg (r = 0.84, P < 0.001)

Medial SLL off a platform

Appendix VIII: The results of One Sample T test and Bland-Altman plots for the validity study

Between 2D FPPA and 3D knee valgus

1- Between 2D FPPA and 3D kV during FH right leg

One-Sample Statistics

	N	Mean	Std. Deviation	Std. Error Mean
Difference	34	-6.5691	4.79246	.82190

One-Sample Test

	Test Value = 0					
					95% Confidence Interval of the Difference	
	t	df	Sig. (2-tailed)	Mean Difference	Lower	Upper
Difference	-7.993	33	.000	-6.56912	-8.2413	-4.8969

2- Between 2D FPPA and 3D kV during FH left leg

One-Sample Statistics

	N	Mean	Std. Deviation	Std. Error Mean
Difference	34	-4.9906	4.64930	.79735

One-Sample Test

	Test Value = 0					
					95% Confidence Interval of the Difference	
	t	df	Sig. (2-tailed)	Mean Difference	Lower	Upper
Difference	-6.259	33	.000	-4.99059	-6.6128	-3.3684

3- Between 2D FPPA and 3D kV during FH_STEP right leg

One-Sample Statistics

	N	Mean	Std. Deviation	Std. Error Mean
Difference	34	-6.4971	5.66732	.97194

One-Sample Test

	Test Value = 0					
					95% Confidence Interval of the Difference	
	t	df	Sig. (2-tailed)	Mean Difference	Lower	Upper
Difference	-6.685	33	.000	-6.49706	-8.4745	-4.5196

4- Between 2D FPPA and 3D kV during FH_STEP left leg

One-Sample Statistics

	N	Mean	Std. Deviation	Std. Error Mean
Difference	34	-5.7282	6.38726	1.09541

One-Sample Test

	Test Value = 0					
					95% Confidence Interval of the Difference	
	t	df	Sig. (2-tailed)	Mean Difference	Lower	Upper
Difference	-5.229	33	.000	-5.72824	-7.9569	-3.4996

5- Between 2D FPPA and 3D KV during LSLL right leg

One-Sample Statistics

	N	Mean	Std. Deviation	Std. Error Mean
Difference	34	-2.3253	4.20286	.72078

One-Sample Test

	Test Value = 0					
					95% Confidence Interval of the Difference	
	t	df	Sig. (2-tailed)	Mean Difference	Lower	Upper
Difference	-3.226	33	.003	-2.32529	-3.7917	-.8588

6- Between 2D FPPA and 3D KV during LSLL left leg

One-Sample Statistics

	N	Mean	Std. Deviation	Std. Error Mean
Difference	34	-1.4174	3.36356	.57685

One-Sample Test

	Test Value = 0					
					95% Confidence Interval of the Difference	
	t	df	Sig. (2-tailed)	Mean Difference	Lower	Upper
Difference	-2.457	33	.019	-1.41735	-2.5910	-.2438

7- Between 2D FPPA and 3D KV during LSLLP right leg

One-Sample Statistics

	N	Mean	Std. Deviation	Std. Error Mean
Difference	34	-3.4944	4.54041	.77867

One-Sample Test

	Test Value = 0					
					95% Confidence Interval of the Difference	
	t	df	Sig. (2-tailed)	Mean Difference	Lower	Upper
Difference	-4.488	33	.000	-3.49441	-5.0786	-1.9102

8- Between 2D FPPA and 3D KV during LSLLP left leg

One-Sample Statistics

	N	Mean	Std. Deviation	Std. Error Mean
Difference	34	-2.3429	5.02305	.86145

One-Sample Test

	Test Value = 0					
					95% Confidence Interval of the Difference	
	t	df	Sig. (2-tailed)	Mean Difference	Lower	Upper
Difference	-2.720	33	.010	-2.34294	-4.0956	-.5903

9- Between 2D FPPA and 3D KV during MSLL right leg

One-Sample Statistics

	N	Mean	Std. Deviation	Std. Error Mean
Difference	34	-7.6956	7.29186	1.25054

One-Sample Test

	Test Value = 0					
					95% Confidence Interval of the Difference	
	t	df	Sig. (2-tailed)	Mean Difference	Lower	Upper
Difference	-6.154	33	.000	-7.69559	-10.2398	-5.1513

10- Between 2D FPPA and 3D KV during MSLL left leg

One-Sample Statistics

	N	Mean	Std. Deviation	Std. Error Mean
Difference	34	-5.9641	5.35813	.91891

One-Sample Test

	Test Value = 0					
					95% Confidence Interval of the Difference	
	t	df	Sig. (2-tailed)	Mean Difference	Lower	Upper
Difference	-6.490	33	.000	-5.96412	-7.8337	-4.0946

11- Between 2D FPPA and 3D KV during MSLLP right leg

One-Sample Statistics

	N	Mean	Std. Deviation	Std. Error Mean
Difference	34	-9.2521	10.07913	1.72856

One-Sample Test

	Test Value = 0					
					95% Confidence Interval of the Difference	
	t	df	Sig. (2-tailed)	Mean Difference	Lower	Upper
Difference	-5.352	33	.000	-9.25206	-12.7688	-5.7353

12- Between 2D FPPA and 3D KV during MSLLP left leg

One-Sample Statistics

	N	Mean	Std. Deviation	Std. Error Mean
Difference	34	-6.9685	6.43002	1.10274

One-Sample Test

	Test Value = 0					
					95% Confidence Interval of the Difference	
	t	df	Sig. (2-tailed)	Mean Difference	Lower	Upper
Difference	-6.319	33	.000	-6.96853	-9.2121	-4.7250

2D and 3D HADD

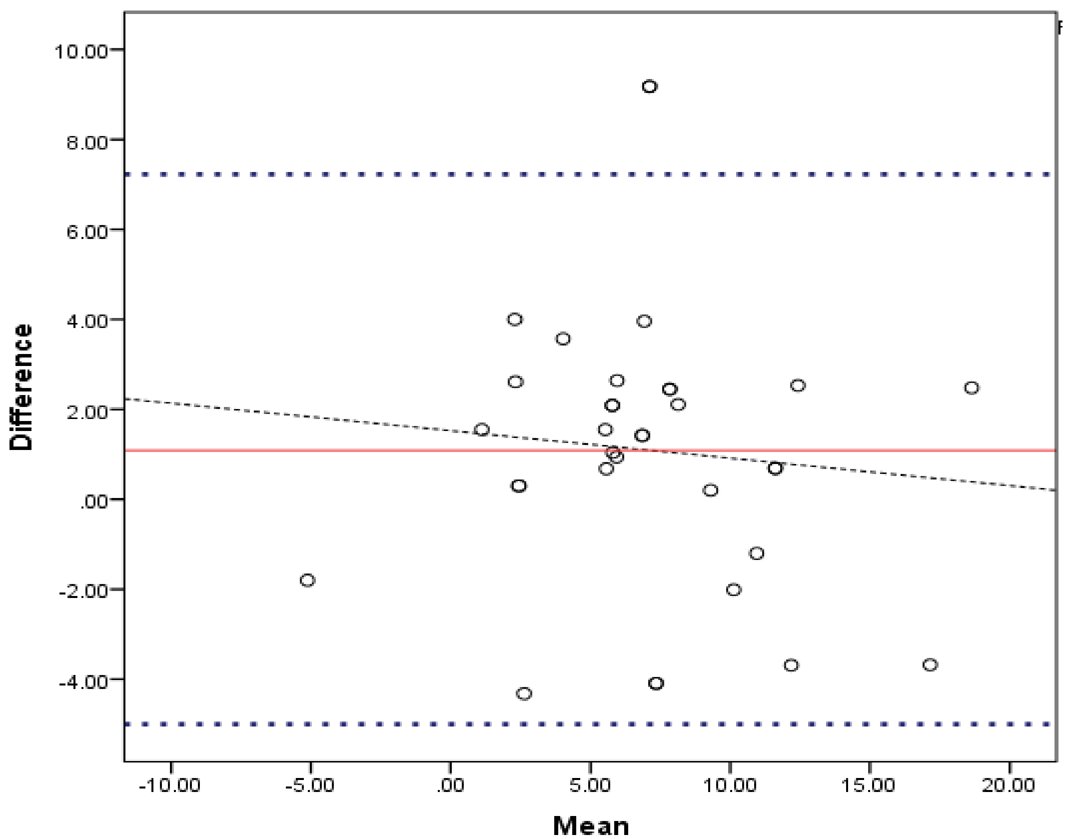
1- Between 2D and 3D HADD during FSLL right leg

