The effect of change of direction angle on knee and hip biomechanics: implications for anterior cruciate ligament injury

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DEDICATION

I dedicated this thesis to my parents:

Abdullah Alhammad and Heiam Alaqrabawi

To my wife:

Nashwa Hazza

And to my children

Abdullah, Juri, Abdullmalik and Tulin

This is very much a shared accomplishment

that would not have been achieved without your support.

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List of Abbreviations

COD	Change of direction
ACL	Anterior cruciate ligament
3D	Three-dimensional
ACLR	Anterior cruciate ligament reconstruction
CI	Confidence intervals
EKAM	External knee abduction moment
PEKAM	Peak external knee abduction moment
GRF	Ground reaction force
VGRF	Vertical ground reaction force
PVGRF	Peak vertical ground reaction force
IC	Initial contact
ICC	Intra class correlation coefficient
KAM	Knee abduction moment
PKVA	Peak knee valgus angle
PKFA	Peak knee flexion angle
MVIC	Maximal voluntary isometric contraction
PT	Peak torque
SEM	Standard error of measurement
SD	Standard deviation
60 ms	60 milliseconds
ROM	Range of motion
RTS	Return to sport

Abstract

ACL injuries have been referred to poor mechanics as they frequently occur without contact. Changes in the knee valgus (abduction) angle and knee valgus (external abduction) moment and limb asymmetry have been linked to greater risk of ACL injury. Change of direction (COD) manoeuvres are important for many field sports, however they are unfortunately associated with non-contact anterior cruciate ligament (ACL) injuries. There is limited literature exploring the associations between lower-limb biomechanical variables during COD manoeuvre associated with ACL injuries. Although players frequently COD at >90° angles, limited knowledge is available on hip and knee joints kinematics and kinetics in term of limb asymmetry and differences between COD at 90° and 135° manoeuvres. In addition, high knee valgus angle and moment during COD manoeuvre is associated with joint positions including increased hip flexion, abduction and internal rotation angles. In addition, isometric hip muscle strength has been reported to predict ACL injuries, indicating that weakness in hip muscles is a modifiable risk factor of the non-contact ACL injury. However, the relationship between knee valgus angle and moment with hip kinematics and muscle strength during COD at 90° and 135° manoeuvres still unknown. Currently, there has been no published research correlating the hip abductor, extensor, and external rotator strength on frontal plane hip and knee biomechanics during 90° and 135° COD manoeuvres. Therefore, the purposes of this thesis was to (1) determine whether asymmetry in knee and hip biomechanics kinematics and kinetics and hip muscle strength between preferred and non-preferred limbs during COD manoeuvres at 90° and 135° angles exists, (2) determine whether differences in knee and hip biomechanics kinematics and kinetics between COD manoeuvres at 90° and 135° angles exists and (3) explore the relationships between ACL injury risk factors (knee valgus angle and moment) and hip kinematics and muscles strength during 90° and 135° COD manoeuvres.

Three-dimensional (3D) motion analysis technique is a gold stander to assess biomechanical lower-limb during functional activities. In fact, the gold standard for examining lower limb biomechanics is 3D motion analysis system and allows researchers to calculate all three motion planes during dynamic manoeuvres. In addition, the isokinetic dynamometer has been considered as a gold standard measurement tool for assessing isometric hip muscle strength and become more

popular in sport, research and clinic setting. Healthy male recreational soccer players performed COD manoeuvres at 90° and 135° angles and maximal voluntary isometric contractions of the hip abductors, extensors, and external rotators. From recorded motion capture, ground reaction force data and hip and knee biomechanics as well as, hip muscles peak torque were calculated. To determine limb asymmetry, a paired sample t-test was conducted using a Holm method correction. Then, pearson's correlation coefficient (r) was used to explore the relationships between hip kinematics and strength and knee valgus angle and moment.

36 individuals took part in the study (24.25 ±6.21 years, 1.72 ±0.06 m and 66.41 ±10.83 kg). At 135° COD, participants showed greater knee valgus angles at initial contact and greater peak external knee abduction moments than at 90°. However, no effect of COD angles on knee flexion angle and peak vGRF were found. The results suggest that there were no differences between preferred and non-preferred limbs. Furthermore, the results highlight an important role the hip motion play in controlling kinematic and kinetic risk factors of ACL injury during COD manoeuvres. The findings provided some support that excessive knee valgus angle and moment is potentially associated with poor hip control in all planes. However, there were no significant correlation between hip muscles strength and knee frontal plane kinetics and kinematics.

It can be concluded that different COD angles demand different hip and knee kinematics and kinetics. The results suggest that sharper COD angle place the knee at more risk for ACL injuries. COD manoeuvres at 90° may be useful for evaluating of individuals but may not be challenging enough to reveal poor neuromuscular control over hip and knee motion. Therefore, sharper angles of examination should be utilized in the evaluation of individuals. Moreover, these results may help provide an appropriate manipulation and intervention on COD manoeuvre to reduce the risk of ACL injury. The findings of this study will increase the knowledge base of ACL injury and can aid in the design of more appropriate neuromuscular and plyometric training protocols for injury prevention.

1 Chapter 1: Introduction

1.1 Background:

Knee injuries are common amongst the sporting population, especially for sports that require changes in direction, stopping and starting, and jumping and landing movements. The largest and most nationally representative epidemiological study of knee injuries in the United States of America (USA) found that from 2005/06 to 2010/11, 5,116 knee injuries were recorded with the most commonly injuries sustained to the medial collateral ligament (MCL) (36.1%), followed by the anterior cruciate ligament (ACL) (25.4%), lateral collateral ligament (LCL) (7.9%), and posterior cruciate ligament (PCL) (2.4%) (Swenson et al., 2013). Anterior cruciate ligament (ACL) injuries are common, as approximately 130,000 primary ACL reconstructive surgeries are performed annually in the USA (Mall et al., 2014). The incidence of ACL reconstruction in the USA rose from 86,687 (32.9 per 100,000 person-years) in 1994 to 129,836 (43.5 per 100,000 person-years) in 2006 (Mall et al., 2014). In England, there were 133,270 cases of ACL reconstruction (124,489 patients) between 1997–1998 and 2016–2017, and the rate of ACL reconstruction increased 12 times from 2 per 100,000 person-years in 1997–1998 to 24.2 per 100,000 person-years in 2016–2017 (Abram, Price, Judge, & Beard, 2019). When considering sporting injuries, ACL injuries have a high financial impact both personally (lost working time) and professionally (lost playing time). Thus, ruptures to the ACL are considered one of the costliest knee injuries in sport (Mather et al., 2013). The lifetime burden of ACL tears in the U.S. was estimated to be $7.6 billion annually (expressed as the net present value) when treated with ACL reconstruction, and $17.7 billion annually when treated with rehabilitation (Mather et al., 2013). Furthermore, the financial impact of ACL reconstruction every year was estimated to be 17.4 million New Zealand Dollar in New Zealand and Australia (Gianotti, Marshall, Hume, & Bunt, 2009; Janssen, Orchard, Driscoll, & van Mechelen, 2012). Apart from the cost, more worryingly, injuries to the ACL sometimes have devastating results in the long term, as athletes are forced to lower their levels of activity, and there is up to 25 percent risk of re-injury on return to competitive sport, as well as an increased risk of osteoarthritis (Eckstein, Wirth, Lohmander, Hudelmaier, & Frobell, 2015; Paterno et al., 2010). Moreover, only 55% of athletes returned to competitive sport within three years after ACL reconstruction (ACLR) (Ardern, Taylor, Feller, & Webster, 2014).

In general, 70-85% of ACL injuries occur during non-contact situations (not as a result of a direct blow to the knee), and almost all occur during either a change of direction (COD) manoeuvre or a single leg landing (SLL) (Cochrane, Lloyd, Buttfield, Seward, & McGivern, 2007; Johnston et al., 2018; Koga et al., 2010; Waldén et al., 2015). From an injury prevention perspective, as the majority of ACL injuries occur during non-contact situations during COD manoeuvres, it is possible to prevent or reduce the risk of ACL injuries by understanding non-contact ACL injury mechanisms and risk. According to both self-reported studies and video analysis, ACL injuries typically happen during movements that require a sudden deceleration and a change in direction, or involve a jump, just after the foot comes into contact with the ground (Benis, Torre, & Bonato, 2018; Grassi et al., 2017; Johnston et al., 2018; Montgomery et al., 2018). ACL injuries have been linked with poor mechanics as they frequently occur without contact. Various risk factors have been reported to be associated with the increased risk of ACL injuries, particular; any alteration in the sagittal and frontal plane biomechanics and muscle strength (Hughes, 2014). It has been postulated that changes in the knee valgus (abduction) angle and knee valgus (external abduction) moment are thought to increase risk of non-contact ACL injury (Hewett et al., 2005; Krosshaug et al., 2007; Pollard, Stearns, Hayes, & Heiderscheit, 2015). The risk of an ACL injury is increased when there is an extended knee combined with an increased knee valgus angle and moment, and with internal tibia rotation (McLean, Huang, Su, & Van Den Bogert, 2004; Montgomery et al., 2016; Myer et al., 2015; Olsen, Myklebust, Engebretsen, & Bahr, 2004). Montgomery et al. (2016) describe the mechanism of 36 ACL injuries from rugby games played in top professional leagues and international matches using systematic video analysis. They found that non-contact ACL injuries had lower knee flexion angles than the control group (10° vs 20°). Similarly, increased knee abduction moment with less knee flexion (less than 30 degree) increases the ACL strain measured in cadaveric knee models (Markolf et al., 1995). In addition, in a prospective study of 205 women's soccer, basketball and volleyball players, Myer and colleagues (2015) demonstrated that a knee abduction moment of 25.25 Nm during landing is the most sensitive and specific threshold to dichotomise those who suffered non-contact ACL injury from those who did not. With peak external knee abduction moment observed during the weight acceptance (the first 20-30% of the stance phase) of COD manoeuvres (Besier, Lloyd, Cochrane, & Ackland, 2001b), this is when ACL injury risk is thought to be greatest. Furthermore, ACL injury videos studies found that the estimated time of injury ranged between 17 and 60 milliseconds after the initial contact (IC) (Koga et al., 2010; Krosshaug et al., 2007b). In addition, simulation study suggest that noncontact ACL injuries are expected to occur between 48 and 61 milliseconds after initial contact during simulated landing and injury events (Bates, Schilaty, Ueno, & Hewett, 2020).

Soccer is generally considered the most popular sport in the world (Dvorak, Junge, Graf-Baumann, & Peterson, 2004; Junge & Dvorak, 2015). Professional soccer players are required to perform a large amount of COD manoeuvres within high intensity movement during competitive matches; this involves numerous directional changes from various angles (Ade, Fitzpatrick, & Bradley, 2016; Bloomfield, Polman, & O'Donoghue, 2007; Robinson, O'Donoghue, & Wooster, 2011; Taylor, Wright, Dischiavi, Townsend, & Marmon, 2017). Taylor et al. (2017) performed a systematic review to evaluate literature that characterizes, quantifies, and compares the demands of multi-directional sports. Their study revealed that the most frequent COD manoeuvres were performed during soccer games with up to 800 CODs per game. Notational analysis has been used with English Premier league football players, and this revealed that, on average, players carried out 727 turns/swerves during a 90 minute game, approximately eight turns/swerves every minute to the right or left, and the performance of 100 CODs at angles between 90° and 180° across all positions (Bloomfield et al., 2007). However, many of the CODs recorded in the study by Bloomfield et al. (2007) would have been performed at low intensities during periods of "ball related movement". In comparison, the investigation by Robinson et al. (2011) included all CODs performed at speeds of 4 m/second. Robinson et al. (2011) reported 233 CODs of 45°–135° and 61 CODs ≥ 135° per match, made to the left and right and across all positions in English FA Premier League soccer matches. Furthermore, elite soccer athletes perform up to 32% of directional changes at 90°-180° (Ade et al., 2016). Moreover, over two seasons (2016 and 2017) during home matches with the first team in a Norwegian elite football club, 90˚ to 180˚ COD angles were performed more often by all positions (Baptista, Johansen, Seabra, & Pettersen, 2018). In the aforementioned literature, studies have shown how frequently COD manoeuvres at angles between 45° and 135° have been used during soccer. However, COD manoeuvres are unfortunately associated with non-contact anterior cruciate ligament (ACL) injuries at varies COD angles (30°-180°) (Montgomery et al., 2018; Waldén et al., 2015).

Furthermore, data has shown that COD manoeuvres at 90° have a greater risk of ACL injury compared to single-leg landing tasks. For example, 90° COD manoeuvres resulted in significantly greater knee abduction moments compared with single-legged landings, which were four times greater (Jones, Herrington, Munro, & Graham-Smith, 2014).

Moreover, data has shown that COD movements at small angles (45°) have a greater risk of ACL injury compared to single-leg landings (single-leg drop landings, single-leg countermovement jumps and single-leg jump landings) (Chinnasee, Weir, Sasimontonkul, Alderson, & Donnelly, 2018). External knee abduction moments were 8, 6 and 2 times higher in 45° COD compared with single-leg drop landings, single-leg countermovement jumps and single-leg jump landings, respectively (0.93 ± 0.53 Nm/kg vs 0.15 ± 0.07 Nm/kg, 0.12 ± 0.10 Nm/kg and 0.45 ± 0.27 Nm/kg, respectively). In addition, the external knee abduction moment was found to be six times higher in 45° COD manoeuvres compared to the drop jump (1.58 ± 0.60 Nm/kg vs 0.25 ± 0.16 Nm/kg) (Kristianslund & Krosshaug, 2013). These results suggest that COD manoeuvres have a greater risk of ACL injury compared to single-leg landing tasks. In addition, using a landing task to screen athletes may not be as beneficial to soccer in which COD manoeuvres are more common (Faude, Junge, Kindermann, & Dvorak, 2005). Therefore, the COD manoeuvre has become our riskiest measure to determine ACL injury risks in sports involving a large amount of such movements.

As previously mentioned, soccer players are required to perform a diverse range of COD angles ranging between 45° and 135° during match games (Ade et al., 2016; Bloomfield et al., 2007; Robinson et al., 2011; Taylor et al., 2017). A plethora of biomechanical investigations have investigated a range of angled direction changes (30°–135°) to provide an understanding of the biomechanical risk factors associated with increased ACL injury risk. However, the majority of the studies that have evaluated COD manoeuvres are limited to smaller change of direction angles (30, 45, 60 degree) (Dempsey et al., 2007; Frank et al., 2013; Kristianslund, Faul, Bahr, Myklebust, & Krosshaug, 2014; McLean, Huang, & van den Bogert, 2005; Sigward, Cesar, & Havens, 2015; Sigward & Powers, 2007), with some analysing COD manoeuvres at a 90° angle (Havens & Sigward, 2015a; Jones, Herrington, & Graham-Smith, 2015) and only one study investigating COD manoeuvres at 135° angles (Schreurs, Benjaminse, & Lemmink, 2017). COD at 45° is a common manoeuvre for assessing knee kinetics and kinematics for ACL injury; however the mechanical demands placed on the knee increase for COD manoeuvres at angles greater than 45°. For example, Sigward et al. (2015) found that athletes displayed greater knee valgus moments, greater hip abduction angles and more ground reaction force when performing COD at a 110° compared to a 45° angle. Generally, the knee valgus moment was found to be 2.4 times greater during the 110° COD and the vertical ground reaction force was recorded at 24.76 (N/kg) and 21.91 (N/kg) when comparing 110° and 45° COD, respectively. This finding is corroborated by Havens and Sigward (2015a), who reported greater knee abduction moments when comparing 90° to 45° COD. Corroborating the results of previous studies (Havens & Sigward, 2015a; Sigward et al., 2015), Schreurs et al. (2017) found that greater knee abduction moments were found in athletes demonstrating sharper CODs (90°, 135° and 180°) compared to 45° COD. Overall, these results suggest that the COD angle influences COD biomechanics, including the magnitude of knee joint loading during COD manoeuvres at greater angles than 45° when the mechanical demands placed on the knee increase thus also increasing the risk of ACL. However, a limited number of investigations have inspected 90° and 135° COD biomechanics from the perspective of risk of injury. Thus, further research is warranted to investigate 90° and 135° COD biomechanics and gain an understanding of the associated ACL injury risk factors.

Overall, current COD tests reflect the discrete aspects involved in athletes' movements, yet most COD assessments involve planned or unplanned, basic and reactive decision responses, which are very different to the complex decision-making processes of one, or even several, actions performed during real-life soccer matches. For example, two closely spaced movements (stimuli) may be needed if an attacking player carries out a fake ball pass (Henry, Dawson, Lay, & Young, 2012; Schmidt & Lee, 2005). This leads to the double-stimulation paradigm, whereby a player's reaction to the first out of two closely timed stimuli is normal. In comparison, their reaction to the second is more delayed than if it had happened in isolation (Schmidt & Wrisberg, 2008); therefore, current COD assessment methods do not consider these important aspects. Implementing a suitable control with acceptable test/re-test reliability regarding the aforementioned factors that also takes place in a chaotic environment, is challenging and perhaps even unrealistic. It is therefore necessary to assess players from a mitigating or rehabilitating framework perspective in order to reduce the risk of suffering a primary or secondary ACL injury.

In the rehabilitative context, clinicians would routinely use planned COD. However, the mitigating injury framework adopts the most ecologically valid test, which would mean that we needed to understand the mechanism of different COD tests before moving onto more complex scenarios. This is especially true in rehabilitation whereby one would want to be sure that the player could consistently achieve the desired outcome before moving forward in the rehabilitative cycle. Therefore, understanding such a manoeuvre is essential to mitigate injury risk and to develop successful ACL injury prevention/rehabilitation programs. This allows a practitioner to start with planned scenarios and to progress to unplanned activities, which transition from more closed (planned) to more open (unplanned) skills practice. Such progression will allows participants to move from performing in a predictable environment at a time of their choice (closed skill), to performing in an unpredictable environment that includes various external factors (open skill) (Magill, 2001). In addition, in order to rehabilitate a sportsperson to return to sport and enable reactiveness to unplanned situations, it is important to have first completed the planned stage. This would involve first understanding how individuals execute the planned COD manoeuvres and whether there is a difference when they consciously perform the tasks. Whilst it is understood that this may not explain what will happened in an unplanned environment, we aim to ensure that there are no significant differences in the individual during rehab when they are performing COD movements. This would have transferability to unplanned movements where we would perceive less risk if impairments were already corrected during the planned environment. It is unknown whether changing the impairments in a planned environment have any subsequent impact on an unplanned environment but this is an interesting area for future work.

Looking at planned versus unplanned movement is a perceptual control area of work and difficult for unplanned lab studies to undertake. As such, both planned and unplanned COD tasks are recommended during biomechanical testing when assessing the effectiveness of prophylactic training protocols; however this adopts the framework that starts with planned and moves to unplanned movements. Furthermore, it should be noted that there are currently no injury surveillance data to support or refute that athletes are likely to sustain an ACL injury due to excessive knee loading by an unplanned COD manoeuvre. Whilst this information is not available, it would seem plausible that injuries would occur in an unanticipated scenario, such as match situations. However, in considering prevention and rehabilitation to reduce the risk of ACL injuries, planned movements allow for the completion of task goals before moving on to unplanned movements. Therefore, the planned COD manoeuvre has become our measure task.

1.2 Statement of the problem

ACL injuries are often serious, costly and debilitating. The majority of ACL injuries occur during non-contact COD manoeuvres (Waldén et al., 2015). With non-contact injuries, it is possible to prevent these injuries through prevention programmes, either in the pre-injured state or as part of the rehabilitation from an injury. Subsequently, a great deal of ACL injury research has been dedicated to injury prevention strategies. However, before successful preventative efforts can be achieved, potential injury prevention solutions and appropriate prevention measures must be researched within models or frameworks. The most widely cited injury prevention model is the Sports Injury Prevention model by Van Mechelen, Hlobil, & Kemper (1987, 1992). Van Mechelen et al. (1992) developed the ‘sequence of prevention’ in 1992 (see Figure 1-1). This injury prevention model encompasses incidence and epidemiological evidence with the modifiable biomechanical factors associated with ACL injury. This model provides a rationale for the design of injury prevention programs with the goal of changing policy on how to prevent and manage these injuries on a national level.