One-Sample Statistics

	N	Mean	Std. Deviation	Std. Error Mean
Difference	34	1.0953	3.13409	.53749

One-Sample Test

	Test Value = 0					
					95% Confidence Interval of the Difference	
	t	df	Sig. (2-tailed)	Mean Difference	Lower	Upper
Difference	2.038	33	.050	1.09529	.0018	2.1888



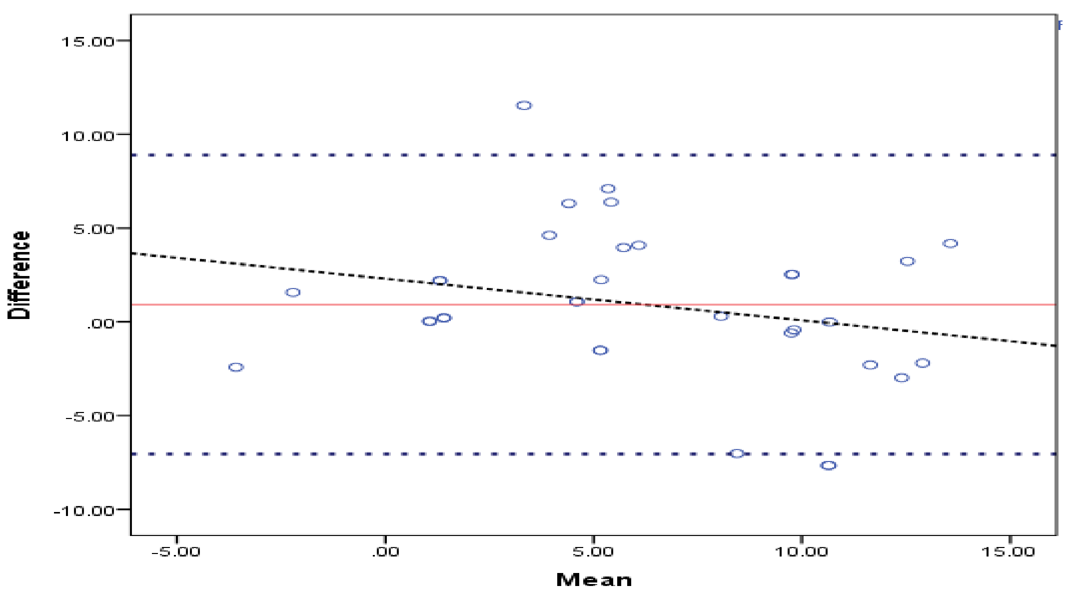
2- Between 2D and 3D HADD during FSLL left leg

One-Sample Statistics

	N	Mean	Std. Deviation	Std. Error Mean
Difference	34	.9188	4.06774	.69761

One-Sample Test

	Test Value = 0					
					95% Confidence Interval of the Difference	
	t	df	Sig. (2-tailed)	Mean Difference	Lower	Upper
Difference	1.317	33	.197	.91882	-.5005	2.3381



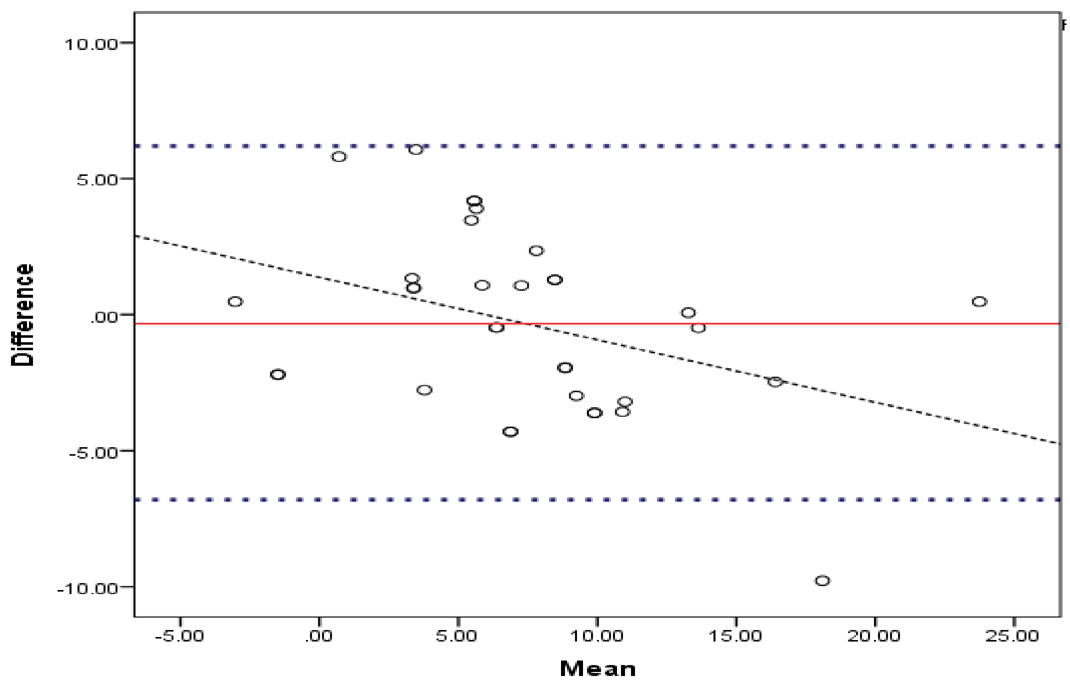
3- Between 2D and 3D HADD during FSLLP right leg

One-Sample Statistics

	N	Mean	Std. Deviation	Std. Error Mean
Difference	34	-.3326	3.33310	.57162

One-Sample Test

	Test Value = 0					
					95% Confidence Interval of the Difference	
	t	df	Sig. (2-tailed)	Mean Difference	Lower	Upper
Difference	-.582	33	.565	-.33265	-1.4956	.8303



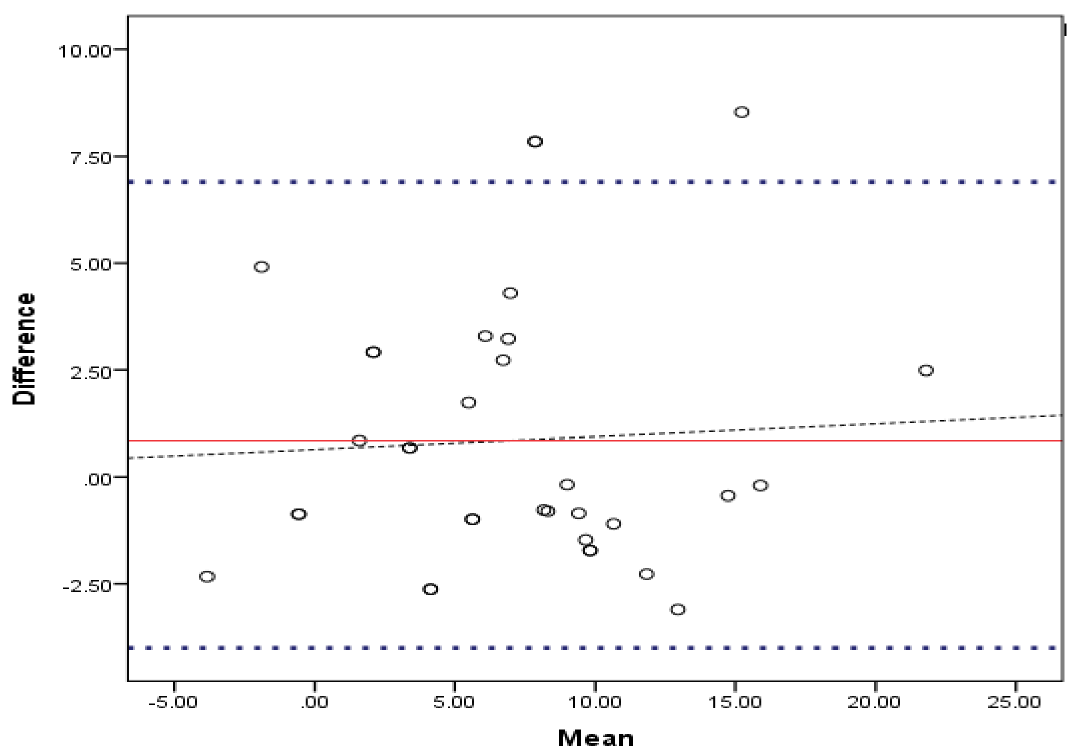
4- Between 2D and 3D HADD during FSLLP left leg

One-Sample Statistics

	N	Mean	Std. Deviation	Std. Error Mean
Difference	34	.8535	3.10994	.53335

One-Sample Test

	Test Value = 0					
					95% Confidence Interval of the Difference	
	t	df	Sig. (2-tailed)	Mean Difference	Lower	Upper
Difference	1.600	33	.119	.85353	-.2316	1.9386



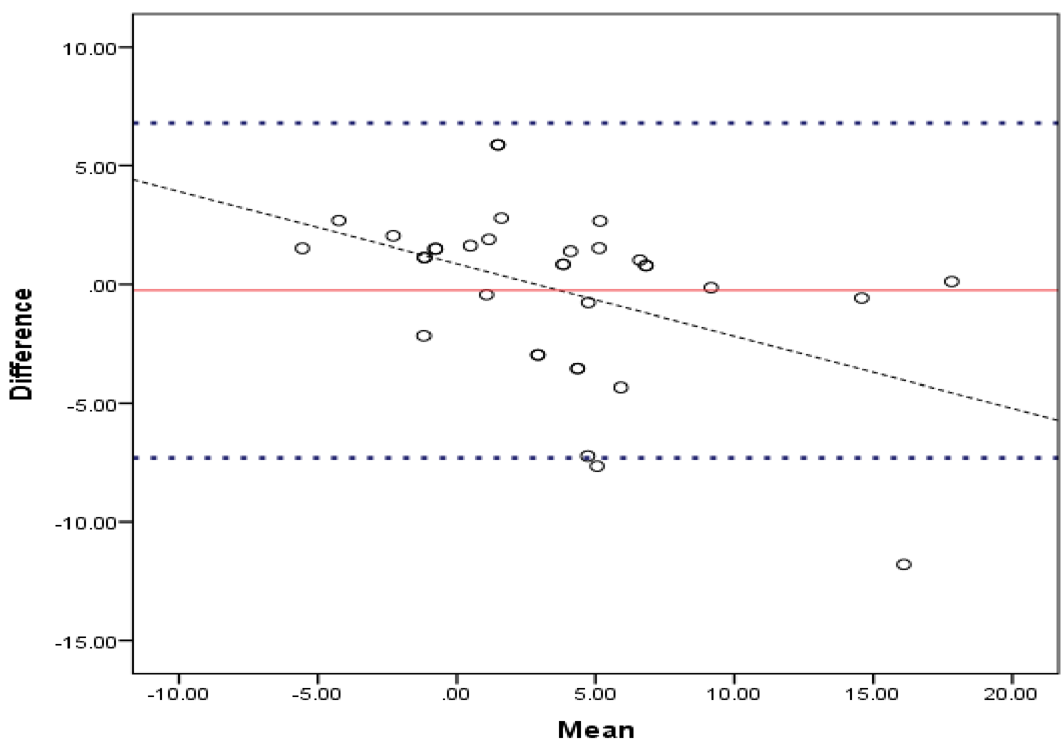
5- Between 2D and 3D HADD during LSLL right leg

One-Sample Statistics

	N	Mean	Std. Deviation	Std. Error Mean
Difference	34	-.2500	3.62901	.62237

One-Sample Test

	Test Value = 0					
					95% Confidence Interval of the Difference	
	t	df	Sig. (2-tailed)	Mean Difference	Lower	Upper
Difference	-.402	33	.691	-.25000	-1.5162	1.0162



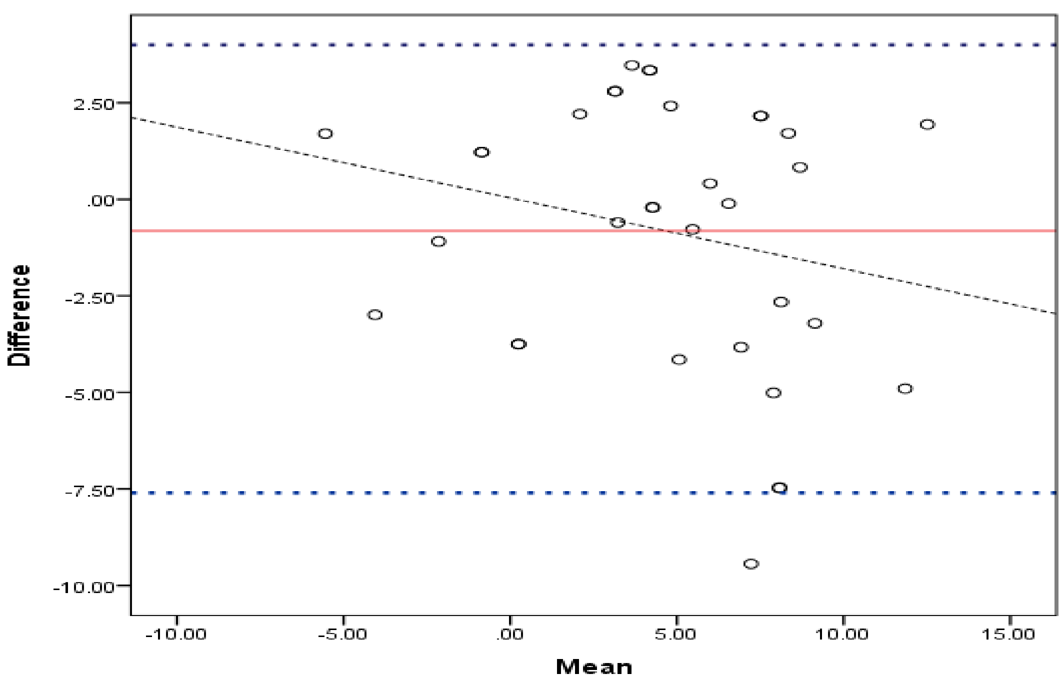
6- Between 2D and 3D HADD during LSLL left leg

One-Sample Statistics

	N	Mean	Std. Deviation	Std. Error Mean
Difference	34	-.8206	3.47737	.59636

One-Sample Test

	Test Value = 0					
					95% Confidence Interval of the Difference	
	t	df	Sig. (2-tailed)	Mean Difference	Lower	Upper
Difference	-1.376	33	.178	-.82059	-2.0339	.3927



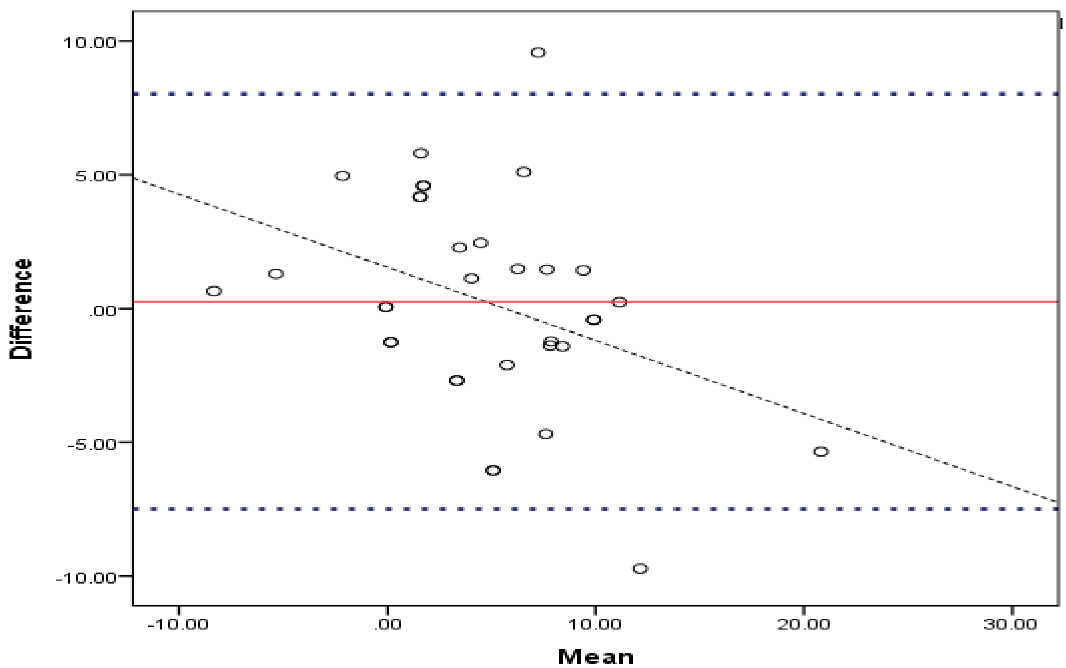
7- Between 2D and 3D HADD during LSLLP right leg

One-Sample Statistics

	N	Mean	Std. Deviation	Std. Error Mean
Difference	34	.2582	3.96130	.67936

One-Sample Test

	Test Value = 0					
					95% Confidence Interval of the Difference	
	t	df	Sig. (2-tailed)	Mean Difference	Lower	Upper
Difference	.380	33	.706	.25824	-1.1239	1.6404