This model consists of four stages, and its intended use was to inform the development of applied injury prevention strategies. The model must be considered for the development of an injury prevention strategy; however, the first two stages are of greatest importance for the advancement of knowledge of ACL injury. The first stage of the respective model directed the researcher to ‘establish the extent of the problem’; a typical approach to fulfilling the requirements of this stage is to identify high-risk groups which should be targeted through the appraisal of epidemiological research. On agreement that a specific pathology within a certain population merits the need to develop a prevention initiative, the second stage of Van Mechelen et al.’s (1992) model was to ‘establish [the] aetiology and mechanisms of injury’. Knowledge of predisposing factors for injury is essential to develop an understanding of how the pathologies develop; thus, the appraisal of etiological research findings is invaluable for the development of injury prevention strategies. Understanding the causes and injury risks are prerequisites for prevention strategies. In addition, understanding the underlying factors that increase the risk of ACL injury can provide insight to the particular vulnerability of the soccer population to ACL injury. By identifying these causal relationships, ACL injury prevention training programs can be created to target the biomechanically relevant factors associated with ACL injury (stage 3).

In order to develop an ACL injury prevention program, it is essential that the relationship between the biomechanical factors and the ACL injury is understood. Due to the potentially dangerous impact from ACL injuries amongst soccer players, individual athletes as well as sports clubs are interested in injury prevention strategies. However, while great efforts have been made to understand the aetiology of ACL injuries, several risk factors and various underlying factors that can increase the risk of an ACL injury are yet to be fully explored (Shultz et al., 2015).

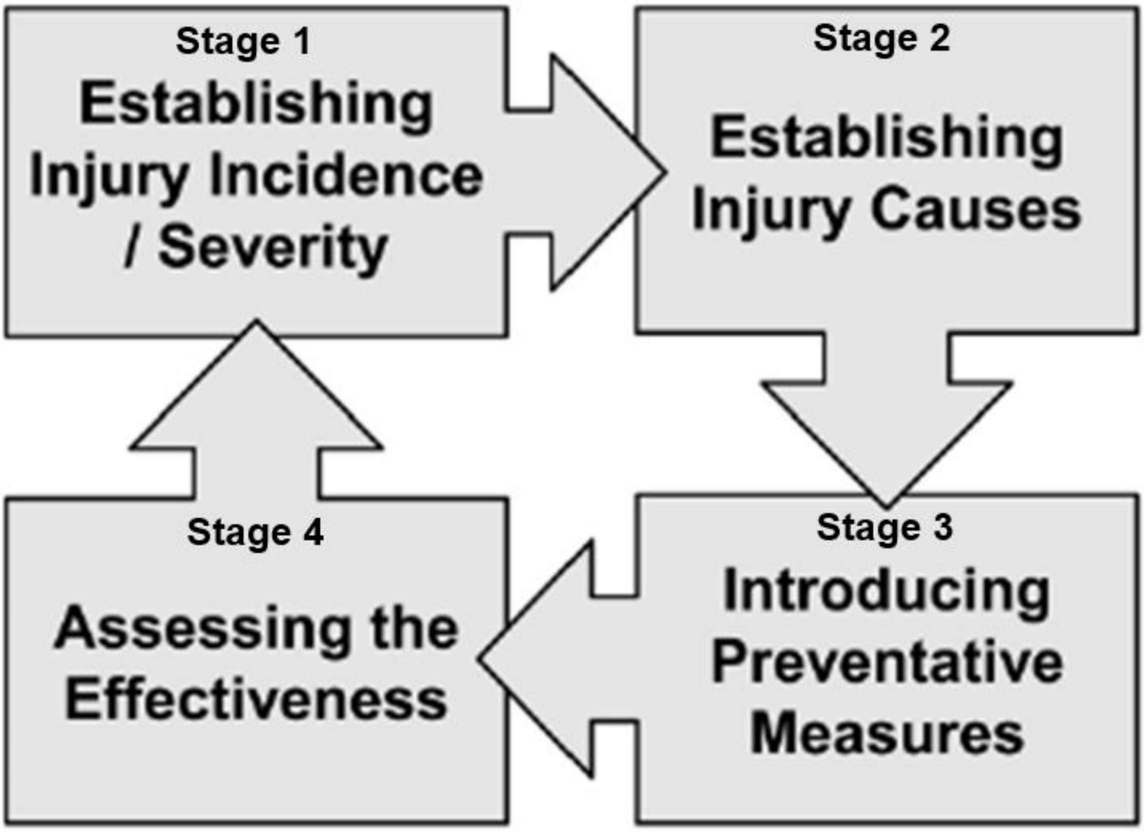


Figure 1-1 A four-stage process of the sequence of injury prevention (Van Mechelen et al., 1992)

While assessing ACL injury risk during 90° and 135° COD manoeuvres, a variety of biomechanical characteristics are engaged in order to perform the task without injury. Furthermore, there may be shortcomings in these biomechanical characteristics, such as asymmetrical players who face greater risk of injury than symmetrical players. The influence of limb asymmetry on ACL injury risk is an important factor, as, during match play, the athlete performs COD manoeuvres in both directions and at different angles; thus, they need to use both the preferred and non-preferred limb to push-off. Despite the various mechanisms involved in non-contact ACL injury (Quatman, Quatman-Yates, & Hewett, 2010), the dynamic control between limb differences, namely side-to-side differences and asymmetries, represent potential ACL injury risk factors (Brophy, Silvers, Gonzales, & Mandelbaum, 2010; Hewett et al., 2005; Pappas, Shiyko, Ford, Myer, & Hewett, 2016). As such, there has been increasing interest in the impact that limb differences have on COD biomechanics. For example, Hewett et al. (2005) carried out a prospective study of ACL injury, and discovered that there were significantly greater limb-to-limb differences in knee abduction moments amongst athletes who went on to suffer an ACL injury compared with athletes who remained un-injured. In addition, non-contact ACL injuries may not always happen on the dominant or preferred limb, as ACL injury rates for the non-dominant limb range are between 43% and 67% according to previous research (Brophy et al., 2010; Goerger et al., 2014; Matava, Freehill, Grutzner, & Shannon, 2002; Negrete, Schick, & Cooper, 2007).

Research studies that focus on between-limb differences during COD offer further understanding of the possible mechanisms involved in non-contact ACL injury. Moreover, to prevent injury, athletes should be encouraged to develop their ability to use both limbs equally to change direction safely (Dos'Santos, Thomas, Jones, & Comfort, 2018). However, many athletes have been found to have strength deficits between limbs, as well as deficient neuromechanical and dynamic control (Bishop, Turner, & Read, 2018; Brown, 2018). It is posited that such deficits may increase asymmetries in COD biomechanics, as one limb may have higher risk mechanics, and therefore an increased risk of injury. However, a very limited number of studies have analysed the impact of limb asymmetry on COD biomechanics, and only at a 45° angle (Greska, Cortes, Ringleb, Onate, & Van Lunen, 2016; Pollard et al., 2018). The link between limb preference and knee mechanics, and how these variables affect risk factors for ACL injury requires further research, as the scientific data on whether limb preference influences mechanical knee joint loading is inconclusive. Therefore, there is a need to identify the impact associated with increased ACL injury risk when one limb displays a greater biomechanical deficit than the other, and to improve our understanding of limb strength and biomechanical asymmetries during COD manoeuvres. However, no studies to date have examined the limb strength and biomechanical asymmetries during COD manoeuvres at 90° and 135° angles. Thus, one of the aims of this thesis was to determine whether strength and mechanical asymmetries exist during COD manoeuvres at 90° and 135° angles.

Hip mechanics are involved in frontal plane knee loading while performing COD manoeuvres. Researchers have previously found a link between peak knee abduction moments and angles and initial hip internal rotation, hip abduction and hip flexion while performing 45° COD manoeuvres (McLean, Huang, et al., 2005; Sigward & Powers, 2007). A greater peak valgus moment was associated with larger initial hip internal rotation and abduction during COD manoeuvres at 45° and 90° (Havens & Sigward, 2015a; Sigward & Powers, 2007). It is important to understand the lower extremity mechanics involved in COD manoeuvres to prevent injury as a result of the mixture of deceleration and change in direction required for COD; this has been shown to potentially lead to injury, especially ACL injury. Therefore, it is important that hip biomechanics are further researched to assess the potential risk factors for ACL injuries during COD manoeuvres so that ACL injury prevention programmes can also effectively correct abnormal hip biomechanics. Further investigations are needed to understand the role of the hip in the frontal, sagittal and transverse plane mechanics on the mechanism of ACL injuries during COD at 90° and 135° angles. Therefore, another aim for the thesis was to explore the relationship between ACL injury risk factors (knee valgus angle and moment) and hip kinematics and muscle strength during 90° and 135° COD manoeuvres.

As well as identifying the mechanics involved in sustaining an ACL injury, it is also essential to examine the causal factors. For example, if an athlete does not have enough strength to perform a particular manoeuvre, they may use a strategy that involves poor hip mechanics when conducting COD manoeuvres. Poor strength is thought to be an important factor in injury risk with studies that link hip weakness and ACL injury; these provide some support for a potential relationship between hip muscle strength and ACL injury. In a recent prospective study Khayambashi, Ghoddosi, Straub, and Powers (2016) showed that isometric hip abduction and external rotation strength independently predict noncontact ACL injury, as increased hip strength has a protective function against future ACL injury (Khayambashi et al., 2016). This study performed a prospective case-control study involving 501 athletes from various sports. Baseline hip external rotation and abduction isometric strength were measured before the start of the respective competitive seasons. Fifteen athletes sustained noncontact ACL injuries. This subgroup had significantly lower mean baseline isometric hip strength measures compared with non-injured athletes for external rotation (17.2 ± 2.9% body weight [BW] and 22.1 ± 5.8% BW, respectively) and abduction (30.8 ± 8.4% BW and 37.8 ± 7.6% BW, respectively). Using a logistical regression model, clinical cut-offs believed to define high risk for ACL injury were established for isometric external rotation strength (<20.3% BW) and isometric adduction strength (<35.4% BW).

In addition, most of the studies found a negative relationship between the knee valgus moments and angles and isometric hip muscle strength during landing and squatting (Hollman, Hohl, Kraft, Strauss, & Traver, 2013; McCurdy, Walker, Armstrong, & Langford, 2014; Ramskov, Barton, Nielsen, & Rasmussen, 2015; Stickler, Finley, & Gulgin, 2015; Suzuki, Omori, Uematsu, Nishino, & Endo, 2015), in that decreased isometric hip muscle strength is a risk factor for ACL injury. A moderate relationship was noted between the knee valgus and isometric hip abductor strength during a landing task, suggesting that the isometric hip abductor strength may be important for controlling hip adduction during double-limb tasks (Jacobs, Uhl, Seeley, Sterling, & Goodrich, 2005). More recently, Stickler et al. (2015) examined the relationship between the frontal plane kinematics of a single-leg squat and isometric hip strength. These authors found that isometric hip abductor strength was a strong predictor of frontal plane projection angle, whilst those with decreased hip abductor strength had an increased frontal plane projection angle. Moreover, Lawrence, Kernozek, Miller, Torry, and Reuteman (2008) found that during single-leg drop landings women with strong isometric external hip rotation strength saw a decrease in the knee abduction angle and vertical ground reaction force compared with a weaker group. In contrast, other studies failed to find similar relationships. For example, (Nilstad, Krosshaug, Mok, Bahr, & Andersen, 2015) reported that no relationship between hip strength and frontal plane knee valgus angles during a drop-landing manoeuvre in 279 Norwegian elite female soccer players. In addition, Homan, Norcross, Goerger, Prentice, and Blackburn (2013) looked at the impact of hip abductor and external rotator peak strength on knee abduction; hip adduction, and hip internal rotation angles while landing, and participants with lower peak strength had similar frontal and transverse plane hip and knee kinematics, whereas those with greater peak strength, demonstrated that peak strength on its own cannot be used to predict hip and knee kinematics while performing dynamic manoeuvres. Moreover, the effect of implementing a hip strengthening protocol for knee kinematics was analysed by Stearns and Powers (2014), and they found that four weeks post training, there was an increase in the peak strength of the hip abductors and extensors, yet there was only a 1.2° reduction in peak knee abduction angle for a landing manoeuvres, although it was not considered statistically significant ($p = 0.07$). While it is intuitive to consider hip muscle weakness a risk factor for ACL injury, experimentally the relationship between hip strength and injury mechanics continues to be inconsistent. These inconsistencies may be because of the variety of manoeuvres used for each study. However, all the aforementioned studies have examined the relationship between hip muscle strength and Knee abduction angle and moment only during landing (Gehring et al., 2009; Jacobs et al., 2007; Lawrence et al., 2008; Suzuki, Omori, Uematsu, Nishino, & Endo, 2015), squat (Willson et al., 2006) and single leg step-down (Hollman et al., 2009). Although 60-70% of non-contact ACL injuries occurred while a player performed a COD manoeuvre (Johnston et al., 2018; Montgomery et al., 2018), there has been no published research correlating the hip strength and frontal plane knee biomechanics during COD manoeuvres. Therefore, this is the first study to investigate the isometric hip muscle strength and knee biomechanics correlation during COD manoeuvre to 90° and 135°.

In addition, in the clinic and research field, muscle strength measurement is important in terms of risk injury or rehabilitation. The isokinetic dynamometer has been considered a gold standard for strength measurement, and the test can be performed in isometric, concentric and eccentric. Measuring the isometric, eccentric and concentric strength of a hip muscle is important. However, to measure all of these in one study would increase the susceptibility to fatigue. From previous work at the University of Salford (Ziyad, 2019), this study examined the correlation between the isometric, concentric and eccentric of hip muscles. This study shows large correlations between isometric and concentric peak torque (r = 0.54) and between isometric and eccentric peak torque (r = 0.52). This showed that there is strong concurrent validity between eccentric, concentric and isometric hip muscle strength. The most reliable of these three tests is isometric and is less likely to have a high measurement error. Furthermore, because there is a learning effect in undertaking an eccentric and concentric test, an isometric test would be perceived as more reliable. In addition, from a practical clinical point of view, a hand-held dynamometer (HHD) is an alternative device to assess muscle strength. The hand-held dynamometer is used to measure only the isometric lower limb muscle strength, and not the eccentric or concentric muscle strength. A hand-held dynamometer has the advantage of being a portable device, in a suitable size, available at a low cost and offering a convenient way to assess isometric muscle strength in a clinical setting. Moreover, HHD has strong validity in comparison to the Biodex system, and the results reveal a high correlation between HHD and the Biodex System 4 PRO dynamometer (Kim et al., 2014; Mentiplay et al., 2015). Isometric strength can be measured with a HHD whereas eccentric and concentric strength would be difficult to determine. Therefore, measuring isometric hip muscle strength has become our muscle strength measure in this thesis.

Ultimately, the results of these studies show that isometric hip strength may alter the lower extremity biomechanics associated with ACL injury risk. It is therefore essential that hip muscle strength is investigated as a potential risk factor for ACL injuries during sporting tasks, such as COD manoeuvres, so that an ACL injury prevention program can also effectively correct abnormal hip biomechanics. However, all these studies have only investigated the relationship between hip muscle strength and ACL injury risk factors during landing, single leg squats or double leg landings. Moreover, almost 60-70% of non-contact ACL injuries occurred while a player performed a COD manoeuvre (Johnston et al., 2018; Montgomery et al., 2018). Furthermore, COD has a greater risk of ACL injury compared with landing from a drop jump (Chinnasee et al., 2018). The knee valgus moment was 6 times higher in COD compared to drop jump landings (1.58 ± 0.60 Nm/kg vs 0.25 ± 0.16 Nm/kg) (Kristianslund & Krosshaug, 2013). Therefore, a COD manoeuvre would produce larger magnitudes of knee abduction motion, which would be more appropriate to look for, especially when changing direction at sharper angles, such as 90° and 135°. However, to the author's knowledge, there has been no published research correlating the hip abductor, extensor, and external rotator strength on frontal plane knee biomechanics during 90° and 135° COD manoeuvres. Furthermore, hip abductors, extensors and external rotator muscle strength are important points of focus and may decrease the risk of an ACL injury. Therefore, a further aim for the thesis was to explore the relationship between ACL injury risk factors (knee valgus angle and moment) and hip muscles strength during 90° and 135° COD manoeuvres.

1.3 Thesis aims:

The aims of the thesis are therefore to:

1. Review the literature related to Anterior Cruciate Ligament injuries, including their occurrence, mechanism and proposed risk factors (chapter 2).
2. Review the literature regarding screening tools to identify potential Anterior Cruciate Ligament injury risk (chapter 2).
3. Establish the reliability of 3D during change of direction manoeuvres (chapter 3).
4. Establish the reliability of the Biodex system to measure hip muscle strengths (chapter 3).
5. Establish whether asymmetry in knee biomechanics kinematics and kinetics and hip kinematics as well as hip muscle strengths between preferred and non-preferred limbs during COD manoeuvres at 90° and 135° angles exists. (chapter 4).
6. Establish whether differences in knee biomechanics kinematics and kinetics and hip kinematics between 90° and 135° change of direction manoeuvres exists (chapter 5).
7. Establish the relationship between the hip kinematic and muscle strengths and the non-contact ACL injury risk factors during 90° and 135° COD manoeuvres (chapter 6).

1.4 Thesis Structure:

Chapter 1
- Introduction.
- Thesis aims.

Chapter 2
- Literature review.
- Research questions and hypothesis.

Chapter 3
- Method.

Chapter 4
- Biomechanical and strength asymmetry in preferred and non-preferred legs during 90 and 135 change of direction manoeuvres and MVIC test.

Chapter 5
- Biomechanical differences between 90° and 135° change of direction manoeuvres.

Chapter 6
- The relationship between the hip kinematic and muscle strengths and the non-contact ACL injury risk factors during 90° and 135° COD manoeuvres.

Chapter 7
- Overall Summary, Conclusion and Future recommendations.

2 Chapter 2: Literature Review

This literature review provides the background and rationale for the work conducted in this thesis. The following are therefore discussed:

- Knee Joint Anatomy and Biomechanics (2.1).
- Knee injuries in sport (2.2)
- Anterior cruciate ligament (ACL) injury (2.3).
- Incident of ACL injury (2.3.1).
- Anterior cruciate ligament reconstruction (ACLR) (2.3.2).
- Economic costs of ACL injury (2.3.3).
- Returning to sport after ACL injury (2.3.4).
- The role of ACL injury in future degenerative conditions (2.3.5).
- Mechanism (2.3.6) risk factors (2.3.7) for ACL injury.
- Injury prevention (2.4)
- Change of direction (COD) manoeuvre (2.5).
- Hip motion and loading (2.6).

2.1 Knee joint anatomy and biomechanics

The knee is the largest joint in the human body and is surrounded by an osseous anatomy made up of the femur, patella, tibia, and fibula (Figure 2-1). Two separate fibrocartilage pads, the medial and lateral meniscus, are found intra-articularly between the femoral and tibia, as well as articular cartilage. The knee’s normal range of motion is ensured by two cruciate ligaments and two collateral ligaments (Figure 2-1). These are the anterior cruciate ligament (ACL); the posterior cruciate ligament (PCL) primarily restrict anterior and posterior translation of the tibia on the femur respectively. The medial collateral ligament (MCL) restrains valgus forces and the lateral collateral ligament (LCL) varus forces applied to the knee. The ACL and PCL work together in harmony to prevent sagittal plane translation of the tibia in relation to the femur (Butler, Noyes, & Grood, 1980; Gollehon, Torzilli, & Warren, 1987; Markolf, Mensch, & Amstutz, 1976).