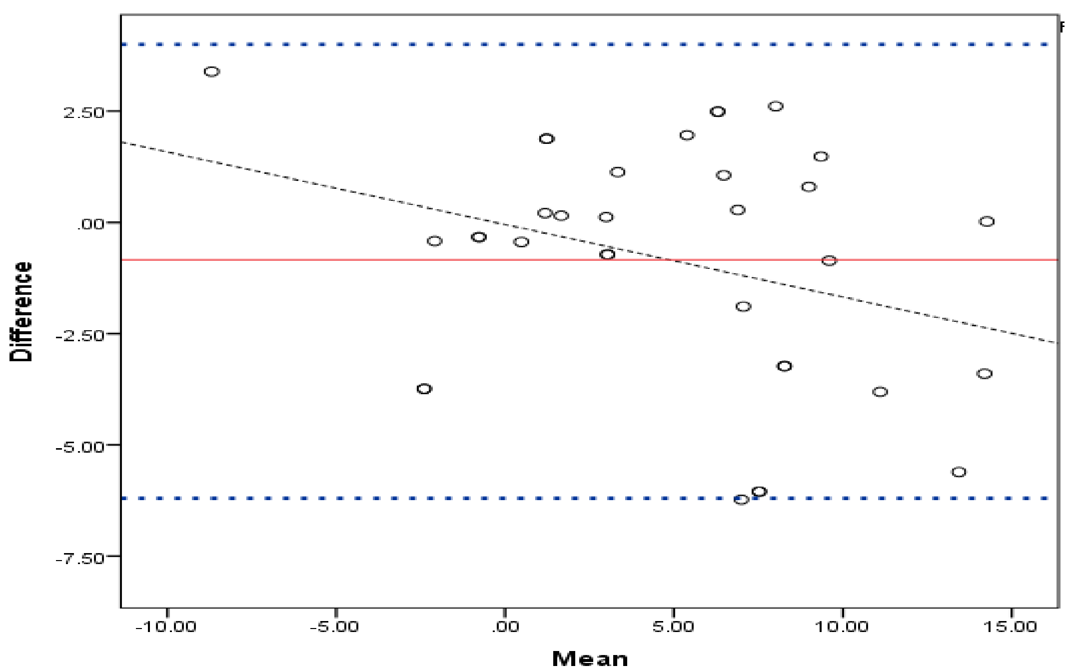
8- Between 2D and 3D HADD during LSLLP left leg

One-Sample Statistics

	N	Mean	Std. Deviation	Std. Error Mean
Difference	34	-.8485	2.73512	.46907

One-Sample Test

	Test Value = 0					
					95% Confidence Interval of the Difference	
	t	df	Sig. (2-tailed)	Mean Difference	Lower	Upper
Difference	-1.809	33	.080	-.84853	-1.8029	.1058



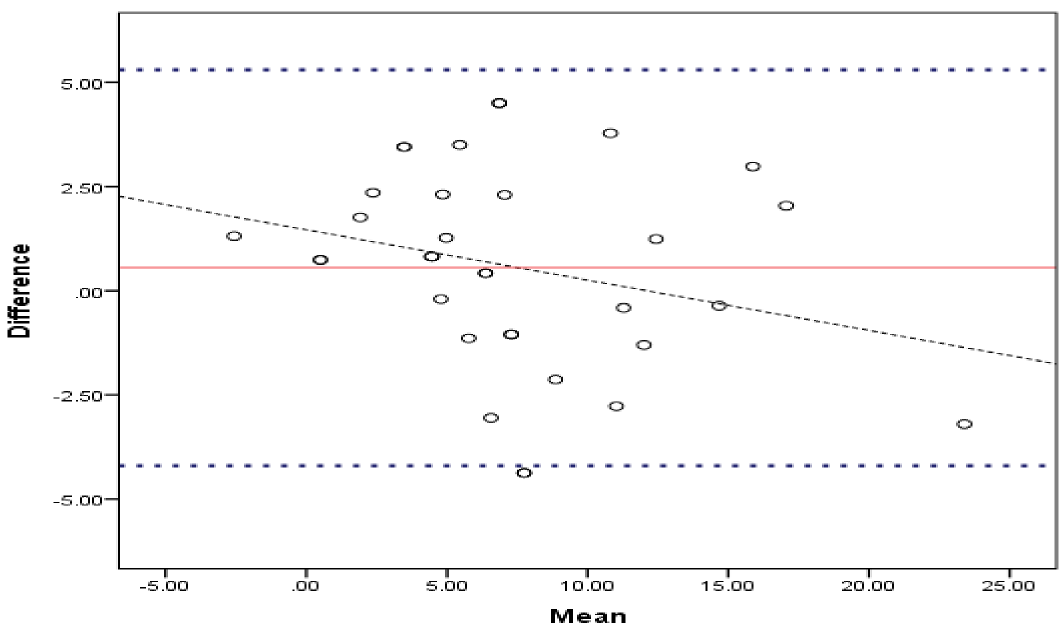
9- Between 2D and 3D HADD during MSLL right leg

One-Sample Statistics

	N	Mean	Std. Deviation	Std. Error Mean
Difference	34	.5676	2.43282	.41722

One-Sample Test

	Test Value = 0					
					95% Confidence Interval of the Difference	
	t	df	Sig. (2-tailed)	Mean Difference	Lower	Upper
Difference	1.361	33	.183	.56765	-.2812	1.4165



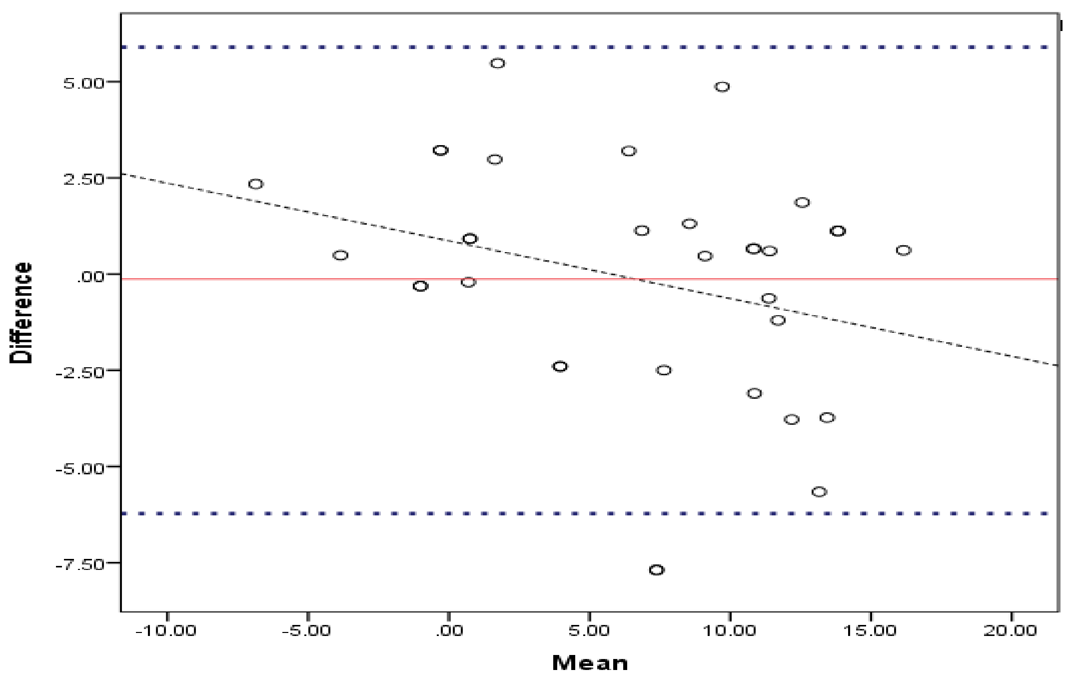
10- Between 2D and 3D HADD during MSLL left leg

One-Sample Statistics

	N	Mean	Std. Deviation	Std. Error Mean
Difference	34	-.1300	3.11009	.53338

One-Sample Test

	Test Value = 0					
					95% Confidence Interval of the Difference	
	t	df	Sig. (2-tailed)	Mean Difference	Lower	Upper
Difference	-.244	33	.809	-.13000	-1.2152	.9552