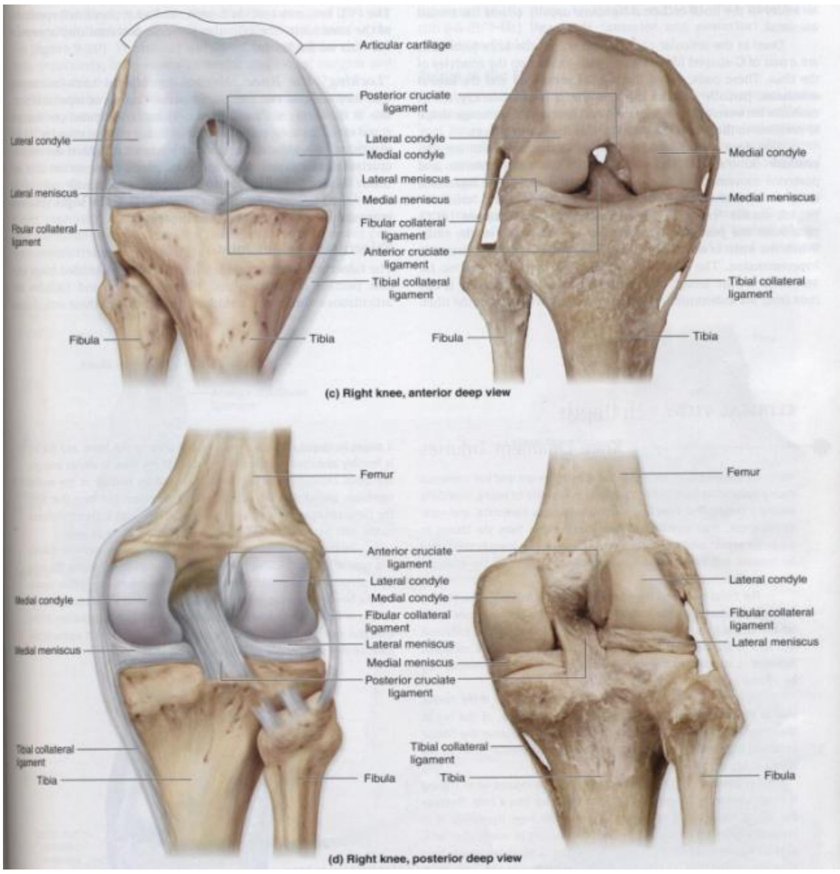


Figure 2-1 Knee joint anatomy (McKinley, 2008)

A full description of the biomechanics of the knee is important in attempting to understand the mechanical aspects of the joint in relation to knee disorders. As the main focus of this thesis in ACL injury, the biomechanics of tibiofemoral articulation are examined. It is commonly understood that tibiofemoral articulation happens in all three anatomical planes-sagittal, frontal and transverse, and that there are six degrees of freedom (6-DOF), which include three rotations and three translations that occur in the middle of the tibial plateau and the femoral condyles (Woo, Abramowitch, Kilger, & Liang, 2006) (Figure 2-2). Apart from flexion and extension, the range of motion of the tibiofemoral in all other directions is quite limited. The amount of laxity in the range of motion of the joint varies greatly among the general population and has been shown to differ according to age, stage of puberty, sex, and race (Quatman, Ford, Myer, Paterno, & Hewett, 2008). Moreover, tibiofemoral joint motion that goes beyond the normal physiologic joint laxity may result in tissue damage affecting the knee’s internal structures.

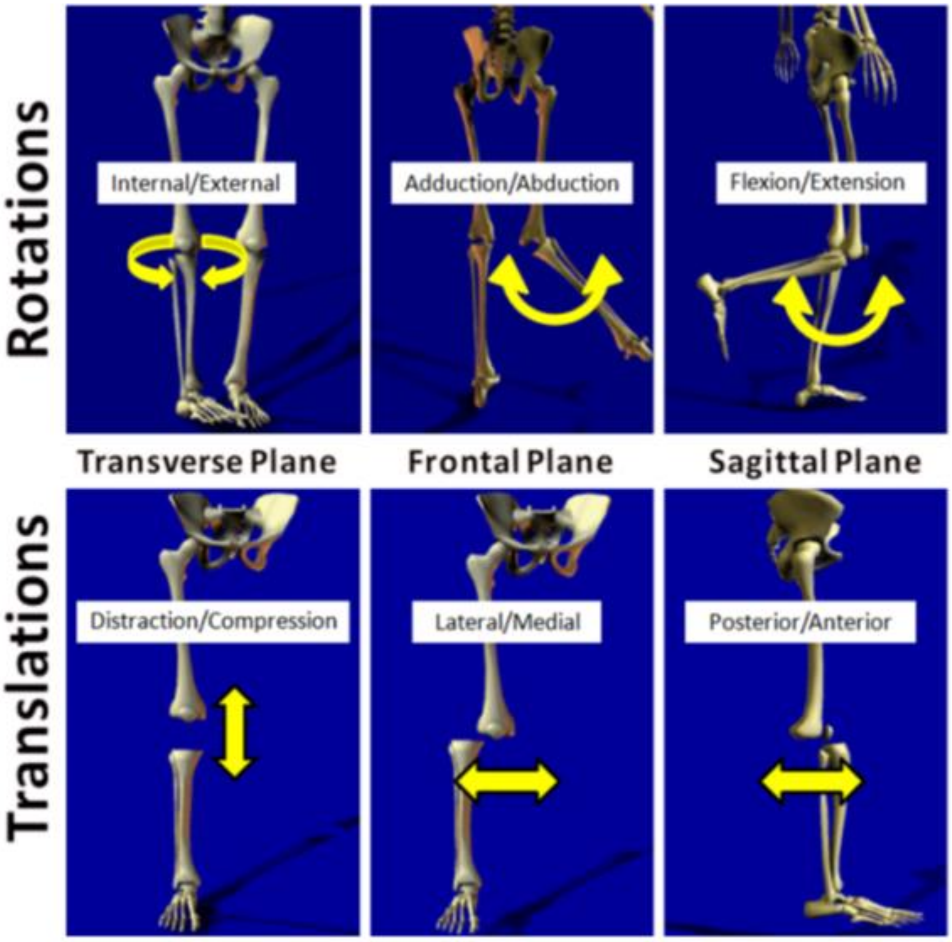


Figure 2-2 Tibiofemoral joint motions (Quatman 2009)

2.2 Knee injuries in sport

A potential risk of injury comes with physical activity, and an increased participation in sport in particular leads to the possibility of injuries being sustained, resulting in costs to the affected individual due to temporary or long-term disability and the subsequent loss of earnings; an impact on the healthcare system, and therefore the economy as well. Approximately 50 to 75% of injuries affect the lower limb in a range of sports and at various levels of expertise (Agel et al., 2007). The most commonly injured joint in the lower limbs is the knee, and it often causes a major loss of training time and partaking in competition (Agel et al., 2007; Dallalana, Brooks, Kemp, & Williams, 2007). Around 15-25% of injuries among football, basketball, volleyball and rugby players, whether high school, college or professional, are to the knee (Agel et al., 2007; Dallalana et al., 2007). Furthermore, knee injuries are often serious, and may lead to the affected individual being unable to return to sport, needing to change their job, and even developing osteoarthritis (Utting, Davies, & Newman, 2005).

Injury to the knee can involve the ligaments, tendons or fluid-filled sacs (bursae) surrounding the knee joint, and the meniscus. ACL injuries make up at least 50% of all knee injuries (Arna Risberg, Lewek, & Snyder-Mackler, 2004), which makes it essential to focus on this ligament when discussing knee injuries. Moreover, surgical repair along with extensive long-term rehabilitation is often necessary for anterior cruciate ligament injuries (Arna Risberg et al., 2004). Therefore, injuring the anterior cruciate ligament results in an extensive amount of time-loss in sport , and those affected may be unable to return to previous activity levels, as well as being at further risk of early onset knee osteoarthritis (OA) (Lohmander, Englund, Dahl, & Roos, 2007; Utting et al., 2005). Ardern, Webster, Taylor, and Feller (2011) conducted a cohort study and found that only 33% of 503 patients returned to the same level of competition 12 months post anterior cruciate ligament reconstruction (ACLR).

2.3 Anterior cruciate ligament (ACL) injury

ACL injury is catastrophic and have a dramatic effect on a patient's return to sports participation, activity level and long-term quality of life. Ardern, Taylor, Feller, and Webster (2012b) surveyed 314 ACL reconstruction (ACLR) individuals 2-7 years after reconstruction. The investigators found that only 41% of their participants had attempted competitive sport at follow-up, and only 29% were actively participating at their pre-injury competitive level (Ardern et al., 2012b). A systematic review by Ardern, Taylor, Feller, and Webster (2014) revealed that following ACL reconstruction, just 65% of non-elite athletes returned to a pre-injury level of sport, and only 55% returned to competitive level sport. Previous research conducted by Ardern et al. (2011) also found that just 33% of the participants returned to pre-injury level and competition 12 months after having ACL surgery. In addition, Shah, Andrews, Fleisig, McMichael, and Lemak (2010) states that a study conducted with American Football players showed that 37% of players who had ACL surgery did not return to playing the sport. Myklebust, Holm, Maehlum, Engebretsen, and Bahr (2003b) found that following ACL reconstruction, 58% of Norwegian elite handball players returned to competition at the same level, with the 42% either competing at a lower level or not returning to competitive sport. Research by Lohmander, Ostenberg, Englund, and Roos (2004) found that more than half of female Swedish football players were unable to return to sport after an ACL injury, with just 15% returning to pre-injury levels of activity.

Furthermore, in the long term, an ACL injury could increase the risk of early onset osteoarthritis of the knee (Zabala, Favre, & Andriacchi, 2015), with several researchers claiming that the majority of individuals suffering from an ACL injury experience early onset of osteoarthritis (OA), including the pain associated with that, as well as limited functioning (Ahlden et al., 2012; Lohmander et al., 2007; Oiestad, Engebretsen, Storheim, & Risberg, 2009). In addition, Lohmander et al. (2004) discovered radiographic patellofemoral or tibiofemoral OA in 51% of ACL injured female soccer athletes 12 years post injury. Ahlden et al. (2012) conducted the largest known study reporting results in almost 18,000 patients with a history of ACL reconstruction through the Swedish National ACL Register. The study collected results KOOS scores from registry respondents at 1, 2, and 5 years postoperatively. Ahlden et al. (2012) found that patients who underwent a second surgery had significantly poorer knee related quality of life compared to those who had had their first reconstruction. Participants with an additional ACL reconstruction also displayed no significant improvement in symptoms, pain, and activities of daily living at 5 years post-surgery compared with their preoperative values (Ahlden et al., 2012). According to Oiestad et al. (2009), a patient who suffers an ACL injury without injury to the meniscus has a 0-13% chance of developing knee OA more than 10 years after the injury; but if the meniscus is involved the likelihood increases to between 21% and 48%. Several studies have found that up to 80% of ACL injured knees show radiographic evidence of osteoarthritis five to 15 years after the initial injury, in particular when there is concomitant meniscal damage (Neuman et al., 2008). Following ACLR, patients with severe radiographic osteoarthritis have a poorer quality of life in relation to health; therefore, the clinical impact is significant (Filbay, Ackerman, Russell, Macri, & Crossley, 2014). Barenius et al. (2014) conducted a trial and in a 14 years follow-up after ACL reconstruction, they found an incidence rate of 57% for osteoarthritis, and that this figure is far higher than the rate of 18% for osteoarthritis of the contralateral knee; in particular, they found that osteoarthritis is most commonly affects the medial compartment. Leiter, Gourlay, McRae, de Korompay, and MacDonald (2014) conducted a retrospective case study and also found that knees that had undergone ACLR showed a significantly higher incidence and greater severity of osteoarthritis compared to non-ACL-injured counterparts.

2.3.1 Incidence of ACL injury in sports

In England, between 1997–1998 and 2016–2017, there were 133 270 cases of ACL reconstruction (124 489 patients), the rate of ACL reconstruction increased 12 times from 2 per 100,000 person-years in 1997–1998 to 24.2 per 100,000 person-years in 2016–2017 (Abram, Price, Judge, & Beard, 2019). The incidence of ACLR in the Australia increased 43% from 54 per 100,000 person-years in 2000 to 77.4 per 100,000 person-years in 2015

(Zbrojkiewicz, Vertullo, & Grayson, 2018). Furthermore, the incidence of ACLR in the United States rose from 86,687 (32.9 per 100,000 person-years) in 1994 to 129,836 (43.5 per 100,000 person-years) in 2006 (Mall et al., 2014). The incidence of ACL injury in Finland of 60.9 per 100,000 person-year in a cohort study of 46,000 youths (Parkkari, Pasanen, Mattila, Kannus, & Rimpela, 2008). In Swedish, the incidence rate per 100,000 person-years was 78 for ACL injury (Nordenvall et al., 2012). The incidence of ACLR in New Zealand of 36.9 per 100,000 person-years (Gianotti, Marshall, Hume, & Bunt, 2009). Several research studies have examined the incidence of ACL injury in different types of sports; for example, Hootman, Dick, and Agel (2007) analysed data from the National Association (NCAA) of injury surveillance during a 16-year period. They discovered that American football players had the highest number of reported ACL injuries (2538 injuries from 4800 in total, 53% of all recorded ACL injuries). The second highest number to sustain ACL injuries were basketball players (677 injuries of 4800 in total, 10% of all recorded ACL injuries). Contrastingly, a meta-analysis revealed that the incidence of ACL ruptures was highest amongst basketballers, followed by soccer players and then lacrosse players (Prodromos, Han, Rogowski, Joyce, & Shi, 2007).

2.3.2 Anterior cruciate ligament reconstruction (ACLR)

ACLR is considered to be essential for those playing sports requiring rapid jumping and COD, and the literature shows that a number of different procedures have been investigated, and while some studies recommend surgical options (Fink, Hoser, Hackl, Navarro, & Benedetto, 2001), other suggest a non-surgical approach (Giove, Miller, Kent, Sanford, & Garrick, 1983). The choice to undergo surgery often depends on the recommendation of the surgeon, and the patients' age, activity levels, severity of the injury, and cost of surgery and rehabilitation, all need to be considered (Gokeler et al., 2016; Macaulay, Perfetti, & Levine, 2012). While it is generally believed that ACLR is necessary for return to sport, there are some shortcomings in the evidence to support this (Frobell et al., 2013, 2015), and rehabilitation without surgery has also shown good results in the management of ACL injuries (Weiler, Monte-Colombo, Mitchell, & Haddad, 2015); however, surgery may be necessary at a future time (Myklebust & Bahr, 2005).

ACLR surgery typically involves an allograft or an autograft. An allograft is where a tissue graft is surgically transplanted from one person to another, whereas autograft uses a tissue graft that is surgically transplanted from one area of the patient's own body to another (Macaulay et al., 2012; Prodromos et al., 2007). Allografts have the advantage of no harvest site morbidity, as well as less postoperative pain (Macaulay et al., 2012), but the risk of graft failure postoperatively is two to four times higher with allografts compared to autografts (Krych, Jackson, Hoskin, & Dahm, 2008; Prodromos et al., 2007). A meta-analysis showed a ratio of 5.03 times greater risk of the graft failing with allografts compared to autografts (Krych et al., 2008). Therefore, autografts are preferred, and they have other benefits such as better joint stability, less risk of infection, and typically an earlier return to sport (Bonasia & Amendola, 2012).

The most common autografts are from the ipsilateral semitendinosus and gracilis or the patellar tendon (Gobbi & Francisco, 2006), with patellar tendon grafts the more popular surgical choice; however, hamstring tendon grafts are becoming increasingly common (Cerulli et al., 2013). Moreover, there is no gold standard treatment for an ACL injury, and the graft selection usually depends on the opinion of the individual surgeon (Bonasia & Amendola, 2012; Macaulay et al., 2012). No matter which graft is chosen, there still may be deficits in eccentric and concentric knee extensor (Knezevic, Mirkov, Kadija, Nedeljkovic, & Jaric, 2014; Konishi, Aihara, Sakai, Ogawa, & Fukubayashi, 2007; Mirkov et al., 2017) and flexor strength (Kramer, Nusca, Fowler, & Webster-Bogaert, 1993; Tengman, Brax Olofsson, Stensdotter, Nilsson, & Hager, 2014), which can continue in the long term, even 25 years post-surgery (Tengman et al., 2014).

2.3.3 Economic costs of ACL injury

The standard treatment for an ACL rupture is ACL-Reconstruction (ACLR), as it has been designed to reduce intra-articular damage in the long term and restore stability and function (Van Grinsven, Van Cingel, Holla, & Van Loon, 2010). Surgical reconstruction is not conducted for all ACL deficient patients, and the decision is made according to their level of knee function (Beynnon, Johnson, Abate, Fleming, & Nichols, 2005). It is estimated that two-thirds of patients choose reconstruction, which is costly itself, and rehabilitation is estimated to be $17,000-25,000 per operation (Hewett, Ford, Hoogenboom, & Myer, 2010). Even so, the costs in the long-term show that ACL reconstruction is not that much more expensive than conservative treatment, as the latter can result in associated meniscal injuries and the early development of osteoarthritis if reconstructive surgery is not carried out (Farshad et al., 2011). ACL injuries cause a significant economic burden, especially because 175,000 and 250,000 ACL injuries are reported in the US annually (Dunn & Spindler, 2010; Herzog, Marshall, Lund, Pate, & Spang, 2017; Mather et al., 2013), making a total cost of 2.6 billion US Dollars, which does not include rehabilitation or the treatment of secondary knee joint injuries or disorders

(Herzog et al., 2017; Mather et al., 2013). In Australia, ACL injuries cost about A$75 million each year (Janssen, Orchard, Driscoll, & van Mechelen, 2012), and in New Zealand and Australia, the figure is 17.4 million New Zealand Dollar (Gianotti et al., 2009; Janssen et al., 2012). However, there are a lack of studies on the economic and social impact in other countries, for example, no studies have evaluated the cost from ACL injuries in the United Kingdom.

ACL injury does not only have health-related effects on the patient (pain, disability, limited QoL, and psychological impact), it can also have separate economic and societal effects, some of which have long-term implications. Even though the direct economic burden of treatment (e.g. cost of medication, cost of surgery and rehabilitation sessions) in this population is fairly well-documented, the long-term, indirect effects on society (societal effects), such as loss of productivity and claims due to disability, have not been adequately reported (Mather et al., 2013). Currently, the main two treatment options that exist in relation to ACL injury are surgical reconstruction and focused rehabilitation. Early reconstruction is widely used for young, active patients, whereas rehabilitation is typically preferred by older and lower-demand individuals (Mather et al., 2014). Additionally, two types of costs of ACL injury can be identified: direct costs (mainly costs of diagnosis and treatment) and indirect costs (lost wages, lost productivity, disability claims) (Mather et al., 2013). In accordance, even though societal costs are related to both types, direct costs tend to be reported more often as they are easier to measure (Mather et al., 2013). Meanwhile, time taken off work due to ACL injury management, which translates to lost wages, is usually estimated to be on average 28 workdays that are missed due to ACL reconstruction and 40 workdays missed due to knee arthroplasty surgery. Furthermore, disability claims can be variable depending on the welfare system, the sex and age of patients. In total, the average annual disability payment ranged from $9,000 to $17,000 per claimant per year in the USA based on the 2011 Current Population Survey (Mather et al., 2013).

Considering these two different treatment options, short-term direct costs of early reconstruction are estimated to be $20,000 per patient compared to $21,500 associated with rehabilitation with the option of delayed reconstruction based on 2012 figures in the USA (Mather et al., 2014). However, when indirect long-term societal costs are considered, the average lifetime cost of the management of ACL injury becomes $38,000 per patient undergoing early reconstruction compared to $88,500 per patient for rehabilitation with delayed reconstruction. This translates to an economic burden of $7.6

billion each year for reconstruction and $17.7 billion for rehabilitation in the USA. Other long-term complications also represent a considerable concern, due to the added costs and health implications for the patients.

Table 2.1 shows estimations of the cost of management of complications associated with ACL injury. Based on 2012 statistics in the USA, nearly 16% of all patients who underwent early reconstruction tended to develop symptomatic osteoarthritis and most of these patients would eventually need complete knee arthroplasty (Mather et al., 2013). These figures tended to increase three-fold in the presence of meniscal tears (Oiestad et al., 2009). With rehabilitation, nearly 20% of patients who received physiotherapy without reconstruction were expected to develop symptomatic osteoarthritis, and subsequently this resulted in the necessity of knee arthroplasty in most of these patients (Mather et al., 2013; Muraki et al., 2012).

Table 2-1 Estimated economic burden associated with complications following ACL injury based on 2012 USA based statistics (Mather et al., 2013). The estimations include direct and indirect costs.

Cost of management option (US Dollars)	**ACL injury with meniscal symptoms**	**ACL tear with radiographic osteoarthritis**	**ACL tear with symptomatic osteoarthritis**	**Total knee arthroplasty due to ACL injury**
ACL reconstruction	**$44,000-$44,100**	**$38,500-$49,000**	**$40,000-$47,700**	**$42,800-$46,000**
Rehabilitation	**$83,500-$86,000**	**$85,000-$89,000**	**$81,800-$91,700**	**$84,700-$88,700**

2.3.4 Returning to sport after an ACL Injury

The goal of most many patients who have suffered an ACL injury or ACLR is to resume their activities and return to sport (RTS) as soon as possible (Bauer, Feeley, Wawrzyniak, Pinkowsky, & Gallo, 2014). A satisfactory activity level without ACLR could be achieved, despite impairments and decreased activity level. The activity level is affected by the time since injury, and physical and psychological factors (Österberg, Kvist, & Dahlgren, 2013). The rate of return to pre-injury activity level is similar with or without surgical treatment (Frobell et al., 2015). Even return to elite professional football is possible, although uncommon, after an ACL injury treated non-surgically (Weiler et al., 2015).

RTS is an important clinical outcome after ACLR (Ardern et al., 2014). A meta-analysis involving 69 studies which included 7556 patients, 81% returned to some kind of sport;

after unilateral ACLR 65% of patients returned to preinjury levels of sport after a mean follow-up time of 40 months, yet only 55% returned to competitive sport 3 years following ACLR, despite having good physical function (Ardern et al., 2014). In an earlier review, Ardern et al. (2011) reported that only 44 % of patients returned to competitive sport. More recently, a 15-year prospective study in men's professional football by Walden, Hagglund, Magnusson, and Ekstrand (2016) showed that 85 % of players returned to training and 65 % of players returned to competitive play within three years following ACLR. The long term sport participation after an ACLR is not known, but after 5 years, one out of five was still active regardless of treatment (ACLR or not) (Frobell et al., 2015). Roos, Ornell, Gardsell, Lohmander, and Lindstrand (1995) reported that 30% of ACL-injured football players were active in football three years after injury compared with 80% in an un-injured control population. Young athletes (<25 years) are 1.5 times more likely, and elite athletes are >2 times more likely to RTS after ACLR (Ardern et al., 2014). Fewer females RTS than males (Tan, Lau, Khin, & Lingaraj, 2016), males are approximately 50% more likely to return to their previous level of sport or to competitive sport (Ardern et al., 2014). Females also RTS later than males after ACLR (Ardern et al., 2011). When RTS, females showed more concern over environmental conditions and the risk of re-injury compared to males 2-7 years after ACLR (Ardern et al., 2012b). In two cohorts with female and male elite ACLR football players, 86% vs. 100% had returned to football training within 12 months (Walden, Hagglund, Magnusson, & Ekstrand, 2011). In two football player cohorts of different ages and level of play, the return rate was only 46-67% for females compared to 60-76% for males (Brophy et al., 2012; Sandon, Werner, & Forssblad, 2015). Male, sex, and younger age are factors linked to a return to football after ACLR, with activity-related knee pain and cartilage injury factors associated with not returning to football (Brophy et al., 2012; Sandon et al., 2015). Higher quadriceps strength, less pain and less effusion are factors associating with RTS, although the evidence supporting these factors is weak (Czuppon, Racette, Klein, & Harris-Hayes, 2014).

Rehabilitation and return to sport outcome measures are widely discussed in the literature (Bradley, Klimkiewicz, Rytel, & Powell, 2002; Lyman et al., 2009) and a number of outcome measures are used in rehabilitation programs to determine the time course for when athletes can return to sport (Bauer et al., 2014; Webster, Feller, Whitehead, Myer, & Merory, 2017). For many years, isokinetic dynamometry has been considered the gold standard measure of the strength and dynamic stability of the knee joint and is frequently used for the evaluation of the success of the rehabilitation program (Cvjetkovic et al., 2015). This method provides an objective measure of both concentric and eccentric knee flexor and extensor strength (Aagaard & Andersen, 1998). General recommendations are that strength and ROM should be near pre-injury level or equal to the uninjured side, and that there should be no instability, tenderness, inflammation or effusion at the time of RTS (Creighton, Shrier, Shultz, Meeuwisse, & Matheson, 2010). However, the optimal time or set of criteria for when athletes are allowed to return to sport following ACLR remains uncertain (Ardern et al., 2014; Webster et al., 2017). Typically, athletes require 9 to 12 months recovery before making a return to competitive sport (Ardern et al., 2014; Ardern et al., 2011). Recent work by Webster et al. (2017) reported that just a third of athletes had returned to their pre-injury competitive sport 12 months after ACLR while another one-third had only returned to training.

The contralateral limb is typically used as the comparator, mostly through muscle strength as according to a Limb Symmetry Index (LSI). The LSI is applied in routine clinical practice to assess the performance of the injured limb compared with the non-injured limb; this is presented as a percentage score that accords with the performance, as access to normalised scores may be difficult. The validity of LSI is based on the following two assumptions: firstly, that symmetry represents the individual's pre-injury functional state, and secondly that the non-injured limb is representative of a healthy normality that has not been affected by the injury (Herrington, 2013). However, literature on this issue is split between those who recommend using the LSI (Logerstedt, Lynch, Axe, & Snyder-Mackler, 2013) and those who warn against its use (Welling et al., 2018) and thus suggest comparisons to healthy control values (Zwolski, Schmitt, Thomas, Hewett, & Paterno, 2016) or use absolute measures to provide a context to the symmetry values (Gokeler, Welling, Zaffagnini, Seil, & Padua, 2017). Logerstedt et al. (2013) supports the use of the LSI, as they claim that the non-injured limb represents the healthy state of the individual. However, a number of studies have revealed that impairments to the muscle strength of the contralateral limb often occur following an ACL injury (Chung et al., 2015). This would create similarities and those assumed to have acceptable symmetry could have reduced strength that is not properly acknowledged. The contralateral limb could then be at risk as it is only as strong as the ACL injured side.

Researchers who recommend using the LSI claim that a standard of 90% (comparison of injured versus contralateral) indicates recovery (Thomee et al., 2011). Barber, Noyes, Mangine, McCloskey, and Hartman (1990) were the first to suggest a cut-off for acceptable performance; this was based on 90% of healthy participants scoring an LSI of > 85%. Thomee et al. (2012) have shown the importance of standardising levels for LSI, with their data for success ranging from 80% to 100%, which highlights how an increase in the LSI cut off level has a major impact on the number of individuals classed as being recovered. However, questions still remain around what is an appropriate LSI for defining recovery or recommending rehabilitation interventions. The LSI clearly requires further research and testing on its use as an outcome for rehabilitation (Wellsandt, Failla, & Snyder-Mackler, 2017). The European Board of Sports Rehabilitation (EBSR) has recommended that the LSI should be presented along with absolute values, including at group level, as well as reporting the proportion of individuals who achieve each standard (Thomee et al., 2011). Logerstedt et al. (2013) claim that whilst symmetry is an important goal for post-operative rehabilitation and that this remains valid, it should also be considered alongside absolute values. In order to provide an accurate context for LSI measures, and to make further recommendations about its validity, a greater understanding of the strength of the non-injured limb amongst healthy subjects is required. Comparisons with healthy subjects/limbs are essential to measure success among the ACL injured population.

With regard to LSI, caution is necessary as it can hide bilateral deficits where the non-injured limb is affected by the injury and length of inactivity (Gokeler et al., 2017). Moreover, some researchers disagree on the value of symmetry measures and claim that unilateral normalised values can better predict outcome measures within the ACLR population (Kester, Behery, Minhas, & Hsu, 2017), which supports the use of normalised measures that follow ACLR (Pietrosimone et al., 2016). In addition, contralateral weakness could increase the symmetry values while providing a false representation of the strength and function of the affected limb; this, again, suggests the use of normative data. According to Zwolski et al. (2016), using just the LSI may not provide all the information necessary on the extent of the impairment, and performance strength, which can be better assessed by conducting a comparison of strength performance values with the normative values displayed by healthy controls. This is in line with the findings of Wellsandt et al. (2017) who claim that subjects with the required 90% symmetry criterion for strength at six months after the ACLR would not have passed this criterion if the performance of the contralateral limb was compared to its performance prior to ACLR surgery, and not after. Therefore, symmetry, assessed on strength through comparison to the contralateral limb, does not necessarily mean that adequate recovery has taken place, or confirms that the patient is ready to return to sport. As explained, an ACL injury may be described as a double limb problem, rather than a single limb injury, as muscle strength deficits have been found in the contralateral uninjured limb, as well as the injured limb (Trulsson, 2019).

A gradual progression through sport specific training is important and movement quality is as important as quantitative performance before RTS (Myer, Paterno, Ford, & Hewett, 2008a). Functional objective parameters are typically used to assess an athlete's readiness to return to sport (Bauer et al., 2014). The criteria for RTS that has been suggested combines an evaluation of functional performance, including muscle strength (power and endurance); knee stability; bilateral limb symmetry; postural control; agility; technique for sport-specific tasks, and PROMs (Ellman et al., 2015; Myer, Paterno, Ford, Quatman, & Hewett, 2006). Almost all rehabilitation protocols after ACLRs involve distinct goal-based phases where progression is determined by successful attainment of specific outcome measures. For example, the primary goals in phase one (1-2 weeks) are typically to reduce knee swelling and improve knee range of motion before the start of phase two where basic strength and range of motion exercises are started. Two recent studies have shown that the ACL reinjury risk can be significantly reduced if the return to sport criteria has been met (Grindem, Snyder-Mackler, Moksnes, Engebretsen, & Risberg, 2016; Kyritsis, Bahr, Landreau, Miladi, & Witvrouw, 2016). Kyritsis et al. (2016) demonstrated that athletes who did not achieve the discharge criteria before returning to professional sport were four times more likely to sustain an ACL injury than those who met the return to sport criteria. In addition, it appears that a low H:Q ratio is linked to an increased risk of ACL injury (Kyritsis et al., 2016). Functional performance tests are important pieces of the RTS puzzle, but should not be used independently, a test battery is likely more appropriate (Thomee et al., 2011). The RTS decision should be taken collaboratively between the coach, the physiotherapist, the surgeon and the patient.

Previous research has tended to focus on functional performance tests in the evaluation of a patient with an ACLR and RTS. However, recently, there has been increased attention on psychological factors including psychological readiness to RTS, low fear of sustaining a new injury, and trust in the knee (Langford, Webster, & Feller, 2009; Tjong, Murnaghan, Nyhof-Young, & Ogilvie-Harris, 2014). The traumatic nature of an ACL injury is often followed by psychological effects such as negative emotions, depression and reduced self-confidence (Kvist, Ek, Sporrstedt, & Good, 2005). Moreover, negative emotions surrounding re-injury are a large obstacle for any athlete's mind to block out during rehabilitation and recovery from ACLR (Kvist et al., 2005; Webster et al., 2017). Assessing the athlete's psychological profile is helpful when identifying those with a high chance of returning to preinjury activity levels (Gobbi & Francisco, 2006). Psychological factors likely have a strong influence on RTS. Low fear of re-injury, self-motivation, confidence, psychological readiness to RTS, positive mood and emotions facilitate RTS (Ardern, Taylor, Feller, & Webster, 2012a; Ardern, Taylor, Feller, Whitehead, & Webster, 2013; Czuppon et al., 2014). Fear of reinjury has been found to be the most important reason for not returning to pre-injury sport (Ardern et al., 2012a; Kvist et al., 2005; Tjong et al., 2014). Patients who had undergone ACLR within 3 months after injury had a lower fear of re-injury than those who had waited longer. Those who had RTS to their pre-injury level participated in sport with low fear of re-injury (Ardern et al., 2012a). Motivation to RTS is also an important factor (Tjong et al., 2014). Those with high motivation to RTS preoperatively (Gobbi & Francisco, 2006) and 1 year after ACLR (Ardern, Taylor, Feller, Whitehead, & Webster, 2015) were also more likely to RTS. RTS after ACLR may also be affected by personality factors, such as cautiousness, being pessimistic, a lack of self-confidence, and low levels of self-motivation (Everhart, Best, & Flanigan, 2015; Tjong et al., 2014), yet more research is needed into this. According to (Lentz et al., 2012), incomplete rehabilitation can cause the failure to return to pre-injury participation in sport if athletes are given the go ahead to RTS before their impairment has been fully resolved. However, the lack of research into how exactly to ensure the patient's rehabilitation status means it is difficult to make decisions based on solid evidence. Moreover, there are no standardised clinical guidelines to assist clinicians in making RTS decisions, and there is little consensus on which outcome measures should be used to effectively evaluate the athlete's functional status.

Furthermore, most of the research into the development of postsurgical outcome measures has addressed measures of impairment and disability, with strength, ROM, laxity, girth, and swelling, shown to correlate loosely to measures of activity (Lephart et al., 1992; Ross, Irrgang, Denegar, McCloy, & Unangst, 2002). Therefore, it can be argued that static measures of impairment do not accurately represent the high dynamic load placed on the knee during strenuous sports participation, which throws the validity of current measures for evaluating postoperative levels of sport participation into question, as current constructs are typically based on measures of impairment to assess readiness to return to play. An athlete's functional status is usually measured using functional performance tests, as it is assumed that testing physical performance simulates stresses on the knee experienced during athletic activities. However, as there is a lack of strong research evidence supporting such a correlation, caution is needed when interpreting the results beyond levels of disability in patients that have had ACLR. Ardern et al. (2011) used hop tests to test their usefulness for measuring return to pre-injury levels of activity among patients post ACLR, and they discovered high correlations between level of performance on functional tests and return to pre-injury levels of activity. Even so, further research is needed to further investigate the relationship between functional performance measures and readiness to participate in pre-injury levels of sport. Lephart, Ferris, Riemann, Myers, and Fu (2002) were amongst the first to study the association between impairment based measures of physical function and functional performance. They found a poor correlation ($r = 0.01$ to $r = 0.42$) between measures of strength, laxity, ROM, thigh girth, and functional performance (Lephart et al., 2002). In a similar study, Lentz et al. (2009) explored the relationship between knee impairment, kinesiophobia and function. Their findings demonstrated that pain, quad strength, kinesiophobia and knee flexion restriction correlated with self reports of function only (Lentz et al., 2009). Only knee effusion was associated with a performance based test, the single leg hop test. Barber-Westin and Noyes (2011a) performed a systematic review of published studies to identify which clinical criteria had been investigated over the past decade to determine RTS status following ACLR. Their study revealed serious discrepancies between objective criteria used to make return to sport decisions. Of the 716 studies they identified, only 35 (13%) presented objective criteria for their decisions (muscle strength or thigh circumference: 28; general knee exam: 15; single leg hop test: 10; Lachman:1;) (Barber-Westin & Noyes, 2011a). Barber-Westin and Noyes (2011b) subsequently published a list of proposed criteria for release to full sports participation (limb symmetry index on single leg hop test, quadriceps strength, lack of pain or effusion, full ROM, functional knee stability, surgical and psychosocial factors). The authors should be commended for their comprehensive multifaceted approach; however, the criteria are based primarily on expert opinion. Future study is warranted to validate the predictive value of the various constructs for successful RTS and reduced risk of re-injury. Due to the complexities involved and the physical and mental demands placed on athletes to perform well at sport, a complex diagnostic tool that has been validated and correlates multidimensional deficits with measures of participation would be highly useful. In addition, (Lentz et al., 2009) suggest using a combination of patient self-report and performance-based measures to evaluate function, as they can produce differing conclusions.

Bearing in mind the great personal burden of pain and disability after an ACL injury, it is likely that fear of re-injury is an important psychological variable that could slow down or prevent RTS following ACLR (Kvist et al., 2005). A high fear of re-injury has been shown to correlate with poor self-reporting on function (Kvist et al., 2005). The link between kinesiophobia and sports has been explored using the Tampa Scale of Kinesiophobia (TSK), with Kvist et al. (2005) demonstrating that high TSK score correlate with decreased activity levels among ACLR patients compared with athletes who had returned to pre-injury levels of sport. In addition, the ACL-RSI is being used to measure the psychological impact of returning to sport after an ACLR surgery (Webster, Feller, & Lambros, 2008). Therefore, the impact of kinesiophobia in determining RTS readiness in athletes following ACLR should be researched further. Controversy still exists in the clinical practice of prescribing a functional brace for rehabilitation and return to athletic activities. Proponents of functional bracing following ACLR cite their belief that post-surgical outcomes may be improved by increasing passive knee extension, decreasing pain and graft strain (Möller, Forssblad, Hansson, Wange, & Weidenhielm, 2001; Wright & Fetzer, 2007). Although the use of bracing is widely used, a systematic review does not support its efficacy in improving functional outcomes (Kruse, Gray, & Wright, 2012). Regardless of type of brace (immobilization, functional or rehabilitation) bracing was not found to provide protection against post-operative injury, or decrease pain, alter range of knee motion, or improve stability following ACLR (Kruse et al., 2012). The authors concluded that post-operative bracing provided no benefit, and added an unnecessary expense to rehabilitation. This systematic review supports several previous prospective randomized controlled studies which showed no statistically significant correlation between bracing and measures of strength, functional hop tests, ROM, knee circumference, Lysholm scale, IKDC, and the Tegner activity level scale (Kartus et al., 1997; McDevitt et al., 2004; Möller et al., 2001). In a systematic review to determine if sufficient evidence exists to support the use of post ACLR functional bracing, Wright and Fetzer (2007) reported that brace use provided no improvements in ROM, graft stability or protection from subsequent injury. Although small but significant improvements in static proprioception have been demonstrated with functional bracing, this does not appear to translate into improvements in functional hop tests (Birmingham et al., 2001; Risberg, Beynnon, Peura, & Uh, 1999; Wu, Ng, & Mak, 2001). Current evidence refutes the use of bracing following ACLR, and should be reflected in clinical practice standards.

There is currently a lack of consensus regarding when an athlete should be deemed fit to RTS, with discrepancies between RTS protocols ranging from four to 12 months postoperatively (Cascio, Culp, & Cosgarea, 2004; Kvist et al., 2005). The decision seems to be increasingly based on the athlete's desire to RTS, rather than being based on strong evidence. In addition, surgeons are attempting to push the boundaries to speed up RTS times following ACLR, especially as the success of rehabilitation is often based on how quickly the athlete returns to their athletic career. The high ACLR re-injury rates, leads to the question: Are we returning our athletes to sport before they are safely ready to resume athletic activity, and with insufficient evidence to support our decisions? Fundamental to this question is an understanding of current clinical practice patterns compared with evidence-based measures. Prospectively identifying which of measures effect successful participation in pre-injury levels of sporting activities is essential to developing rules on clinical decisions for athletes returning to sport.