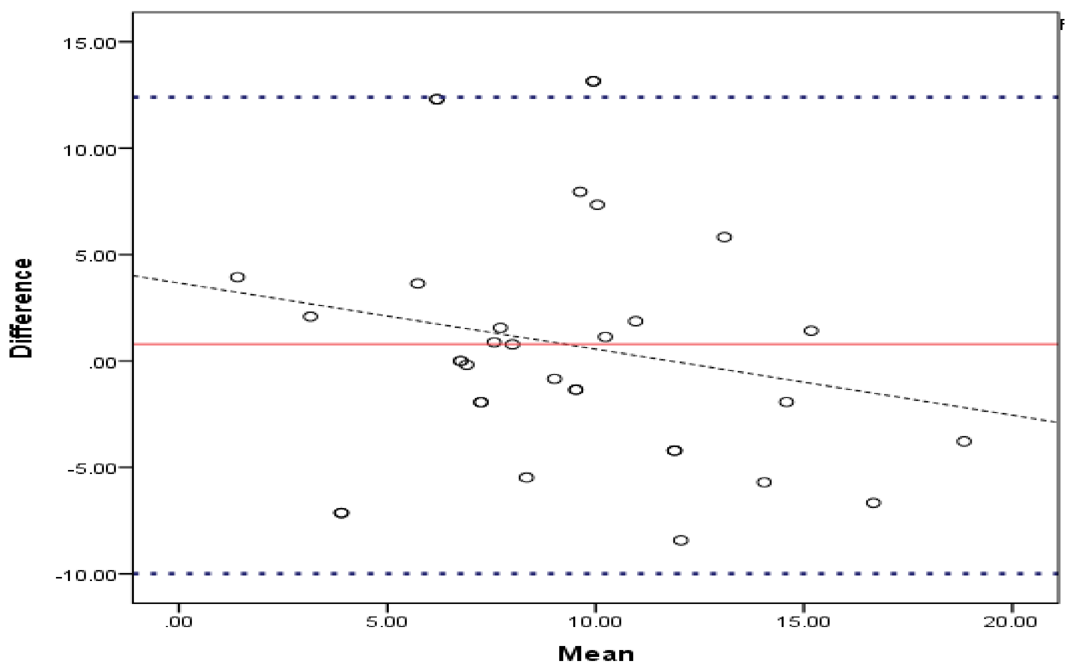
11- Between 2D and 3D HADD during MSLLP right leg

One-Sample Statistics

	N	Mean	Std. Deviation	Std. Error Mean
Difference	34	.7950	5.94683	1.01987

One-Sample Test

	Test Value = 0					
					95% Confidence Interval of the Difference	
	t	df	Sig. (2-tailed)	Mean Difference	Lower	Upper
Difference	.780	33	.441	.79500	-1.2799	2.8699



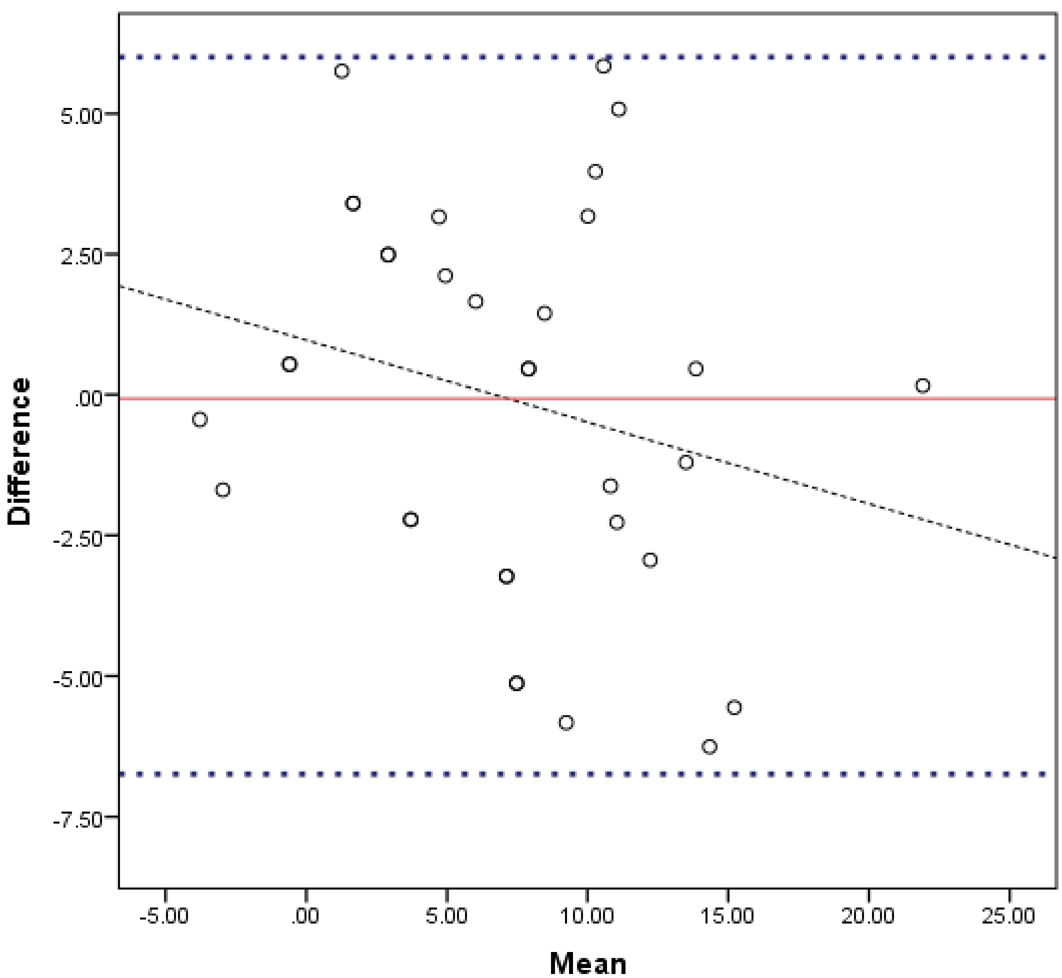
12- Between 2D and 3D HADD during MSLLP left leg

One-Sample Statistics

	N	Mean	Std. Deviation	Std. Error Mean
Difference	34	-.0697	3.40711	.58432

One-Sample Test

	Test Value = 0					
					95% Confidence Interval of the Difference	
	t	df	Sig. (2-tailed)	Mean Difference	Lower	Upper
Difference	-.119	33	.906	-.06971	-1.2585	1.1191



Appendix IX: Correlation between 2D FPPA and 2D HADD angle

Pearson correlation and linear regression between 2D FPPA and 2D HADD angle for right leg during all tasks.

FPPA	2D HADD angle R (P value) (r²)					
	FSLL	FSLLP	LSLL	LSLLP	MSLL	MSLLP
FSLL	**0.58**(<0.001) (0.33)**					
FSLLP		**0.47** (0.005) (0.22)**				
LSLL			**0.68** (<0.001) (0.64)**			
LSLLP				0.33 (0.057) (0.11)		
MSLL					**0.40* (0.20) (0.16)**	
MSLLP						**0.39* (0.023) (0.15)**

FPPA= frontal plane projection angle, HADD= hip adduction angle, FSLL= forward SLL, FSLLP= forward SLL off a platform, LSLL= lateral SLL, LSLLP = lateral SLL off a platform, MSLL = medial SLL, MSLLP = medial SLL off a platform. * Correlation is significant at the 0.05 level, ** Correlation is significant at the 0.01 level

Pearson correlation and linear regression between 2D FPPA and 2D HADD angle for left leg during all tasks.

FPPA	2D HADD angle R (P value) (r²)					
	FSLL	FSLLP	LSLL	LSLLP	MSLL	MSLLP
FSLL	0.31 (0.07) (0.1)					
FSLLP		**0.58** (<0.001) (0.33)**				
LSLL			0.19 (0.28) (0.04)			
LSLLP				0.21 (0.23) (0.04)		
MSLL					**0.70** (<0.001) (0.49)**	
MSLLP						**0.48** (0.004) (0.23)**

FPPA= frontal plane projection angle, HADD= hip adduction angle, FSLL= forward SLL, FSLLP= forward SLL off platform, LSLL= lateral SLL, LSLLP = lateral SLL off a platform, MSLL = medial SLL, MSLLP = medial SLL off a platform. * Correlation is significant at the 0.05 level, ** Correlation is significant at the 0.01 level

Appendix X: Result for normality test for correlation study.