2.3.5 The role of ACL injury in future degenerative conditions

Concomitant joint injury

As a result of the high forces placed on the knee joint during the time of an ACL rupture, concurrent damage to other knee structures often occurs (Noyes, Bassett, Grood, & Butler, 1980). Due to knee trauma at the time of ACL rupture, bone marrow lesions (bone bruises) are apparent in around 70% of knees and may persist for over 12 months following the injury (Papalia et al., 2015; Yu & Cook, 1996). In addition, these lesions can cause a high degree of knee pain (Frobell et al., 2009; Papalia et al., 2015). Even so, bone marrow lesions have not been linked to knee function throughout the first two years after an ACL injury (Papalia et al., 2015); however, research is needed in order to explore whether bone marrow lesions have an impact on longer-term joint function, and/or the development of osteoarthritis (Papalia et al., 2015). At the time of ACL rupture, damage to the articular cartilage may also occur, and full thickness cartilage lesions typically result in greater knee pain, poorer knee function, and reduced quality of life QOL at two years post-surgery (Rotterud, Risberg, Engebretsen, & Aroen, 2012; Røtterud, Sivertsen, Forssblad, Engebretsen, & Årøen, 2013). Early-onset knee osteoarthritis has been associated with concomitant cartilage and meniscus injury, with or without surgical repair, more so than isolated ACL ruptures (Claes, Hermie, Verdonk, Bellemans, & Verdonk, 2013; Keays, Newcombe, Bullock-Saxton, Bullock, & Keays, 2010; Magnussen, Mansour, Carey, & Spindler, 2009; Oiestad et al., 2009; Van Meer et al., 2015). Moreover, patients with baseline meniscal damage reported worse outcomes, such as reduced physical activity level and knee function, and greater pain, 16 years following ACLR (Gerhard et al., 2013). Over 60% of ACLR surgeries conducted in New York state between 1997 and 2006 (70,547 procedures altogether) also involved a concomitant surgery, with one in two ACLRs including concurrent surgery to the meniscus (Lyman et al., 2009). However, concomitant meniscus surgery at the same time as ACLR has been associated with worse outcomes two to 15 years after surgery, such as greater pain, reduced knee function, and poorer QOL, compared to patients who did not undergo meniscus surgery (Barenius et al., 2014; Cox et al., 2014; Dunn et al., 2015; Neuman et al., 2008). This suggests that individuals who suffer associated or additional injuries may be at greater risk of poor QOL outcomes following ACLR surgery.

Subsequent injury and revision surgery

It has been found that individuals who experience a subsequent knee injury after ACLR suffer from more pain, worse symptoms, poorer function and an overall decrease in QOL (Swirtun & Renström, 2008). Additional surgery after ACLR is not unusual, and up to 7% of patients undergo additional knee surgery to either knee within a year post-operatively (Lyman et al., 2009), with 19% of patients having surgery on their ACL-reconstructed knee within six years after ACLR (Hettrich, Dunn, Reinke, & Spindler, 2013). Some patients who have undergone ACLR will experience a re-rupture of the ACL graft, or the contralateral ACL may rupture. However, studies have differed regarding rates of revision, although larger studies found a revision rate of two percent at two years follow-up (Andernord et al., 2015; Bjornsson et al., 2015), and four to five percent at five years follow-up after a primary ACLR (Lind, Menhert, & Pedersen, 2012; Persson et al., 2014; Webster, Feller, Leigh, & Richmond, 2014). Young adults, adolescents and individuals returning to high impact sports are more likely to undergo revision after the primary ACLR (Andernord et al., 2015; Lind et al., 2012; Persson et al., 2014). Moreover, adolescents are at a higher risk of suffering a contralateral ACL rupture in comparison to adults (Leroux et al., 2014; Webster et al., 2014). A study conducted in Australia discovered an average contralateral ACL injury rate of 8% five years after having a primary ACLR, but this figure was far higher, at 29%, among those aged under 20 years (Webster et al., 2014). Rates of sustaining a graft re-rupture or contralateral ACL rupture 15 years after primary ACLR are often as high as one in every four individuals according to Bourke, Salmon, Waller, Patterson, and Pinczewski (2012). According to Leroux et al. (2014), patients who later on need to undergo a re-revision procedure tend to engage in lower levels of activity and suffer more cartilage injuries than those undergoing their first revision surgery. Moreover, it is difficult to determine the exact rate of ACL graft re-ruptures as some incidences may go undiagnosed or not involve surgical reconstruction. In addition, patients that require ACLR revision tend to display meniscal and chondral damage, more so than patients undergoing their first ACLR (Ahn, Lee, & Ha, 2008; Kievit, Jonkers, Barentsz, & Blankevoort, 2013; Thomas, Kankate, Wandless, & Pandit, 2005; Widener, Wilson, Galvin, Marchant, & Arrington, 2015).

Poorer outcomes have been reported following revision surgery in comparison to primary ACLR, such as the development of osteoarthritis, more severe pain and symptoms, reduced activity levels, and poorer QOL (Gifstad, Drogset, Viset, Grøntvedt, & Sofie Hortemo, 2012; Kievit et al., 2013; Lind et al., 2012). If at the time of revision ACLR, a high rate of chondral lesions is noticed, there is likely to be an increased risk of the patient developing osteoarthritis after the revision (Salmon, Pinczewski, Russell, & Refshauge, 2006). While levels of patient expectations for revision ACLR are lower than for primary ACLR, 96% of patients expect there to be no risk or a slight increase in the risk of going on to develop osteoarthritis compared to the healthy knee, 10 years after revision ACLR (Feucht et al., 2016). In addition, 88% percent expect that they will return to the same level of sport as before, and all patients expect to have a normal functioning, or almost normal, knee following revision surgery (Feucht et al., 2016). However, despite such expectations, it has been found that only one out of two patients will return to pre-injury levels of sport after revision ACLR (Grassi et al., 2015), and knee osteoarthritis has been reported in 37% to 80% of patients four to eight years after revision ACLR (Kamath, Redfern, Greis, & Burks, 2011). These results can lead to reduced psychological wellbeing as expectations are not met, as well as reduced QOL.

Knee osteoarthritis

Osteoarthritis is a synovial joint disease that is progressive and causes changes to articular cartilage, subchondral bone, synovium, peri-articular muscles, the meniscus and ligaments (Lane et al., 2011); it is a major cause of disability around the world (Cross et al., 2014) and affects one out of every three individuals over 60 years old (Felson, 2004). There is no curative treatment, making total knee arthroplasty a common option to deal with pain and improve the function of the knee in people with end-stage osteoarthritis of the knee. Rupturing of the ACL has been linked to an increase in the risk of knee osteoarthritis, which may be due to changes in the knee biomechanics; modified joint loading, and a number of other intra-articular pathogenic processes occurring at the time of the ACL rupture (Lohmander et al., 2007). This risk is greater for who suffer a concomitant injury along with the ACL rupture, with one in two developing osteoarthritis of the knee 10 to 20 years after an ACL injury (Oiestad et al., 2009). This is a short time period between ACL injury and the development of osteoarthritis development, and it is particularly concerning when considering the high rate of ACL injury among adolescents and the active populations that usually undergo ACLR (Renstrom et al., 2008). Moreover, Ackerman et al. (2015) found that young and middle-aged adults suffering from knee osteoarthritis have described the greater psychological stress that results, as well as a reduction in health-related QOL and problems related to employment compared to an age-matched population; even so, little research has been conducted into the impact of symptomatic early knee osteoarthritis.

Psychological outcomes

Throughout the acute postoperative period, the psychological impact of an ACL rupture and reconstructive surgery becomes apparent (Brewer et al., 2007; Heijne, Axelsson, Werner, & Biguet, 2008; Langford et al., 2009; Tripp, Stanish, Ebel-Lam, Brewer, & Birchard, 2007), and this can continue for several years after ACLR, perhaps leading to a negative impact on long-term outcomes (Ardern et al., 2012a; Wierike, van der Sluis, van den Akker-Scheek, Elferink-Gemser, & Visscher, 2013). Research has shown that individuals who do not return to pre-injury sport one-year post ACLR, have a greater fear of reinjury, suffer negative emotions and have less confidence when compared to those who do return to sport (Kvist et al., 2005; Webster et al., 2008). In addition, Langford et al. (2009) note that emotional disturbances were described by individuals who had not returned to sport six and 12 months after ACLR, whereas those who had returned to sport were not affected, even though knee function and symptoms were similar. Fear of re-injury is a common psychological outcome following ACLR (Gignac et al., 2015; Kvist et al., 2005; Tripp et al., 2007), and this is concerning because fear of re-injury is linked to worse knee related QOL outcomes (Kvist et al., 2005); moreover, psychological factors before undergoing ACLR are indicative of postoperative outcomes (Everhart et al., 2015). In particular, pessimism (Swirtun & Renström, 2008) and negative predictions of knee self-efficacy in the future (Thomeé et al., 2008) can predict poor post-operative outcomes. Furthermore, an external locus of control that causes individuals to not feel in control of their health is associated with lower self-perceived function before ACLR (Nyland, Johnson, Caborn, & Brindle, 2002). Thomeé et al. (2008) found lower knee self-efficacy one year after ACL injury, and other researchers have discovered poorer functional outcomes and health-related QOL two years post ACLR (Nyland, Cottrell, Harreld, & Caborn, 2006). The aforementioned findings highlight the possibility of ACL injury and reconstructive surgery leading to psychological impacts that can persist in the long term and affect knee function, return to sport rates, and QOL.

Quality of life (QOL)

During the time of an acute injury and the early postoperative periods, quality of life is often affected, and the negative impact may continue until a return to pre-injury knee function or until the individual accepts any restriction they have concerning the function of their knee. (Lynch et al., 2015) explain that ACLRs are usually conducted to make knee function pain free, and to reduce any swelling, or restriction of movement, so that the person can partake in activities. However, it may not be possible for some individuals to return to normal levels of activity due to continued knee problems, or fear of re-injury (Ardern et al., 2014). A number of studies have explored the impact from specific factors on QOL in relation to health and knee function post ACLR. For example, McCullough et al. (2012) found worse knee-related QOL outcomes two years after ACLR for those who did not return to pre-injury levels of sport, and Ardern et al. (2014) found similar results one to seven years after ACLR.

Regarding outcome measures, a full thickness cartilage lesion occurring at the time of ACLR has been linked to the lower Knee injury and Osteoarthritis Outcome Score (KOOS) and QOL scores at two years (Røtterud et al., 2013) and two to five years post ACLR (Rotterud et al., 2012). In addition, research has shown poorer Short-Form 36 (SF-36) scores at two-year follow-ups in patients with concomitant chondromalacia of the lateral tibial plateau when the ACLR was conducted (Dunn et al., 2015). As mentioned previously, a revision ACLR has been shown to lead to poorer knee-related and health-related QOL outcomes within five years post-surgery in comparison to primary ACLR (Bjornsson et al., 2015; Dunn et al., 2015; Kievit et al., 2013; Lind et al., 2012). Additional factors affecting knee-related or health-related QOL following ACLR are fear of re-injury (Kvist et al., 2005), as well as poor performance in single-leg triple-leg hop tests (Reinke et al., 2011).

A number of demographic factors have been found to worsen health-related QOL outcomes within five years of ACLR, which are: being a smoker (Dunn et al., 2015; Kvist, Kartus, Karlsson, & Forssblad, 2014); older age; a high BMI, and poor levels of education (Dunn et al., 2015). Using self-report measures, lower pre-operative knee-related and health-related QOL scores are associated with lower scores on the same measures postoperatively (Bryant, Stratford, Marx, Walter, & Guyatt, 2008; Dunn et al., 2015). In addition, other pre-operative factors that have been found to reduce postoperative QOL outcomes are: lower levels of physical activity (Dunn et al., 2015; Mansson, Kartus, & Sernert, 2013), and reporting anterior knee pain before the ACLR is conducted (Heijne, Ang, & Werner, 2009). Some studies have found no impact from the type of ACL autograft type (patellar tendon, quadruple-stranded or double-bundle hamstring tendon) on Quality of Life Outcome Measures (Questionnaire) for Chronic Anterior Cruciate Ligament Deficiency (ACL-QOL) two years post-operatively (Mohtadi, Chan, Barber, & Oddone Paolucci, 2015); in addition, the KOOS-QOL or Euro-QoL 5D (EQ-5D) scores at one and two years (single or double bundle hamstring autografts) (Bjornsson et al., 2015), and SF-

36 scores at six, 12 and 24 months after ACLR (single or double-bundle) (Nunez et al., 2012; Ochiai, Hagino, Senga, Saito, & Haro, 2012) showed little difference in outcomes. Fleming et al. (2013) state that differences in graft tension, whether low graft tension or high graft tension, were found not to affect knee-related or health-related QOL scores in a three-year follow-up. In addition, a concomitant meniscal lesion or partial-thickness cartilage lesion was not found to affect KOOS-QOL scores two years post-operatively in 3476 patients listed on Norwegian and Swedish Registries (Røtterud et al., 2013). The timing of the operation has also been researched, and it was found that having an early ACLR within four weeks of injury, as opposed to an optional delayed after a structured program of exercise, led to similar SF-36 and KOOS-QOL scores two and five years after an ACL rupture (Frobell, Roos, Roos, Ranstam, & Lohmander, 2010; Frobell et al., 2013).

2.3.6 ACL injury mechanism

The understanding of injury situations and mechanisms could be valuable for injury prevention. ACL injuries are characterized by a contact or non-contact mechanism. A non-contact mechanism involves no contact with an opposing player at the time of injury. Mechanisms of non-contact ACL injury normally involve multi-planar knee loading events (Shimokochi & Shultz, 2008). 70-85% of ACL injuries occur in non-contact situations (Benis et al., 2018; Johnston et al., 2018; Montgomery et al., 2018; Walden et al., 2015). It is often possible to prevent non-contact injuries, which is essential when considering that these are the most common type of ACL injury, therefore, the mechanism of such injuries needs to be better understood to lower the risk of injury.

Benis et al. (2018) conducted a study using questionnaires to explore ACL injury mechanisms, and the majority of respondents stated that their injury occurred while landing or during COD manoeuvre. The individual's ability to recall and comprehend the details of the actual injury and the lack of precise definitions to describe injury mechanisms are important limitations to this relatively inexact approach. Several studies have shown that analysing videotape footage of the occurrences of ACL injuries provides evidence that COD and landing led to the majority of non-contact injuries (Grassi et al., 2017; Johnston et al., 2018; Koga et al., 2010; Krosshaug et al., 2007; Montgomery et al., 2018; Walden et al., 2015); for example, Krosshaug et al. (2007) found that non-contact injuries accounted for 27 out of 39 cases noted in video analysis, and Johnston et al. (2018) discovered 50 out of 69 cases in the videos they analysed. Furthermore, ACL injury happens during the deceleration phase of COD and landing (Koga et al., 2010; Krosshaug et al., 2007). The estimated time of injury ranged between 17 and 60 milliseconds after initial contact (IC)

(Bates et al., 2020; Koga et al., 2010; Krosshaug et al., 2007). Figures 2-3 show the change of direction mechanisms of non-contact ACL injury in rugby players.

Figure 2-3 Change of direction mechanisms of non-contact ACL injury (Montgomery et al., 2018)

In addition to the type of action performed during the occurrence of an injury, the position of the body during these actions also needs to be assessed. Several studies have explored lower limb joint angles by examining videotape footage of the occurrence of ACL injuries (Brophy, Stepan, Silvers, & Mandelbaum, 2015; Grassi et al., 2017; Johnston et al., 2018; Koga et al., 2010; Walden et al., 2015). The results show that almost 60-70 % of non-contact ACL injuries occurred while a player performed a change of direction manoeuvre (Johnston et al., 2018; Montgomery et al., 2018). The results show that the most lower extremity common position of the non-contact ACL injuries at the time of the injury include a lower knee flexion angle (≤30°) and Knee under valgus stress combined with hip abduction and flexion (Boden, Torg, Knowles, & Hewett, 2009; Grassi et al., 2017; Johnston et al., 2018; Koga et al., 2010; Montgomery et al., 2018). The knee was flexed at IC and then had increased till the time of injury, knee abduction was neutral at IC, but had extremely increased till the time of injury, hip was flexed at IC and was nearly constant till the time of injury and hip was abducted at IC and then tendency toward adduction till the time of injury (Boden et al., 2009; Brophy et al., 2015; Grassi et al., 2017; Johnston et al., 2018; Koga et al., 2010; Montgomery et al., 2018). These positions all place a valgus load on the knee, which can lead to ACL injury (Alentorn-Geli et al., 2009; Boden, Dean, Feagin, & Garrett, 2000). This position can be seen in figures 2-3 and has been termed dynamic knee valgus or the ‘position of no-return’ (Hewett et al., 2005; Ireland, 1999).

The majority of ACL injuries occur during non-contact COD manoeuvres (Koga et al., 2010; Waldén et al., 2015). Waldén et al. (2015) assessed videos from 39 complete ACL tears to describe ACL injury mechanisms in male professional football players. They reported that 85% of ACL injuries in football result from non-contact mechanisms. One limitation of this study is that the recorded video sequences had a relatively low frame rate (60 Hz) and limited picture resolution (768 x 576 pixels). This low frame rate and limited picture resolution impedes: the accurate assessment of the anterior translation of the tibia; the estimation of changes in angular velocities, the determination of IC timing, and the identification of the timing when the ACL injury occurred. Although video analyses have been limited to simple visual inspection, the accuracy of these methods has been poor, even among experienced researchers (Krosshaug et al., 2007a). Moreover, a simple visual inspection is not an accurate method to extract a time for joint angle and velocities. It is therefore difficult to determine the exact time of an ACL injury. In contrast, model-based image-matching is a technique that extracts joint kinematics from video recordings, and has been used to try to explain the mechanisms of ACL injuries in more detail (Koga et al., 2010). However, the difficulty in matching body parts due to other players or clothes blocking the view, as well as problems with the assessment of axial rotations, means the results may not totally accurate. Even so, despite these limitations, model-based image-matching provides the most detailed description of ACL injury mechanisms to date.

In-vitro and 3D modelling studies have been conducted to examine the strain on the ACL during certain movements at the knee joint, and these studies confirm the importance of the dynamic knee mechanism. It is possible to achieve knee joint stabilisation through a variety of active muscular and passive ligament control methods, and it is likely that more than one excessive movement is necessary to create enough force to injure the ACL, with forces of at least 1500-2000N required to injure the ACL (Chandrashekar, Mansouri, Slauterbeck, & Hashemi, 2006), although the tensile properties of the ACL are not the same in all individuals, and Chandrashekar et al. (2006) found that maximum load at failure (1266 N (SD 527)) of ACL for female cadavers was lower than for male cadavers (1818 N (SD 699)). The most strain on the ACL is caused by anterior tibial shear causes, although that alone does not produce enough force to result in ligament rupture (Berns, Hull, & Patterson, 1992; McLean, Huang, Su, & Van Den Bogert, 2004). In addition, computer simulations have shown that for extreme cases of sagittal plane injury mechanisms, the force on the ACL did not exceed 900N (McLean et al., 2004). However, a combination of anterior tibial shear and combined knee valgus and/or rotational moments does place much greater strain on the ACL, leading to a greater likelihood of injury

(Markolf et al., 1995; Shin, Chaudhari, & Andriacchi, 2011), in particular at angles close to full knee extension (Berns et al., 1992; Ireland, 1999).

Knee valgus moment induced through poor neuromuscular coordination is one of the more notable biomechanical noncontact mechanisms of injury (Boden et al., 2000; Hewett et al., 2005). Valgus angle and ensuing loads placed at the knee joint during screening measures of drop landings highly predict future ACL injury risk as shown by a prospective study of female soccer players by Hewett et al. (2005). Additionally, a large number of biomechanical mechanisms of ACL injury are dictated by both gender and subject-specific neuromuscular patterns that control kinematic posture during sport-related activities including natural mechanical axes in the lower limb, motor patterns, and kinematic landing variables. Understanding these neuromuscular patterns and how to train against these high-risk patterns is imperative to mitigate injury risk.