	Shapiro-Wilk		
	Statistic	df	Sig.
2D_FPPA_FSLL_RT	.985	34	.901
2D_FPPA_FSLL_LT	.976	34	.635
2D_FPPA_FSLLP_RT	.955	34	.172
2D_FPPA_FSLLP_LT	.975	34	.610
2D_FPPA_LSLL_RT	.840	34	.132
2D_FPPA_LSLL_LT	.885	34	.112
2D_FPPA_LSLLP_RT	.929	34	.129
2D_FPPA_LSLLP_LT	.937	34	.050
2D_FPPA_MSLL_RT	.985	34	.911
2D_FPPA_MSLL_LT	.949	34	.112
2D_FPPA_MSLLP_RT	.955	34	.175
2D_FPPA_MSLLP_LT	.967	34	.382
2D_HADD_FSLL_RT	.968	34	.403
2D_HADD_FSLL_LT	.980	34	.785
2D_HADD_FSLLP_RT	.905	34	.116
2D_HADD_FSLLP_LT	.968	34	.415
2D_HADD_LSLL_RT	.938	34	.054
2D_HADD_LSLL_LT	.952	34	.137
2D_HADD_LSLLP_RT	.978	34	.718
2D_HADD_LSLLP_LT	.983	34	.869

2D_HADD_MSLL_RT	.929	34	.128
2D_HADD_MSLL_LT	.955	34	.179
2D_HADD_MSLLP_RT	.967	34	.375
2D_HADD_MSLLP_LT	.973	34	.561
3D_KV_FSLL_RT	.930	34	.131
3D_KV_FSLL_LT	.973	34	.548
3D_KV_FSLLP_RT	.968	34	.419
3D_KV_FSLLP_LT	.954	34	.160
3D_KV_LSLL_RT	.973	34	.549
3D_KV_LSLL_LT	.955	34	.177
3D_KV_LSLLP_RT	.976	34	.639
3D_KV_LSLLP_LT	.955	34	.174
3D_KV_MSLL_RT	.959	34	.221
3D_KV_MSLL_LT	.978	34	.697
3D_KV_MSLLP_RT	.962	34	.282
3D_KV_MSLLP_LT	.978	34	.695
3D_HADD_FSLL_RT	.960	34	.241
3D_HADD_FSLL_LT	.934	34	.142
3D_HADD_FSLLP_RT	.958	34	.206
3D_HADD_FSLLP_LT	.983	34	.862
3D_HADD_LSLL_RT	.930	34	.130
3D_HADD_LSLL_LT	.979	34	.734
3D_HADD_LSLLP_RT	.960	34	.249
3D_HADD_LSLLP_LT	.968	34	.406
3D_HADD_MSLL_RT	.951	34	.134

3D_HADD_MSLL_LT	.931	34	.133
3D_HADD_MSLLP_RT	.965	34	.333
3D_HADD_MSLLP_LT	.974	34	.571
HADD_Mom_FSLL_RT	.967	34	.374
HADD_Mom_FSLL_LT	.980	34	.778
HADD_Mom_LSLL_RT	.925	34	.122
HADD_Mom_LSLL_LT	.969	34	.429
HADD_Mom_MSLL_RT	.915	34	.012
HADD_Mom_MSLL_LT	.980	34	.787
HADD_Mom_FSLLP_RT	.941	34	.064
HADD_Mom_FSLLP_LT	.979	34	.744
HADD_Mom_LSLLP_RT	.898	34	.114
HADD_Mom_LSLLP_LT	.977	34	.661
HADD_Mom_MSLLP_RT	.912	34	.110
HADD_Mom_MSLLP_LT	.960	34	.242
knee_EXT_Mom_FSLL_RT	.935	34	.044
knee_EXT_Mom_FSLL_LT	.956	34	.183
knee_EXT_Mom_FSLLP_RT	.982	34	.832
knee_EXT_Mom_FSLLP_LTH_ST_LT_1	.961	34	.257
knee_EXT_Mom_LSLL_RT	.906	34	.117
knee_EXT_Mom_LSLL_LT	.949	34	.118
knee_EXT_Mom_LSLLP_RT	.972	34	.511
knee_EXT_Mom_LSLLP_LT	.947	34	.097

knee_EXT_Mom_MSLL_RT	.970	34	.462
knee_EXT_Mom_MSLL_LT	.969	34	.422
knee_EXT_Mom_MSLLP_RT	.945	34	.087
knee_EXT_Mom_MSLLP_LT	.971	34	.491
KV_MOM_FSLL_RT	.949	34	.115
KV_MOM_FSLL_LT	.964	34	.318
KV_MOM_FSLLP_RT	.872	34	.001
KV_MOM_FSLLP_LT	.877	34	.001
KV_MOM_LSLL_RT	.963	34	.291
KV_MOM_LSLL_LT	.943	34	.074
KV_MOM_LSLLP_RT	.971	34	.496
KV_MOM_LSLLP_LT	.963	34	.295
KV_MOM_MSLL_RT	.945	34	.089
KV_MOM_MSLL_LT	.957	34	.196
KV_MOM_MSLLP_RT	.892	34	.113
KV_MOM_MSLLP_LT	.840	34	.110

Appendix XI: Result of repeated measure ANOVA for right leg 2D FPPA

Tests of Between-Subjects Effects

Measure: MEASURE_1

Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Intercept	14340.866	1	14340.866	135.368	.000
Error	3496.025	33	105.940		

Pairwise Comparisons

Measure: MEASURE_1

(I) FPPA_RT	(J) FPPA_RT	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b]	
					Lower Bound	Upper Bound
1	2	.461	.582	1.000	-1.380	2.303
	3	-2.108	.786	.170	-4.596	.380
	4	-1.590	.658	.320	-3.671	.491
	5	1.789	.808	.509	-.769	4.346
	6	4.475*	1.153	.007	.827	8.122
2	1	-.461	.582	1.000	-2.303	1.380
	3	-2.570*	.724	.018	-4.861	-.278
	4	-2.051	.702	.093	-4.271	.168
	5	1.327	.931	1.000	-1.619	4.273
	6	4.013*	1.198	.030	.223	7.803
3	1	2.108	.786	.170	-.380	4.596
	2	2.570*	.724	.018	.278	4.861
	4	.518	.489	1.000	-1.030	2.067
	5	3.897*	1.064	.013	.530	7.263
	6	6.583*	1.395	.001	2.170	10.996
4	1	1.590	.658	.320	-.491	3.671
	2	2.051	.702	.093	-.168	4.271

	3	-.518	.489	1.000	-2.067	1.030
	5	3.379*	1.014	.032	.171	6.586
	6	6.065*	1.387	.002	1.676	10.454
5	1	-1.789	.808	.509	-4.346	.769
	2	-1.327	.931	1.000	-4.273	1.619
	3	-3.897*	1.064	.013	-7.263	-.530
	4	-3.379*	1.014	.032	-6.586	-.171
	6	2.686	.911	.088	-.198	5.570
6	1	-4.475*	1.153	.007	-8.122	-.827
	2	-4.013*	1.198	.030	-7.803	-.223
	3	-6.583*	1.395	.001	-10.996	-2.170
	4	-6.065*	1.387	.002	-10.454	-1.676
	5	-2.686	.911	.088	-5.570	.198

Based on estimated marginal means

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

b. Adjustment for multiple comparisons: Bonferroni.

Result of repeated measure ANOVA for left leg 2D FPPA

Tests of Between-Subjects Effects

Measure: MEASURE_1

Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Intercept	7811.038	1	7811.038	81.925	.000
Error	3146.363	33	95.344		

Pairwise Comparisons

Measure: MEASURE_1

(I) FPPA_LT	(J) FPPA_LT	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b]	
					Lower Bound	Upper Bound
1	2	1.513	.727	.677	-.786	3.811
	3	-1.641	.594	.140	-3.522	.239
	4	-1.131	.598	1.000	-3.022	.760
	5	1.366	.674	.764	-.768	3.500
	6	3.768*	.692	.000	1.579	5.956
2	1	-1.513	.727	.677	-3.811	.786
	3	-3.154*	.890	.018	-5.971	-.336
	4	-2.644	.988	.173	-5.770	.483
	5	-.146	.816	1.000	-2.729	2.436
	6	2.255*	.678	.033	.109	4.401
3	1	1.641	.594	.140	-.239	3.522
	2	3.154*	.890	.018	.336	5.971
	4	.510	.673	1.000	-1.619	2.640
	5	3.008*	.875	.024	.240	5.776
	6	5.409*	.930	.000	2.465	8.353
4	1	1.131	.598	1.000	-.760	3.022
	2	2.644	.988	.173	-.483	5.770
	3	-.510	.673	1.000	-2.640	1.619
	5	2.497	.819	.067	-.093	5.088
	6	4.899*	.872	.000	2.140	7.657
5	1	-1.366	.674	.764	-3.500	.768
	2	.146	.816	1.000	-2.436	2.729
	3	-3.008*	.875	.024	-5.776	-.240
	4	-2.497	.819	.067	-5.088	.093
	6	2.401*	.737	.039	.071	4.732
6	1	-3.768*	.692	.000	-5.956	-1.579
	2	-2.255*	.678	.033	-4.401	-.109
	3	-5.409*	.930	.000	-8.353	-2.465
	4	-4.899*	.872	.000	-7.657	-2.140
	5	-2.401*	.737	.039	-4.732	-.071

Based on estimated marginal means

*. The mean difference is significant at the 0.05 level.

b. Adjustment for multiple comparisons: Bonferroni.

Result of repeated measure ANOVA for right leg 2D HADD angle

Tests of Between-Subjects Effects

Measure: MEASURE_1

Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Intercept	9327.854	1	9327.854	93.581	.000
Error	3289.317	33	99.676		

Pairwise Comparisons

Measure: MEASURE_1

(I) HADD_RT	(J) HADD_RT	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b]	
					Lower Bound	Upper Bound
1	2	.344	.617	1.000	-1.610	2.298
	3	4.046*	.702	.000	1.824	6.268
	4	2.770*	.650	.002	.713	4.828
	5	-.095	.643	1.000	-2.128	1.938
	6	-2.036	.659	.061	-4.120	.049
2	1	-.344	.617	1.000	-2.298	1.610
	3	3.702*	.692	.000	1.514	5.891
	4	2.426*	.667	.014	.317	4.535
	5	-.439	.631	1.000	-2.436	1.559
	6	-2.379*	.588	.004	-4.241	-.518
3	1	-4.046*	.702	.000	-6.268	-1.824
	2	-3.702*	.692	.000	-5.891	-1.514
	4	-1.276	.725	1.000	-3.571	1.020
	5	-4.141*	.697	.000	-6.347	-1.935
	6	-6.082*	.757	.000	-8.476	-3.688
4	1	-2.770*	.650	.002	-4.828	-.713
	2	-2.426*	.667	.014	-4.535	-.317
	3	1.276	.725	1.000	-1.020	3.571
	5	-2.865*	.717	.005	-5.135	-.596
	6	-4.806*	.733	.000	-7.125	-2.487

5	1	.095	.643	1.000	-1.938	2.128
	2	.439	.631	1.000	-1.559	2.436
	3	4.141*	.697	.000	1.935	6.347
	4	2.865*	.717	.005	.596	5.135
	6	-1.941	.678	.109	-4.086	.205
6	1	2.036	.659	.061	-.049	4.120
	2	2.379*	.588	.004	.518	4.241
	3	6.082*	.757	.000	3.688	8.476
	4	4.806*	.733	.000	2.487	7.125
	5	1.941	.678	.109	-.205	4.086

Based on estimated marginal means

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

b. Adjustment for multiple comparisons: Bonferroni.

Result of repeated measure ANOVA for left leg 2D HADD angle

Tests of Between-Subjects Effects

Measure: MEASURE_1

Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Intercept	7591.450	1	7591.450	64.035	.000
Error	3912.182	33	118.551		

Pairwise Comparisons

Measure: MEASURE_1

(I) HADD_LT	(J) HADD_LT	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b]	
					Lower Bound	Upper Bound
1	2	-.823	.440	1.000	-2.215	.569
	3	2.408*	.583	.003	.564	4.252
	4	2.186*	.668	.037	.074	4.299
	5	.111	.750	1.000	-2.261	2.482
	6	-.444	.608	1.000	-2.369	1.481
2	1	.823	.440	1.000	-.569	2.215
	3	3.231*	.763	.003	.818	5.644
	4	3.009*	.797	.009	.488	5.531
	5	.934	.795	1.000	-1.582	3.449
	6	.379	.640	1.000	-1.647	2.405
3	1	-2.408*	.583	.003	-4.252	-.564
	2	-3.231*	.763	.003	-5.644	-.818
	4	-.221	.451	1.000	-1.649	1.206
	5	-2.297*	.659	.021	-4.382	-.213
	6	-2.852*	.713	.005	-5.109	-.595
4	1	-2.186*	.668	.037	-4.299	-.074
	2	-3.009*	.797	.009	-5.531	-.488
	3	.221	.451	1.000	-1.206	1.649
	5	-2.076	.685	.071	-4.243	.091
	6	-2.631*	.732	.016	-4.946	-.315
5	1	-.111	.750	1.000	-2.482	2.261
	2	-.934	.795	1.000	-3.449	1.582
	3	2.297*	.659	.021	.213	4.382
	4	2.076	.685	.071	-.091	4.243
	6	-.555	.686	1.000	-2.725	1.616
6	1	.444	.608	1.000	-1.481	2.369
	2	-.379	.640	1.000	-2.405	1.647
	3	2.852*	.713	.005	.595	5.109
	4	2.631*	.732	.016	.315	4.946
	5	.555	.686	1.000	-1.616	2.725

Based on estimated marginal means

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

b. Adjustment for multiple comparisons: Bonferroni.

Result of repeated measure ANOVA for right leg 3D knee valgus angle

Tests of Between-Subjects Effects

Measure: MEASURE_1

Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Intercept	1186.887	1	1186.887	12.748	.001
Error	3072.333	33	93.101		

Pairwise Comparisons

Measure: MEASURE_1

(I) KV_3D_RT	(J) KV_3D_RT	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b]	
					Lower Bound	Upper Bound
1	2	.534	.453	1.000	-.900	1.967
	3	2.135*	.353	.000	1.019	3.252
	4	1.485*	.443	.031	.082	2.887
	5	.662	.469	1.000	-.823	2.147
	6	1.792*	.564	.048	.008	3.576
2	1	-.534	.453	1.000	-1.967	.900
	3	1.602*	.471	.027	.112	3.091
	4	.951	.361	.189	-.190	2.092
	5	.128	.614	1.000	-1.816	2.073
	6	1.258	.477	.190	-.252	2.768
3	1	-2.135*	.353	.000	-3.252	-1.019
	2	-1.602*	.471	.027	-3.091	-.112
	4	-.651	.461	1.000	-2.110	.809
	5	-1.473*	.454	.040	-2.910	-.036
	6	-.344	.551	1.000	-2.087	1.400
4	1	-1.485*	.443	.031	-2.887	-.082
	2	-.951	.361	.189	-2.092	.190
	3	.651	.461	1.000	-.809	2.110
	5	-.823	.619	1.000	-2.782	1.137

	6	.307	.554	1.000	-1.445	2.059
5	1	-.662	.469	1.000	-2.147	.823
	2	-.128	.614	1.000	-2.073	1.816
	3	1.473*	.454	.040	.036	2.910
	4	.823	.619	1.000	-1.137	2.782
	6	1.130	.462	.301	-.333	2.592
6	1	-1.792*	.564	.048	-3.576	-.008
	2	-1.258	.477	.190	-2.768	.252
	3	.344	.551	1.000	-1.400	2.087
	4	-.307	.554	1.000	-2.059	1.445
	5	-1.130	.462	.301	-2.592	.333

Based on estimated marginal means

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

b. Adjustment for multiple comparisons: Bonferroni.

Result of repeated measure ANOVA for left leg 3D knee valgus angle

Tests of Between-Subjects Effects

Measure: MEASURE_1

Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Intercept	534.876	1	534.876	7.315	.011
Error	2412.821	33	73.116		

Pairwise Comparisons

Measure: MEASURE_1

(I) KV_3D_LT	(J) KV_3D_LT	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b]	
					Lower Bound	Upper Bound
1	2	.775	.331	.379	-.271	1.821
	3	1.932*	.295	.000	.999	2.865

	4	1.517*	.369	.004	.348	2.685
	5	.393	.312	1.000	-.593	1.379
	6	1.790*	.499	.016	.210	3.369
2	1	-.775	.331	.379	-1.821	.271
	3	1.157*	.331	.021	.109	2.205
	4	.742	.413	1.000	-.565	2.048
	5	-.382	.342	1.000	-1.464	.700
	6	1.015	.438	.406	-.373	2.402
3	1	-1.932*	.295	.000	-2.865	-.999
	2	-1.157*	.331	.021	-2.205	-.109
	4	-.415	.383	1.000	-1.628	.797
	5	-1.539*	.373	.004	-2.721	-.358
	6	-.142	.460	1.000	-1.598	1.314
4	1	-1.517*	.369	.004	-2.685	-.348
	2	-.742	.413	1.000	-2.048	.565
	3	.415	.383	1.000	-.797	1.628
	5	-1.124	.376	.079	-2.314	.066
	6	.273	.418	1.000	-1.051	1.597
5	1	-.393	.312	1.000	-1.379	.593
	2	.382	.342	1.000	-.700	1.464
	3	1.539*	.373	.004	.358	2.721
	4	1.124	.376	.079	-.066	2.314
	6	1.397*	.319	.002	.387	2.407
6	1	-1.790*	.499	.016	-3.369	-.210
	2	-1.015	.438	.406	-2.402	.373
	3	.142	.460	1.000	-1.314	1.598
	4	-.273	.418	1.000	-1.597	1.051
	5	-1.397*	.319	.002	-2.407	-.387

Based on estimated marginal means

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

b. Adjustment for multiple comparisons: Bonferroni.