2.3.7 ACL risk factors

This section reviews the proposed risk factors for non-contact ACL injuries. If the risk factors for non-contact ACL injuries are better understood, some may be modified and injuries prevented. The risk factors can be categorised into non-modifiable and modifiable factors. Non-modifiable ACL risk factors are those that science is unable to change at this time because of the design of the human and the way the body's systems affect each other. Modifiable ACL risk factors are biomechanical and neuromuscular which can be mitigated against. Obtaining a better understanding of these risk factors should improve identification of them, as well as assisting in discovering better ways of correcting them, in addition to reducing the likelihood of an injury occurring. Non-modifiable risk factors will be discussed in section 2.2.3.1, Modifiable risk factors will be discussed in section 2.2.3.2.

2.3.7.1 Non-modifiable risk factors

Anatomical Risk Factors:

The anatomical variances between individuals are extensive, but the most common differences found to have an impact on ACL injuries are joint laxity due to the surrounding structures; the width of the intercondylar notch; Q-angle size; the width of the pelvis, and the physical size and strength of the individuals ACL.

Joint laxity

Knee joint laxity is a major anatomical factor when it comes to injury where hypermobility of the joint, along with musculotendinous flexibility, leads to laxity. According to Griffin et al. (2000), joint hypermobility is determined by genetics, whereas muscle flexibility can be increased through stretching exercises and conditioning programs. In addition, female tend to have greater levels of tibio-femoral laxity, which leads to less resistance with the femur during rotation and translation (Renstrom et al., 2008). This greater laxity may be why female are more at risk of hyperextension of the knee whilst moving in an extended position. Boden et al. (2000) found that joint laxity among athletes competing in basketball, running and other sports, increased 30 minutes into activity, anteriorly as well as posteriorly, from 18 to 20% compared to at the start (Boden et al., 2000). According to Johannsen, Lind, Jakobsen, and Kroner (1989), who carried out research with long distance runners, the knee's joint laxity increased post-exercise, and decreased within the joint during the recovery period. During walking, the anterior-posterior displacement of the tibia did not reveal much laxity, although an increase of statistical significance was shown with internal-external tibial rotation. Arendt and Dick (1995), used an arthrometer and found that the knees of athletes with intact ACLs as well as those with ACL deficiencies showed anterior laxity. Myer, Ford, Paterno, Nick, and Hewett (2008a) carried out a study with 1,558 female soccer and basketball players over a four-year period. Before the season started, the anterior-posterior knee laxity measurements of the athletes were taken and noted using a CompuKT knee arthrometer. Myer et al. (2008a) discovered that the participants who had injured their ACLs displayed 1.3 mm side-to-side measurements in total anterior to posterior tibiofemoral translation laxity, and so they were three times as likely to suffer an ACL injury. Boden et al. (2000) states that if the medial compartment of the knee joint is constrained, along with the lateral compartment demonstrating laxity, the joint may shift; the lateral tibial plateau might shift anteriorly and rotate internally, which would increase the strain on the ACL and may lead to injury (Boden, Sheehan, Torg, & Hewett, 2010). More research is required in the area of joint laxity, as using an arthrometer may not be suitable when attempting to detect and determine athletes who are at an increased risk of ACL injuries (Arendt & Dick, 1995). In addition to the aforementioned studies, Shultz and Schmitz (2009) concentrated on knee joint laxity combined with poor neuromuscular biomechanics. 96 participants took part in Shultz and Schmitz (2009), with the aim of discovering whether individuals with greater varus and valgus, and internal and external rotational knee laxity, face more difficulties in controlling the motion of the knee in the frontal and transverse planes. Shultz and Schmitz (2009) found that when participants with varus/valgus and internal/external rotational knee laxity at higher levels landed from a drop jump, they displayed greater frontal and transverse knee motions. 4.3 degree more laxity in varus/valgus values and 6.1 degree more internal/external rotational laxity was found in female compared to the males. However, the female's results were similar to the mens' with regard to anterior knee laxity at 6.6 degrees compared to 6.8 degrees respectively (Shultz & Schmitz, 2009). The study went on to reveal that female with higher knee laxity displayed greater hip adduction and knee valgus movements during the early phase of landing, before going on to display greater hip adduction with internal rotation in the final part of the landing. Shultz and Schmitz (2009) has shown that valgus collapse, which puts the ACL at risk, affects female more due to their greater knee laxity.

Femoral notch

A further anatomical risk factor of ACL injury is the size of the femoral notch as it also affects the likelihood of an ACL injury. With regard to the anatomy of the femur, and as highlighted by Griffin et al. (2000), nine studies have examined the differences between healthy and ACL-injured athletes concerning the anatomical structure of the femoral notch. The studies illustrate that the intercondylar notches of individuals with a history of ACL injury were smaller on average, which is similar to the situation between the sexes, as females tend to have a smaller notch width (Chandrashekar, Slauterbeck, & Hashemi, 2005). The risk of ACL may be greater with a smaller femoral notch, as it could constrict some movements, which puts the ACL in highly strained positions. Even so, it has been noted that too much variability exists at present when it comes to the measurements used, and so it is not possible to totally confirm the risk (Griffin et al., 2000).

Pelvis width

The pelvis width of the individual has been found to add to the likelihood of ACL injury, as it alters the Q-angle. A line from the anterior superior iliac spine to the centre of the patella, and the centre of the patella to the tibial tubercle, forms the Q-angle , and so a wider pelvis would create larger Q-angles (Alentorn-Geli et al., 2009). Buchanan (2004) claims that a higher Q angle could place the knee at greater risk of knee abduction. At rest, the knee would automatically be more abducted, and a wide pelvis would result in the femur being in a slightly more adducted position. Studies have shown that knee abduction loads the ACL significantly, while standing and performing dynamic movements (Imwalle, Myer, Ford, & Hewett, 2009; Nessler, Denney, & Sampley, 2017). Pantano, White, Gilchrist, and Leddy (2005), however, claim to have shown that an increased Q-angle as a result of wider hips did not increase the risk of injury, but rather that the ratio between the width of the pelvis and the length of the femur is more important with regard to bony anatomy. A greater ratio of pelvic width to femur length seems to be statistically significant when it comes to both the static and dynamic stresses managed by the knee (Pantano et al., 2005).

Chandrashekar et al. (2005) examined the specific anatomical differences between males and females and found that on average females' ACLs are shorter in length, volume, mass, and cross-sectional area compared to males. This suggests that only lower forces and loads could be tolerated before an ACL injury tear. Chandrashekar et al. (2005) claim that females have less tensile strength in the ACL than males, meaning that they are more at risk of tensile force leading to an injury or rupture. Taking this into consideration along with the other factors mentioned above, it is apparent that a weaker ACL increases the importance of the other anatomical differences between the sexes with regard to the risk of ACL injuries.

Hormonal Risk Factors

The female menstrual cycle and the impact it has on hormone levels with regard to the ACL and risk of injury is an interesting area of study. The higher levels of estrogen in the female body compared to males is important because estrogen can lead to more relaxed soft tissue. The function of estrogen is to lower collagen synthesis and the number of fibroblasts being produced. The fibroblasts of the ACL make collagen, and this forms the main load bearing mechanism of the ACL; therefore it is important to gain a thorough understanding of fibroblastic activities (Boden et al., 2000). Female athletes may, in theory, be more susceptible to ACL injuries as a result of soft tissue becoming weaker because of collagen synthesis levels, which would mean performing at lower strength levels (Boden et al., 2000). Shultz, Sander, Kirk, and Perrin (2005) also claims that high levels of fluctuation in hormone levels cause collagen to weaken after just a few days; moreover, the fibroblasts of the ACL contain receptor sites for the hormones estrogen and progesterone. During the menstrual period, the levels of estrogen circulating within the body vary, therefore, there may be a correlation between this fluctuation in the level of estrogen and collagen strength (Griffin et al., 2000). Hormone levels fluctuate during the woman's cycle by up to 400 fold over four hour periods around ovulation; these great fluctuations in hormone levels lead to disturbances to the body's equilibrium. In their study, Wojtys, Huston, Boynton, Spindler, and Lindenfeld (2002) found that a higher rate of ACL injuries for females in the ovulatory phase of menstruation, and so it could be assumed that ACL injuries are more common throughout the ovulatory phase because of the subsequent increase in estrogen levels. Myklebust, Maehlum, Holm, and Bahr (1998) carried out a study with a Norwegian handball team and found that ACL injuries were more likely in the weeks before, during and after the start of menstruation. The higher risk of ACL injuries at this time is possibly due to tissue laxity as a result of changes in hormone levels before, during, and after the menses (Myklebust et al., 1998).

<u>Genetic:</u>

One of the most recent additions to the list of potential risk factors for ACL ruptures is the possible genetic component. To date, there are data from only three studies which suggest that genetic factors are associated with ACL ruptures. Two of the studies have investigated a familial predisposition to ACL ruptures and only a single study has shown that a specific genetic element is linked to a greater risk of ACL ruptures.

Familial predisposition

A genetic predisposition to ACL ruptures was first shown in a study that was exploring anatomical risk factors for ACL ruptures (Harner, Paulos, Greenwald, Rosenberg, & Cooley, 1994), with the personal data that was gained showing a significant difference in the frequency of ACL ruptures among close family members in patients with bilateral ACL ruptures in comparison to control subjects. Out of 31 (35%) patients with bilateral ACL ruptures, eleven had a family history of ACL ruptures, compared to only one out of 23 (4%) for the control group (Harner et al., 1994). In another case-control study, familial predisposition towards an ACL rupture was examined in 171 patients with an ACL rupture and 171 matched controls (Flynn et al., 2005), and it was found that the former were twice as likely to have a first, second or third degree relative with an ACL rupture compared to those in the control group; in addition, the risk increased slightly when only first degree relatives took part in the research. The strength of this investigation was that data from a large number of participants were available for this study, which made matching of gender, age and primary sport possible. Of the 732 eligible subjects (348 cases and 384 controls), 171 matched pairs were achieved and used in the analysis. The percentage of cases with first, second or third degree family history of ACL rupture was 31%, compared to only 19.3% amongst the control. Similarly, 23.4% of cases and 11.7% of control participants had a first degree family history of ACL rupture (Flynn et al., 2005). Although it appears that familial predisposition is a significant risk factor ACL rupture, the available evidence is insufficient to accurately predict risk.

COL1A1 Sp1 binding site polymorphism

Khoschnau et al. (2008) identified the first genetic sequence variant associated with ACL ruptures. They found that the TT genotype of the COL1A1, which encodes the α1 chain of type I collagen, Sp1 binding site polymorphism was majorly under-represented in participants that had suffered a cruciate ligament rupture. Only one out of 233 participants with an ACL rupture, compared to six out of 358 control subjects, had a TT genotype at the Sp1 binding site within COL1A1 (Khoschnau et al., 2008). However, the lack of data on this specific polymorphism and the low frequency of the rare TT genotype, means the level of certainty that the COL1A1 Sp1 binding site polymorphism is a risk factor for ACL ruptures is low.

Increase body mass index (BMI):

A high Body Mass Index (BMI) is a risk factor for lower extremity injuries generally, and ACL ruptures specifically (Griffin et al., 2006). This is because a higher BMI causes a more extended knee position on landing, and this increased extension during landing increases the likelihood of ACL ruptures (Brown, Yu, Kirkendall, & Garrett, 2007). Uhorchak et al. (2003) conducted a prospective study at a US military academy and discovered that BMI was a significant predictor of the risk of ACL ruptures among female recruits, but not among their male counterparts, with female cadets with a BMI higher than one standard deviation above the mean 3.5 times more risk of developing an ACL rupture. However, Parkkari et al. (2008) found in their cohort study involving 46500 people that no significant association existed between being overweight (BMI > 25kg/m2) and increased risk of ACL ruptures among male (Hazard ratio = 1.1; 95% CI: 0.8 – 1.7) or female participants (Hazard ratio = 1.5; 95% CI: 0.8 - 3.1). However, the heterogeneity of the large population in Parkkari et al. (2008) study makes the data difficult to interpret. Other studies designed to explore the relationship between BMI and injury risk during training, found no such relationship (Knapik et al., 2001). Thus, just one prospective study shows a link between increased BMI and risk of an ACL rupture in females (Uhorchak et al., 2003), whereas other similar research has not shown such an association, making an increased BMI is a low risk factor for ACL ruptures.

Footwear and playing surface:

The athlete’s footwear and playing surface both affect the likelihood of ACL injuries. Playing surfaces are affected by the weather, and this can have an impact on the athlete’s performance, while also increasing the likelihood of an ACL injury occurring. To mitigate this, athletes often prefer footwear with smaller cleats, as this creates more friction and traction when playing surfaces are dry. However, non-contact ACL injuries occur more often when the playing surface is dry according to Griffin et al. (2000) due to the type of cleat worn by the athlete. Renstrom et al. (2008) found that during rugby matches held in Australia, ACL injuries occurred more frequently during times of low rain levels and high evaporation rates, which may have been due to the traction variables between the athlete's footwear and the playing surface decreasing. However, further studies are required to additional investigate the impact of weather on the risk of ACL injuries. Increasing traction with the surface is thought to cause the athlete to be able to perform better. However, this increased traction could increase the risk of the athlete suffering from an ACL injury, particularly when the athlete is fixated and carries out a pivotal motion while decelerating. Lambson, Barnhill, and Higgins (1996) carried out a study that focused on football cleats, and they discovered that athletes wearing cleats meant to increase the friction ratio, also resulted in an increase in the risk of ACL injuries. Boden et al. (2000) also notes higher rates of ACL injury have been reported among athletes who wear footwear with smaller pointed cleats around the interior of the shoe, as this type of cleat increases its torsional resistance, which increases the ability of the cleat to remain fixed, as well as increasing friction with the surface (Boden et al., 2000). Gehring, Rott, Stapelfeldt, and Gollhofer (2007) investigated soccer players wearing cleats with eight round studs at the forefoot and four at the rear of the foot and found higher levels of stimulation for the medial quadriceps femoris muscle. This discovery adds to the assumptions put forward by Lambson et al. (1996) and Boden et al. (2000), in that greater levels of stimulation of the quadriceps results in greater anterior translation and therefore places strain on the ACL.

Although soccer is usually played on a grass surface, it is occasionally played on a surface made of artificial turf, sand or gravel. Injury to the ACL is often presented as a major risk to athletes playing on artificial turf due to the differences in shoe to surface traction. The increase in the frictional force between the athlete's shoes and the surface of the field, along with the extra force needed to release the foot from the surface, is likely to contribute towards both non-contact and contact related ACL injuries. Dragoo, Braun, and Harris (2013) conducted research with a large cohort of NCAA football players, and found a 40% increased incidence of ACL injury for a range of artificial turf types compared with natural grass. On the other hand, Scranton et al. (1997) found that the risk of ACL injury decreased on first generation artificial turf compared with natural grass in professional football players. Moreover, Hershman et al. (2012) and Powell and Schootman (1992) discovered a major increase in the risk of ACL injury on third generation turf compared to natural grass in professional football players. However, most of the studies, including this research, have explored ACL in relation to flat surfaces with no boots (wearing training shoes), which is a limitation; however, this would be a practical example in some training centres and in the rehabilitation pathway for the return from injury. Therefore, the traction between such shoes and surface may not reflect the actual interaction for some sports, such as football, or other sports that are played on grass. There is a further need for research in this area although only a few facilities in the world have access to 'real-life' biomechanical testing where boots can be worn and data can be collected (e.g., Manchester Institute of Health and Performance 3G Performance Capture).

Sport participation (exposure):

In a large population based prospective cohort study with a 9 year follow up, Parkkari et al. (2008), found that participation in organised sports resulted in a significantly increased risk of ACL ruptures. Furthermore, the frequency of participation was related to the degree of risk. Participation in organised sports > 3 times/week resulted in an 8.5 and a 4.0 times increased risk in females and males respectively (Parkkari et al., 2008). This study provides evidence that participation in organised sports is undoubtedly associated with ACL ruptures. The level of competition that the person is playing at is also a risk factor for ACL injury, for example, Myklebust et al. (1998) carried out research with 24 elite European handball teams and found that the risk of ACL injury was 30 times higher during games than practice, and similar findings have been found in other research (Dragoo, Braun, Durham, Chen, & Harris, 2012; Messina, Farney, & DeLee, 1999). Beynnon et al. (2014) carried out an analysis of injury data from eight colleges and 18 high schools, and they discovered that college athletes were at a significantly higher risk of injury from noncontact ACL injury than high school athletes, which suggests that risk for ACL injury increases in line with the level of competition.

2.3.7.2 Modifiable risk factors

Limb dominance:

Influence of limb asymmetry on ACL injury risk is important issue. During match play, the athlete will be challenged to perform the COD manoeuvre in both directions at different angles, with both the preferred and non-preferred limb as the push-off limb. In addition, non-contact ACL injuries may not always happen on the dominant or preferred limb. ACL injury rates for the non-dominant limb range from between 43 and 67 percent (Brophy, Silvers, Gonzales, & Mandelbaum, 2010; Goerger et al., 2014; Matava, Freehill, Grutzner, & Shannon, 2002; Negrete, Schick, & Cooper, 2007). In addition, there are side-to-side differences in neuromuscular control, as well as the biomechanics of the knee, while dynamic manoeuvres are performed, and this is thought to lead to be a risk factor for ACL injury (Pappas & Carpes, 2012; Paterno et al., 2010). Dynamic knee valgus along with limb asymmetry have been associated with a higher risk of ACL injury (Ford, Myer, & Hewett, 2003; Hewett et al., 2005), with limb asymmetry in knee joint loading shown to be higher in athletes who later on suffered an ACL injury (Hewett et al., 2005). Hewett et al. (2005) discovered a link between asymmetrical frontal plane loading at the knee and an increased risk of ACL injury among female athletes. In addition, Paterno et al. (2010) discovered that the amount of internal knee extensor moment asymmetry during initial contact while performing a drop vertical jump was a predictor of future ACL injury. Thus, limb asymmetry during moments is a risk factor for re-injury while playing sport (Schmitt, Paterno, & Hewett, 2012). Kinematic asymmetries may affect muscle–skeletal relations, for example, force–length relationships; change the distribution of the force between the two limbs; put greater load on the inert structures within a single knee, and place disproportionate demands on the musculature of one lower limb. It is likely that unbalanced strength and conditioning leads to a greater imbalance (asymmetry) between the lower limbs, an increased likelihood of injury, and a 10 to 15 percent threshold that requires attention as it poses a risk (Knapik, Bauman, Jones, Harris, & Vaughan, 1991). Furthermore, 138 female collegiate athletes took part in a preseason strength test, where maximal isokinetic torque of the right and left knee flexors and knee extensors, at 30 and 180 deg/sec were measured (Knapik et al., 1991). These individuals were followed up to assess for injuries during their subsequent sporting seasons. Forty percent of the female athletes went on to have one or more injuries, and they experienced more lower extremity injuries if they had a 10-15% stronger limb than the injured side (Knapik et al., 1991). Paterno, Ford, Myer, Heyl, and Hewett (2007) investigated whether female athletes show lower limb asymmetries during landing and takeoff force after ACLR when they were cleared to return to competitive sport. Their sample comprised fourteen female athletes at an average of 27 months after ACLR, along with 18 healthy female athletes. All of the subjects were asked to perform a drop vertical jump (DVJ) task on two force plates. Paterno et al. (2007) found that although the ACLR group demonstrated side-to-side asymmetries during the takeoff as well as the landing phase of the DVJ, the control group showed no side-to-side differences in the takeoff or landing phases. Moreover, these asymmetries were still apparent up to two years after ACLR surgery. Paterno et al. (2007) note that the non-involved limb encountered increased forces in the landing phase of the

DVJ, and this has been suggested as a mechanism that is highly related to ACL injury (Hewett et al., 2005; Olsen et al., 2004).