Result of repeated measure ANOVA for right leg 3D HADD angle

Tests of Between-Subjects Effects

Measure: MEASURE_1

Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Intercept	8372.514	1	8372.514	49.278	.000
Error	5606.852	33	169.905		

Pairwise Comparisons

Measure: MEASURE_1

(I) HADD_3D_RT	(J) HADD_3D_RT	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b]	
					Lower Bound	Upper Bound
1	2	-1.084	.532	.742	-2.766	.598
	3	2.701*	.548	.000	.967	4.435
	4	1.933*	.557	.022	.170	3.696
	5	-.623	.457	1.000	-2.069	.823
	6	-2.336*	.665	.020	-4.441	-.230
2	1	1.084	.532	.742	-.598	2.766
	3	3.785*	.614	.000	1.841	5.729
	4	3.017*	.604	.000	1.107	4.927
	5	.461	.583	1.000	-1.384	2.307
	6	-1.252	.731	1.000	-3.566	1.063
3	1	-2.701*	.548	.000	-4.435	-.967
	2	-3.785*	.614	.000	-5.729	-1.841
	4	-.768	.751	1.000	-3.145	1.609
	5	-3.324*	.569	.000	-5.123	-1.524
	6	-5.037*	.746	.000	-7.399	-2.675
4	1	-1.933*	.557	.022	-3.696	-.170
	2	-3.017*	.604	.000	-4.927	-1.107
	3	.768	.751	1.000	-1.609	3.145
	5	-2.556*	.652	.006	-4.619	-.492
	6	-4.269*	.594	.000	-6.147	-2.391
5	1	.623	.457	1.000	-.823	2.069
	2	-.461	.583	1.000	-2.307	1.384
	3	3.324*	.569	.000	1.524	5.123

	4	2.556*	.652	.006	.492	4.619
	6	-1.713	.683	.259	-3.875	.448
6	1	2.336*	.665	.020	.230	4.441
	2	1.252	.731	1.000	-1.063	3.566
	3	5.037*	.746	.000	2.675	7.399
	4	4.269*	.594	.000	2.391	6.147
	5	1.713	.683	.259	-.448	3.875

Based on estimated marginal means

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

b. Adjustment for multiple comparisons: Bonferroni.

Result of repeated measure ANOVA for left leg 3D HADD angle

Tests of Between-Subjects Effects

Measure: MEASURE_1

Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Intercept	7631.508	1	7631.508	45.642	.000
Error	5517.690	33	167.203		

Pairwise Comparisons

Measure: MEASURE_1

(I) HADD_3D_LT	(J) HADD_3D_LT	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b]	
					Lower Bound	Upper Bound
1	2	-.888	.587	1.000	-2.747	.970
	3	.669	.428	1.000	-.687	2.024
	4	.419	.582	1.000	-1.421	2.260
	5	-.938	.602	1.000	-2.844	.967
	6	-1.433	.633	.454	-3.435	.569
2	1	.888	.587	1.000	-.970	2.747

	3	1.557	.505	.062	-.041	3.154
	4	1.307	.640	.739	-.719	3.334
	5	-.050	.760	1.000	-2.454	2.354
	6	-.544	.543	1.000	-2.262	1.173
3	1	-.669	.428	1.000	-2.024	.687
	2	-1.557	.505	.062	-3.154	.041
	4	-.249	.449	1.000	-1.671	1.172
	5	-1.607	.648	.276	-3.657	.444
	6	-2.101*	.628	.031	-4.087	-.115
4	1	-.419	.582	1.000	-2.260	1.421
	2	-1.307	.640	.739	-3.334	.719
	3	.249	.449	1.000	-1.172	1.671
	5	-1.357	.786	1.000	-3.845	1.130
	6	-1.852	.618	.078	-3.808	.105
5	1	.938	.602	1.000	-.967	2.844
	2	.050	.760	1.000	-2.354	2.454
	3	1.607	.648	.276	-.444	3.657
	4	1.357	.786	1.000	-1.130	3.845
	6	-.494	.698	1.000	-2.704	1.715
6	1	1.433	.633	.454	-.569	3.435
	2	.544	.543	1.000	-1.173	2.262
	3	2.101*	.628	.031	.115	4.087
	4	1.852	.618	.078	-.105	3.808
	5	.494	.698	1.000	-1.715	2.704

Based on estimated marginal means

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

b. Adjustment for multiple comparisons: Bonferroni.

Result Friedman test for right leg knee valgus moment

Test Statistics[a]

N	34
Chi-Square	49.159
df	5
Asymp. Sig.	.000

a. Friedman Test

Result of Friedman test for left leg knee valgus moment

Test Statistics[a]

N	34
Chi-Square	25.342
df	5
Asymp. Sig.	.000

a. Friedman Test

Result of Friedman test for right leg HADD moment

Test Statistics[a]

N	34
Chi-Square	31.425
df	5
Asymp. Sig.	.000

a. Friedman Test

Result of repeated measure ANOVA for left leg HADD moment

Tests of Between-Subjects Effects

Measure: MEASURE_1

Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Intercept	661.644	1	661.644	2300.634	.000
Error	9.491	33	.288		

Pairwise Comparisons

Measure: MEASURE_1

(I) HADD_MOM_LT	(J) HADD_MOM_LT	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b] Lower Bound	Upper Bound
1	2	.309*	.048	.000	.156	.462
	3	.149	.062	.336	-.048	.346
	4	.329*	.056	.000	.152	.506
	5	.010	.051	1.000	-.151	.170
	6	.330*	.064	.000	.127	.532
2	1	-.309*	.048	.000	-.462	-.156
	3	-.160	.063	.237	-.358	.039
	4	.020	.054	1.000	-.151	.191
	5	-.299*	.056	.000	-.476	-.122
	6	.021	.067	1.000	-.191	.233
3	1	-.149	.062	.336	-.346	.048
	2	.160	.063	.237	-.039	.358
	4	.180*	.048	.011	.027	.332
	5	-.139	.050	.122	-.296	.017
	6	.181	.064	.116	-.021	.382
4	1	-.329*	.056	.000	-.506	-.152
	2	-.020	.054	1.000	-.191	.151
	3	-.180*	.048	.011	-.332	-.027
	5	-.319*	.051	.000	-.481	-.157
	6	.001	.058	1.000	-.181	.183
5	1	-.010	.051	1.000	-.170	.151
	2	.299*	.056	.000	.122	.476
	3	.139	.050	.122	-.017	.296
	4	.319*	.051	.000	.157	.481
	6	.320*	.063	.000	.121	.519
6	1	-.330*	.064	.000	-.532	-.127
	2	-.021	.067	1.000	-.233	.191
	3	-.181	.064	.116	-.382	.021
	4	-.001	.058	1.000	-.183	.181
	5	-.320*	.063	.000	-.519	-.121

Based on estimated marginal means

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

b. Adjustment for multiple comparisons: Bonferroni.

Result of Friedman test for right leg knee extensor moment

Test Statistics[a]

N	34
Chi-Square	96.138
df	5
Asymp. Sig.	.000

a. Friedman Test

Result of repeated measure ANOVA for left leg knee extensor moment

Tests of Between-Subjects Effects

Measure: MEASURE_1

Transformed Variable: Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Intercept	1322.382	1	1322.382	1066.204	.000
Error	40.929	33	1.240		

Pairwise Comparisons

Measure: MEASURE_1

(I) Knee_ext_mom_LT	(J) Knee_ext_mom_LT	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b]	
					Lower Bound	Upper Bound
1	2	-.284*	.071	.005	-.510	-.058
	3	.394*	.055	.000	.219	.570
	4	-.047	.069	1.000	-.264	.170
	5	.423*	.060	.000	.235	.612
	6	-.170	.090	1.000	-.454	.114

2	1	.284*	.071	.005	.058	.510
	3	.678*	.075	.000	.439	.917
	4	.237	.086	.138	-.034	.508
	5	.707*	.079	.000	.458	.956
	6	.114	.105	1.000	-.220	.447
3	1	-.394*	.055	.000	-.570	-.219
	2	-.678*	.075	.000	-.917	-.439
	4	-.441*	.057	.000	-.623	-.259
	5	.029	.057	1.000	-.152	.209
	6	-.565*	.075	.000	-.803	-.326
4	1	.047	.069	1.000	-.170	.264
	2	-.237	.086	.138	-.508	.034
	3	.441*	.057	.000	.259	.623
	5	.470*	.072	.000	.242	.698
	6	-.123	.072	1.000	-.352	.105
5	1	-.423*	.060	.000	-.612	-.235
	2	-.707*	.079	.000	-.956	-.458
	3	-.029	.057	1.000	-.209	.152
	4	-.470*	.072	.000	-.698	-.242
	6	-.593*	.081	.000	-.849	-.337
6	1	.170	.090	1.000	-.114	.454
	2	-.114	.105	1.000	-.447	.220
	3	.565*	.075	.000	.326	.803
	4	.123	.072	1.000	-.105	.352
	5	.593*	.081	.000	.337	.849

Based on estimated marginal means

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

b. Adjustment for multiple comparisons: Bonferroni.