In addition, Kyritsis et al. (2016) found the risk of an ACL re-tear in male athletes was four times higher among individuals who RTS without reaching > 90% symmetry for muscle strength and hopping tasks. Hewett et al. (2005) conducted a prospective study with female athletes that highlighted the factors that could predict ACL injury. They found that uninjured female athletes who later experienced a noncontact ACL injury, had a VGRF that was 20% higher than the cohort who did not subsequently suffer an ACL injury when landing during the DVJ maneuver. In addition, two years after ACLR the athletes who subsequently suffered an ACL injury showed higher forces at an increased rate during manoeuvers. In addition, symmetrical muscle strength prior to a return to sports may be required to decrease the risk of further injury (Grindem et al., 2016). Grindem et al. (2016) discovered that symmetrical quadriceps strength is linked to lower ACL re-injury rates in male and female patients. Moreover, 38% of patients who failed the RTS criteria (≥ 90% LSI in strength) went on to suffer re-injury. Therefore, reducing between-limb biomechanical deficits could be a potential training strategy to reduce the relative risk of non-contact ACL injury.

Differences in the biomechanics of the lower extremity between the dominant and non-dominant limb have been examined during kicking (Ball, 2011; Dorge, Anderson, Sorensen, & Simonsen, 2002) and hopping manoeuvres (Van Der Harst, Gokeler, & Hof, 2007). However, very limited research exists on limb asymmetry during COD. These studies only addressed limb asymmetry during COD for small angles, such as 45 degree. Although a player sustains ACL injuries during COD at various angles (30-180) (Waldén et al., 2015; Montgomery et al., 2018), surprisingly, to date, no study exists on limb asymmetry during COD at 90 and 135 degrees. Knee kinetics and kinematics contribute to the increase angle of COD manoeuvres, where sharper angles affect knee biomechanics in which COD manoeuvres at 90° and 135° angle compare to 45°; thus the mechanical demand placed on the knee increased which meant the ACL injury risk increased (Havens & Sigward, 2015a; Schreurs et al., 2017). A recent systematic review evaluates literature in terms of the effect of limb dominance on COD biomechanics associated with an increased risk of ACL injury and they found conflicting results (Dos'Santos, Bishop, Thomas, Comfort, & Jones, 2019a). Brown, Wang, Dickin, and Weiss (2014b) discovered differences between dominant and non-dominant limbs in healthy females during planned COD to 45° angles; this impacted the knee flexion angle, knee abduction angle, knee internal rotation angle and knee abductor moment, potentially placing greater strain on the ACL, and leading to a greater likelihood of injury (Markolf et al., 1995; Shin, Chaudhari, & Andriacchi, 2011).

In addition, Pollard et al. (2018) examined the differences between dominant and non-dominant limbs in 31 healthy participants (comprising 15 males and 16 females) during planned COD at 45° angles. They found that healthy individuals showed significant difference between the two limbs in which the peak knee internal rotation angle was greater in the non-dominant limb. In terms of other joints, Marshall et al. (2015) investigated differences between dominant and non-dominant limbs during COD at 105° manoeuvres in healthy males and found that ankle internal rotation moments were significantly greater in the non-dominant limb. Moreover, ankle joints were also significantly more dorsiflexed on the non-dominant side.

In contrast, Greska et al. (2016) studied the biomechanical differences between the dominant and non-dominant limb during the performance of a COD manoeuvre at a 45° angle. They found that collegiate female soccer athletes displayed similar lower extremity biomechanics between dominant and non-dominant limbs with regard to peak knee abduction angles and moments. This is in line with other previous research that has shown a lack of difference between dominant and non-dominant limbs while performing 45° COD manoeuvres (Bencke et al., 2013; Matava et al., 2002). Recently, there has been an increase in studies comparing the biomechanical differences between limbs after ACLR, and it has been shown that kinematic and kinetic differences between limbs are apparent during COD tests nine months post ACLR (King et al., 2018). This was found with the lower knee flexion angle, ankle external rotation moment, knee external rotation moment and knee extension moment, as well as lower knee internal rotation in the ACLR limb (King et al., 2018).

It should be noted that previous studies only investigated the differences between limbs during COD manoeuvres at 45°; meanwhile, the current study will utilise different COD angles (at 90° and 135° degrees). It is essential to address limb asymmetry during 90° and 135° COD manoeuvres because the angles of a COD manoeuvre at the time of non-contact ACL injuries ranged between 30-180° (Montgomery et al., 2018; Waldén et al., 2015). In addition, the mechanical demand placed on the knee increased the ACL injury risk increased during 90° and 135° COD manoeuvres compared to 45° COD manoeuvres. However, no study exists on the limb asymmetry during 90° and 135° COD manoeuvre. Furthermore, all previous studies have defined the preferred limb as that which is preferred to kick the ball with. However, in the current study the preferred limb was determined by asking participants which limb they would prefer to use for push-off during COD manoeuvres.

Defining lower limb dominance is not straightforward and varies between the limb chosen for kicking (Gabbard & Hart, 1996); strength (Jones & Bampouras, 2010); braking after being pushed (De Ruiter, De Korte, Schreven, & De Haan, 2010); jumping (Kobayashi et al., 2013); perception (Kong & Burns, 2010); spontaneity of leading limb on a step-up task (De Ruiter et al., 2010), or a combination of these (De Ruiter et al., 2010). It should also be borne in mind that limb dominance can be task-specific (Velotta, Weyer, Ramirez, Winstead, & Bahamonde, 2011), as the athlete adapts their moves to the task they are performing. Therefore, it is important to note that limb dominance is often defined differently based on the author and task. For example, this might include choosing dominance based on which limb the performer prefers to use when performing a COD task. Players using both limbs allow for variety when performing COD manoeuvres which offer a tacticial advantage. However, players may not typically use their dominant limb (the preferred limb to kick the ball with) as a push-off limb during COD manoeuvres; instead, they may use the preferred limb during COD manoeuvres rather than dominant limb. Moreover, previous studies investigating biomechanical symmetry in COD manoeuvres have typically performed the analysis using one discrete point (e.g. peak values). However, there are a number of limitations with this type of analysis. First, asymmetry may occur over phases that are not captured in a single data point. Second, the discrete points utilised may vary between studies; therefore using only one discrete point of analysis may not detect all significant asymmetries. In the current study, biomechanical symmetry during 90° and 135° COD manoeuvres have performed investigated using both multiphase and discrete points (at; IC, PVGRF, 60 ms, PEKAM and PKVA). Analysing the kinetic and kinematic variables during different phases could provide a more complete analysis and a greater understanding than only considering one discrete point.

Ultimately, it remains questionable for COD that limb dominance is an ACL injury-risk (Dos'Santos et al., 2019a). Moreover, there is a lack of scientific data describing the relationship between limb preference and knee mechanics and how these variables affect risk factors for ACL injury (Dos'Santos et al., 2019a). Therefore, more investigation is necessary to determine whether biomechanical asymmetries exist between the preferred and non-preferred lower limbs during COD manoeuvres at 90° and 135° angles.

The impact of limb dominance on non-contact ACL injury risk requires more research to improve our understanding of the link between biomechanics of the preferred and non-preferred limbs during dynamic manoeuvres, including COD. This will assist clinicians when conducting rehabilitation programmes and advising on injury prevention. Analysing asymmetries across hip and knee joints, rather than focusing on a single motion, should lead to a more comprehensive assessment of incorrect motion patterns that could lead to injury. Furthermore, investigating asymmetries among 90° and 135° COD manoeuvres could help to identify the most useful manoeuvre to include in screening.

Therefore, one purpose of this thesis is to determine whether differences in lower extremity neuromechanical exist in ACL injury risk factors between preferred and nonpreferred limbs as push-off limbs during COD manoeuvres. In this thesis we chose two distinctly different COD manoeuvres, namely 90° and 135°. Because players get non-contact ACL injuries during 90° and 135° COD manoeuvres (Waldén et al., 2015) and no study has investigated these COD biomechanic asymmetries from risk of injury perspectives. Instead, we currently only have biomechanics on COD at < 90°. Furthermore, 90° and 135° COD manoeuvres place the knee at greater risk compared to 45° COD manoeuvres (Havens & Sigward, 2015a; Schreurs et al., 2017).

Previous injury:

A major risk factor for suffering a new injury in the same location is a previous injury, probably as a result of the failure to fully complete rehabilitation and returning to play too early. While for senior female players, no link was found between previous knee and ankle injuries and new injuries in the same location (Faude, Junge, Kindermann, & Dvorak, 2006; Soderman, Alfredson, Pietila, & Werner, 2001). Among female elite football players, the risk of ACL rupture was significantly higher among players who reported a previous ACL injury (Faude et al., 2006). In addition, among female youth players, an increased risk of a new injury was found in those with a previous injury (Kucera, Marshall, Kirkendall, Marchak, & Garrett, 2005; Steffen, Myklebust, Andersen, Holme, & Bahr, 2008). Steffen et al. (2008) found that the increase in the risk of injury correlated with the number of previous injuries, and other studies have reported similar outcomes for male professional football players (Arnason et al., 2004; Hägglund, Waldén, & Ekstrand, 2006; Walden, Hagglund, & Ekstrand, 2006).

A past ACL injury is a major risk factor for subsequent re-injury, whether in the contralateral knee or reinjury of the ACL graft (Gianotti et al., 2009; Orchard, Seward, McGivern, & Hood, 2001; Walden et al., 2006). This can be blamed on several factors, such as surgery that was suboptimal; muscular weakness and imbalance; weakened ligaments; changes to kinematics, and lower proprioception following the initial injury (Hewett, Di Stasi, & Myer, 2013; Murphy, Connolly, & Beynnon, 2003). Some studies have found that the risk of future ACL injury is higher for the contralateral uninjured limb compared than the previously injured limb (Boden et al., 2000; Hewett, Myer, & Ford, 2006). The incidence of ACL injury within two years of ACLR and subsequent return to sport was 6-fold higher in athletes with a history of ACL injury compared to uninjured athletes (Paterno, Rauh, Schmitt, Ford, & Hewett, 2014). The greatest risk for re-injury is during the period 12 to 24 months after ACLR, which is usually when athletes return to competitive sport (Paterno, Rauh, Schmitt, Ford, & Hewett, 2012; Paterno et al., 2014). In athletes with a history of ACL injury, deficits in proprioception and their range of motion could alter how their coordination when performing previously learnt movements (Paterno et al., 2012, 2014). For example, Walden et al. (2006) conducted a prospective study with elite soccer players to find out whether ACL reconstruction significantly predicts repeated injury to ACL graft or injury to the contralateral knee. He found a higher incidence of new knee injuries of any type among the soccer players with a history of ACL reconstruction compared with players with no history of an ACL injury (Walden et al., 2006). In addition, Orchard et al. (2001) discovered that a previous ACL reconstruction is a major risk factor for noncontact ACL injury in the reconstructed as well as the contralateral knee. Orchard et al. (2001) found that patients with a previous ACL injury that occurred within the previous 12 months, were 11.3 times more likely to suffer an ACL injury compared to their uninjured counterparts (Orchard et al., 2001). Similarly, those who experienced an ACL injury before the past 12-month period were 4.4 times more likely to suffer an injury to the graft or the contralateral ACL compared those who had not suffered an injury (Orchard et al., 2001). This suggests greater care should be taken from early to late stage rehabilitation to ensure successful recovery before players return to competition in their respective sports (Walden et al., 2015). It may be that after ACLR, persistent neuromuscular and biomechanical risk factors occur which render athletes at greater risk of future re-injury (Alentorn-Geli et al., 2009; Hewett, Lindenfeld, Riccobene, & Noyes, 1999; Hewett et al., 2006; Mandelbaum et al., 2005). Faude et al. (2006) discovered an increased risk of new ACL injury among elite female football players with a previous ACL injury, although no association between previous injuries and new injuries for other lower extremity injuries was found. On the other hand, several injuries have reported this as being a significant predictor of new injuries among youth female players (Steffen et al., 2008) as well as male football players

(Arnason et al., 2004; Engebretsen, Myklebust, Holme, Engebretsen, & Bahr, 2010; Hägglund et al., 2006; Walden et al., 2006)

Biomechanical Risk Factors:

Biomechanical risk factors occur as a result of the movement in the sagittal, frontal and transverse planes contribute to ACL injury, and can usually be corrected. The following sections will review the factors that arise in the sagittal, frontal and transverse planes of movement and how they might influence non-contact ACL injury risk.

Sagittal plane:

Changes in sagittal plane angles at the knee can alter the load imparted on the ACL (Quatman, Quatman-Yates, & Hewett, 2010). Knee flexion angle seems to have an impact on non-contact ACL injury, as research has revealed a reduced ACL load when the knee flexion angle increases (Dai, Mao, Garrett, & Yu, 2014), and ACL strain is usually highest at angles near to full extension (Berns et al., 1992; Markolf et al., 1995). Cadaveric studies provide a better idea about exactly what movements place stress on the ACL, including Anterior Tibial Shear Force (ATSF), which is a type of force that occurs at the knee, involving the anterior motion of the femoral condyles on the tibial plateau. The purpose of the ACL is to reduce ATSF, as without an ACL, there is little to stop the femur sliding off the superior surface of the tibia. ATSF is not harmful, but is important to ensure normal movement; however, too much ATSF can cause the ACL to be overloaded, resulting in a tear. The ATSF is reduced by increasing the knee flexion during movement (Markolf et al., 1995). An anterior shear force to the proximal tibia via the patellar tendon is caused by the contraction of the quadriceps muscle (DeMorat, Weinhold, Blackburn, Chudik, & Garrett, 2004), and it could reach a level that is enough to result in ACL micro-trauma if knee flexion is between 10 to 30 degrees (Griffin et al., 2000). Pandy and Shelburne (1997) found a reduction in knee flexion among female athletes, along with increased the activation of the quadriceps and reduced activation of the hamstring, contributes towards placing greater strain on the ACL and an increased likelihood of injury, which is related to the position typically observed during ACL injury episodes (Johnston et al., 2018; Koga et al., 2010; Krosshaug et al., 2007; Montgomery et al., 2018; Olsen, Myklebust, Engebretsen, & Bahr, 2004). Furthermore, if the knee flexion angle increases, the line of action of the quadriceps will change, and this will reduce its potential to cause anterior tibial shear. DeMorat et al. (2004) conducted a study that shows that the quadriceps force of 4500N applied to the patella tendon can cause an ACL injury at a knee flexion of 20° in six out of 11 cadaveric knees. In comparison, the maximum voluntary quadricep contraction at a 15° knee flexion is almost 3000N (Van Eijden, Weijs, Kouwenhoven, & Verburg, 1987), which is lower than the 4500N quadriceps force previously claimed to risk ACL injury through quadricep contraction (DeMorat et al., 2004). Moreover, to generate the shear component of 1500-2000N supposedly required to damage the ACL it is unlikely that the quadriceps would be contracted in isolation to this extent without other muscles being involved (Chandrashekar et al., 2006). It is also important to bear in mind that compression forces; the dissipation of landing forces at the ankle and hip, and the synergistic action of hamstrings and quadriceps could lower the forces placed on the ACL (McLean et al., 2004). This makes it unlikely that only anterior shears will cause the 1500-2000N load needed for an injury to occur to the ACL (Chandrashekar et al., 2006; Woo, Hollis, Adams, Lyon, & Takai, 1991).

Markolf, Gorek, Kabo, and Shapiro (1990) examined the extent of forces on the ACL from different positions using knees from cadavers. The study found that frontal plane kinetics, that is knee adduction and abduction moments, created an increase in tension on the ACL. They discovered that strain on the ACL as a result of a knee adduction moment was lower with greater knee flexion; whereas the ACL sustained increased strain when the knee abduction moment was between zero and 30 degrees of flexion. However, several researchers have disagreed with the theory of a single plane injury mechanism (DeMorat et al., 2004; Quatman et al., 2010), and claim that knee flexion cannot predict ACL injury (Hewett et al., 2005), with an isolated sagittal-plane force not enough to damage the ACL (McLean, Huang, & van den Bogert, 2008). ACL injury is not caused directly by knee-flexion angle, but it places extra secondary stresses alongside other risk factors, affecting the frontal and transverse planes.

Frontal and transverse planes:

Changes to the hip and knee frontal and transverse-plane motion and loading throughout functional activities have been described as ‘apparent knee valgus’, ‘dynamic valgus’ and ‘dynamic misalignment’. Moreover, knee frontal plane loading has been shown to play an important role in ACL injury (McLean et al., 2004). During landing, running and COD manoeuvres, excessive knee valgus angles and moments are associated with ACL injury (Hewett et al., 2005; Myer, Ford, Khoury, Succop, & Hewett, 2010). Some of the positions of ACL injury identified during dynamic movement are knee abduction collapse with slight knee extension (0-30 degree), and internally rotation of the tibia while the foot is placed on the ground (Johnston et al., 2018; Koga et al., 2010; Montgomery et al., 2018). In addition, external knee abduction moment has been found to place a high force on the ACL (Hewett et al., 2005). A quantitative analyses of injury events found that rapid knee abduction and internal rotation during the early weight-bearing phase tends to happen at the time of injury. Although flexion can change the amount of force placed on the ACL, the interaction with the frontal plane movement is a cause for concern when ACL stress continues at a high level during a valgus force being applied (Markolf et al., 1995). In a similar study, Shin et al. (2011) utilised a three-dimensional cadaveric knee apparatus to test the stress placed on the ACL as a result of a variety of different motions and angles. Shin et al. (2011) found that a combination of knee abduction moment and tibial internal rotation moment placed the most strain on the ACL. Therefore, the claim that if the knee had been subjected to the level of force produced in a full-speed athletic manoeuvre, the strain on the ACL would be even higher than the threshold of rupture, thus greatly increasing the risk of injury. The study did not find significant enough strain to cause injury from knee abduction or tibial internal rotation alone.

Hewett et al. (2005) carried out an important prospective study of ACL injury, with regard to frontal plane factors. The data for those who were injured during the season and those who were not was then compared, and it was revealed that the female athletes that injured their ACL showed 8.4 degrees greater knee abduction angles during initial contact, and 7.6 degrees greater of peak knee abduction angles compared to the uninjured female athletes. In addition, knee abduction moments (KAM) were 6.4 times greater when comparing the differences between their two limbs for those who suffered an injury. Hewett et al. (2005) discovered major increases in KAM among athletes who go on to sustain an ACL injury, but Kristianslund and Krosshaug (2013) believe that the drop vertical jump manoeuvre used in the study was not enough to produce significant KAM. Therefore, they carried out a comparison between drop vertical jumps and COD manoeuvre with 120 elite female handball players from the Norwegian team. The drop vertical jump manoeuvre was almost identical to the one implemented by Hewett et al. (2005). The COD manoeuvre was performed on average 3.4 meters per second, at an angle of 69 degrees. Although a peak was noted in knee abduction moments just after initial contact during COD manoeuvre, no peak was noted at this point with the drop jump, and the knee abduction moments during the COD manoeuvre were six times greater than those during the drop jump manoeuvre. In addition, a 5.8 degree greater knee abduction angle was noted for COD compared to the drop jump; therefore, it might be that an athlete with a measurable knee abduction moments throughout a drop jump manoeuvre may have even higher moments when carrying out a COD manoeuvre during a real-life sporting event. In fact, a more sport-specific COD manoeuvre may be more likely to cause knee abduction moments and angles to be produced that could result in an ACL injury, compared to a drop jump. Thus, according to Hewett et al. (2005) and Kristianslund and Krosshaug (2013), knee abduction moments and angles, both at initial contact and peak values, are significant predictors of ACL injuries occurring.

The manipulation of the position of the body position during different motions is also useful, as this may highlight the impact of stresses on the ACL. Such manipulations enable various body positions to be assessed during a manoeuvre, and the participant's movements are more controlled, which gives a higher rate of accuracy. Thus, Dempsey et al. (2007) carried out a study with 15 men and asked them to perform a COD manoeuvre while in a variety of manipulated body positions (e.g. neutral position, internally rotated foot, externally rotated foot, wide foot placement, narrow foot placement, torso rotation, lateral flexion). They then compared the stresses on the ACL that were found by using markers placed on 50 locations around the body. In this way, Dempsey et al. (2007) found that compared to a neutral body position, a wide foot stance; the torso leaning in the opposite direction to the COD, and the torso rotating away from the direction of the COD, all caused significantly higher valgus or internal rotation stresses on the ACL. If the foot is internally rotated in the direction of the cut, however, the stresses on the ACL decrease. In addition, a wide foot-placement stance during a COD manoeuvre was found to increase both the peak knee valgus stresses and the peak internal rotation at the knee, compared to a normal or narrow stance. Alongside decreased knee flexion, a wide stance results in extreme forces on the knee and the ACL in particular, making it susceptible to injury. Dempsey et al. (2007) discovered that athletes can reveal increased knee valgus or internal rotation at the knee throughout COD manoeuvres. Hewett et al. (2005) claim that ACL injuries may be as a result of the movement being too fast for reflexive contraction and muscle protection. Thus, athletes should consider altering the way they COD so as to protect themselves from movements that will potentially cause an abducted knee position so that they can avoid this risk factor. The suggestion by Hewett, Stroupe, Nance, and Noyes (1996) to ensure a decrease in knee abduction moments by correcting jumping movements, might reduce the risk factors for ACL injuries. Thus, introducing training to improve knee control during COD and landing may be helpful, as shown by Myklebust et al. (2003a) who found that the incidence of ACL injury among female athletes could be reduced, along with reducing the extent of high-risk movements. Thus, training athletes to land or COD with the knee in a safer position following an unexpected load could reduce the risk of injury. The altering of the athlete’s mechanics is dependent relies on the idea that neuromechanical risk factors can be addressed through somatosensory and proprioceptive input to deal with risky motor commands. Dempsey, Lloyd, Elliott, Steele, and Munro (2009) discovered that the risks posed by increased knee valgus or internal rotation at the knee during a COD motion, was reduced by implementing an intervention program. It is important to make athletes and coaches aware of this and train them in the prevention of injury, especially when an athlete is deemed to be at risk.

Although isolated sagittal, frontal and transverse plane factors have been found to increase the risk of an ACL injury, combined knee loading across several planes causes the largest ACL loads and hence presents the greatest risk (Markolf et al., 1995; Shin et al., 2011). Thus, ACL injuries seem to occur due to a multi-planar mechanism (Kiapour et al., 2014; Quatman et al., 2010). Furthermore, knee abduction in combination with internal knee rotation, anterior tibial translation, and increased tibial compression, leads to ACL injuries that are in line with clinical observations of ACL injuries (Levine et al., 2013).

2.4 Injury prevention

In England, the rate of ACL reconstruction increased 12-fold from 2.0 per 100,000 population in 1997–1998 to 24.2 per 100,000 in 2016–2017 (Abram et al., 2019). In addition, the incidence of ACLR in the United States saw an increase from 86,687 (32.9 per 100K) in 1994 to 129,836 (43.5 per 100K) in 2006 (Mall et al., 2014). The reasons for the increase are uncertain but could include an adjustment in intervention rates, increased injury rates, increased surgeon numbers, changes to healthcare commissioning or the development of patient treatment preferences. This massive increase to the ACL injury rate suggests that research into ACL has not led to a reduction in injury rates.

Significant reductions in non-contact ACL injury rates have been shown post intervention when compared to previous injury rates and control groups (Hewett et al., 1999; Myklebust et al., 2003a). However, numerous studies have demonstrated no difference in injury rates between control and intervention groups (Heidt, Sweeterman, Carlonas, Traub, & Tekulve, 2000; Pasanen et al., 2008; Pfeiffer, Shea, Roberts, Grandstrand, & Bond, 2006). Hewett et al. (1999) investigated the effects of a six-week neuromuscular training programme, which included flexibility, strengthening and plyometric exercises, on injury rates among 1263 high-school soccer, basketball and volleyball players. The programme was 60-90 minutes in length and was completed three times per week, becoming progressively harder throughout. In the trained female group, there were no non-contact ACL injuries, whereas in the untrained female group, there were five noncontact ACL injuries, which is similar to the rate for the respective male samples (Hewett et al., 1999). Although there was a high number of athletes in the study sample, only six non-contact ACL injuries occurred, which limited the generalisation of the findings. A continuous programme that aimed to improve balance and landing techniques among women handball players was conducted by Myklebust et al. (2003a) who assessed the effect of the programme over three seasons. The first was the control season when 29 ACL injuries occurred, which is 0.14 per 1000 player hours. During the second and third seasons, the intervention programme was applied. In the second season, 23 ACL injuries occurred which is 0.13 per 1000 player hours and is similar to the control season. In the third season, the number of ACL injuries was lower at 17, which is 0.09 per 1000 player hours; even so, this difference is not significant. However, although there was a reduction in non-contact ACL injuries, the players that suffered injuries during the control season may have been more prone to an ACL injury, which would have reduced the chance of more ACL injuries during the intervention season.

In contrast, some studies have found no effect on the ACL injury rates of other intervention programmes that aim to improve balance, strength, landing technique and agility (Heidt et al., 2000; Pasanen et al., 2008; Pfeiffer et al., 2006; Steffen, Myklebust, Olsen, Holme, & Bahr, 2008). For some of these studies, the small sample size and low number of injuries could have led to insufficient statistical power to detect differences; for example, Heidt et al. (2000) found one ACL injury in the intervention group compared with eight in the control group. Even though this is a clear difference, it did not reveal a statistically significant difference. On the other hand, Pfeiffer et al. (2006) discovered higher rates of non-contact ACL injury in the intervention group (0.107/1000 AE) compared to the control group (0.078/1000 AE) from among a sample of over 1400 female high school athletes.

Despite these differing results, it is clear that biomechanical measurements, such as technique, joint loading, and muscle support, during COD manoeuvres should be considered and measured at the same time as changes in ACL injury rates. This should help to identify the biomechanical mechanisms by which training influences the factors linked to noncontact ACL injuries and the reasons for specific training protocols resulting in positive or inconclusive outcomes. In summary, the main mechanism involved in an ACL injury is the forces applied to the ligament, which is greater than its ability to sustain the load from those forces (Lloyd, 2001). Therefore, ACL injury prevention programs ought to concentrate on reducing the loads applied to the knee joint and, in turn, ACL injury during sporting activities. The externally applied joint loads, and the strength of the muscles capable of supporting these loads affect the loads applied to the ACL. Thus, the focus of biomechanical training interventions is to lower these external joint loads, and/or to improve muscular support. In addition, it is important for training interventions to target the causal factors associated with ACL injury (Lloyd, 2001). However, the role of the hip biomechanic to reduce ACL risk factors are still unknown during 90° and 135° COD manoeuvres. Therefore, identifying the links involved should support the provision of more effective ACL injury prevention/rehabilitation training programs, leading to a reduction in ACL injury rates.

2.4.1 Injury Prevention Framework

To improve our understanding of the various factors involved and their interaction, and to provide a framework for the injury prevention process, sport injury models have been developed. The two most widely recognised models of sports injury prevention are the sequence of injury prevention (Van Mechelen et al., 1992) and Translating Research into Injury Prevention Practice (TRIPP) (Finch, 2006). Both of these are based on injury surveillance, the identification of risk factors for injury, and the implementation and evaluation of injury prevention strategies.

Van Mechelen et al. (1987) published the ‘sequence of prevention’, which has been used in a range of research studies (see Figure 2-4). The model provides a four-stage process to examine the prevention of sports injuries. The first stage involves surveillance of the injury in order to discover the extent of the problem. Injury epidemiology is the study of the occurrence of injuries, including how to deal with the incidence of injuries, and the potential control of factors involved in those injuries. This information creates an overview of the rate and severity of injuries, as well as the distribution of injuries across the body and the tissues prone to injury. This is important information that is key to planning and implementing forms of injury prevention. In England, between 1997–1998 and 2016–2017, there were 133,270 ACL reconstructions carried out on 124,489 patients. Furthermore, the rate of ACL reconstruction increased 12 fold, from two per 100,000 people in 1997–1998 to 24.2 per 100,000 people in 2016–2017 (Abram et al., 2019). Moreover, 70-85% of ACL injuries happen during non-contact events (Benis et al., 2018; Johnston et al., 2018; Montgomery et al., 2018; Walden et al., 2015), and out of these, most occur during COD manoeuvres (Grassi et al 2017; Walden et al., 2015). It may be possible to prevent non-contact injuries, which is important considering that they are the most common type of

ACL injury. Therefore, the mechanisms involved in these injuries need to be fully understood in order to reduce the risk of injury.

The second stage of the model of injury prevention is to establish the mechanisms of injury involved, and the factors associated with the cause and severity of the injury; a biomechanical focus is required because injury surveillance cannot directly establish the mechanisms involved in injuries (Krosshaug, Andersen, Olsen, Myklebust, & Bahr, 2005). More detail about the second stage is provided in the next section (Mechanical Aetiology of ACL Injury). Thirdly, preventative measures are developed and introduced, which aim to reduce future incidences and/or the severity of injuries. Finally, the impact of these preventative measures should be assessed by returning to stage one in order to discover the extent of the problem after the training programme. Therefore, the sequence of injury prevention clearly presents the processes required to provide an evidence base for sports injury epidemiology, and it states that these should be established alongside the causative factors for those injuries, before an injury prevention measure is implemented.

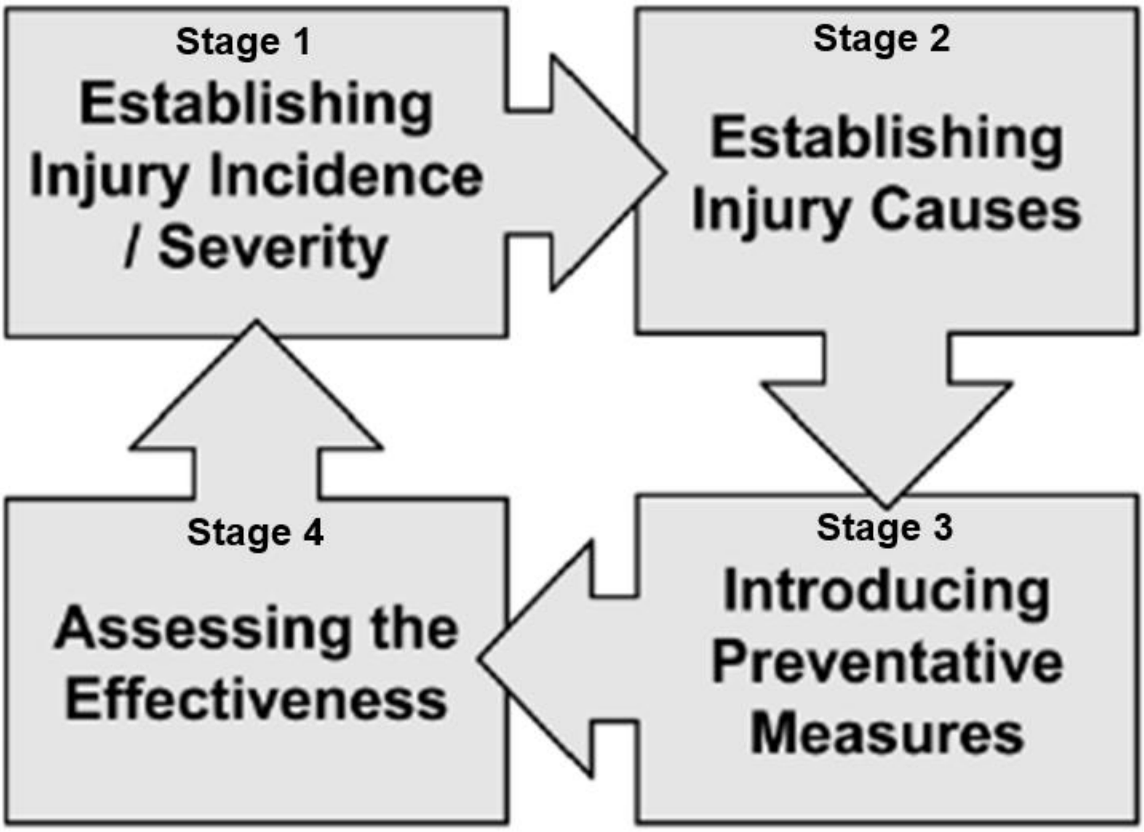


Figure 2-4 A four-stage process of the sequence of injury prevention (Van Mechelen et al., 1992).

Finch (2006) developed the ‘sequence of prevention’ model further, as the Translating Research into Injury Prevention Practice (TRIPP) framework includes two additional stages (see Figure 2-5). Stage five addresses an understanding of how the outcomes from the previous four stages can be implemented in a real-life sporting setting, with the final stage aiming to implement the intervention in a real sporting context to evaluate its effectiveness. Finch (2006) claims that these extra two stages are needed in order to ensure the acceptance of injury prevention measures, and that the athletes they are targeting follow them. The primary considerations of this thesis reflect Stage 2 of the sequence of injury prevention (Van Mechelen et al., 1992) and a TRIPP model (Finch, 2006), which is to establish the mechanism of the ACL injury because we need to understand the mechanical aetiology of ACL injuries during COD manoeuvres to identify and prescribe appropriate interventions for prevention and rehabilitation. This will be reviewed in greater detail in the following section (2.4.2).

Model stage	TRIPP
1	Injury Surveillance
2	Establish aetiology and mechanisms of injury
3	Develop Preventative measures
4	“Ideal conditions” / Scientific evaluation
5	Describe intervention context to inform implementation strategies
6	Evaluate Effectiveness of preventive measures in implementation context

Figure 2-5 The Translating Research into Injury Prevention Practice framework (Finch, 2006)

2.4.2 Mechanical Aetiology of ACL Injury

The second stage in the model of injury prevention is to establish the cause or mechanism of the injury and to identify the risk factors for injury (Van Mechelen, 1992). Understanding the mechanical aetiology of sport injuries is critical to identifying and prescribing appropriate interventions for prevention and rehabilitation (Van Mechelen, 1992). Various types of research, such as experimental laboratory, in vivo/cadaveric, and in-silico, have improved our understanding of the loading patterns, joint kinematics and phases of movement that increase the risk of ACL injury. This information can be used to develop effective countermeasures. It has been found that valgus, internal rotation knee moments and anterior tibial translations relative to the femur increased ACL strain in cadaveric knee models (Markolf et al., 1995; Shin et al., 2011). In addition, simulation studies (in-silico) suggest that the anterior draw on its own is not likely to be the main mechanism involved in ACL injury, as valgus knee moments are also needed to reach injury inducing loads (McLean et al., 2004; McLean et al., 2008). Hewett et al. (2005) found that valgus knee moments can predict the ACL injury status of adolescent females, at rates of 73% specificity and 78% sensitivity. In a single healthy male, peak in-vivo ACL strain was shown to occur mostly during the weight acceptance (WA) phase of their stance (first 20% – 30%) during a COD manoeuvre, when the risk of ACL injury is likely to be greatest (Besier et al., 2001b; Cochrane et al., 2010; Dempsey et al., 2007). Therefore, it follows that the WA phase of COD manoeuvres is when the ACL is at greatest risk of an injury. The knee valgus angle was found to be significantly greater in ACL injured versus uninjured adolescent female populations, and a predictor of ACL injury (R2 = 0.88) (Hewett et al., 2005). Moreover, it is widely acknowledged that dynamic valgus knee postures are associated with a higher risk of ACL injury (Hewett et al., 2005). Even so, the way in which athletes display these postures means it is likely to be a result of poor hip neuromuscular control during WA. However, such a relationship between hip and ACL injury risk is still unknown during 90° and 135° COD manoeuvres, and so the role of the muscle in supporting the hip and knee during COD manoeuvres should not be overlooked. Further research is needed to determine how hip kinematics and musculature influence knee loading during COD manoeuvres.

This provides the rationale for focusing on reductions to loads applied to the ACL, which can be done through two approaches: Firstly, by changing the individual's posture or technique to reduce the size of the loads applied to the knee during 90° and 135° COD manoeuvres. Secondly, by increasing the strength of the muscles to protect the knee when loads are elevated. To do so, it is essential to identify the link between hip kinematics and muscular and ACL risk factors (knee valgus moment and angle) to enable an improved assessment of how hip joint function supports the knee and mitigates ACL strain and injury risk while performing 90° and 135° COD manoeuvres. There is a high possibility that the hip joint is focused on during the kinematic chain to lower knee joint loading. Changing the athlete's technique during a 45° COD manoeuvre has been shown to be effective in reducing valgus knee moments by 36% in nine male team sport athletes (Dempsey et al., 2009). This can be achieved by recommending that athletes place their stance foot closer to the body's midline, keep their torso upright, and rotate toward the desired direction of travel (Dempsey et al., 2009). However, motor control strategies for reducing external knee loading during 90° and 135° COD manoeuvres have not been tested to date. It is known that greater knee loading is seen during 90° and 135° COD manoeuvres compare to 45° COD manoeuvres (Havens & Sigward, 2015a; Schreurs et al. 2017). Moreover, identifying direct, causative links between the athlete's kinematics and knee joint loading is complex when examining COD manoeuvres, as multi-segment, dynamic movements are involved. Therefore, there is limited causal information on the associations between hip kinematics and muscular during 90° and 135° COD manoeuvres concerning knee loading and increased risk of ACL injury (knee valgus moment and angle). Thus, additional research is required to establish these causal links in order to implement more focused and effective ACL injury prevention/rehabilitation training protocols during 90° and 135° COD manoeuvres.

In summary, to develop effective ACL injury prevention programmes, it is important to understand the mechanisms involved in injury, and the risk factors for that injury, prior to the development and implementation of interventions (Finch, 2006). According to injury surveillance literature, most sport-related ACL injuries happen during non-contact COD manoeuvres (Stage 1). The most likely mechanism causing non-contact ACL injuries (Stage 2) are the knee valgus angle and moments during the WA phase of COD manoeuvres when the knee is near full extension. The biomechanical factors linked to ACL injury risk that are addressed in countermeasures should concentrate on two main points: (1) Reducing the magnitude of externally applied valgus knee angle and moments, and (2) Increasing muscular support against the knee joint angle and moments, even though the extent of these still to be properly defined (Stage 3). To carry out stage 3, the role of hip kinematics in reducing the knee valgus angle and moment during 90° and 135° COD manoeuvres must be identified. Furthermore, the role of the hip muscular in reducing the knee valgus angle and moment during 90° and 135° COD manoeuvres must be known.

However, no study has identified how hip biomechanics would affect the biomechanical factors associated with ACL injury during 90° and 135° COD manoeuvres. Therefore, we now need to identify how hip kinematics could help in reducing the knee valgus angle and moment, and whether the hip muscle will increase muscular support during 90° and 135° COD manoeuvres. It should be noted that establishing such relationships is needed to ensure long-term effectiveness, and to evaluate the cost benefits of ACL injury prevention training protocols. Therefore, one aim of the research presented in this thesis is to find the relationship between hip biomechanics and ACL risk factors during 90° and 135° COD manoeuvres.

2.5 **Change of Direction Manoeuvre (COD)**

2.5.1 Change of Direction as a Risk Factor for ACL Injury

Certain types of sports, for example basketball, football and volleyball, involve a great deal of COD manoeuvres, which put athletes at risk of injury (Agel, Arendt, & Bershadsky, 2005; Orendurff et al., 2010; Prodromos et al., 2007). Clearly, the ability to change direction is an integral component of multidirectional sport (Bloomfield et al., 2007; Orendurff et al., 2010). During multidirectional sport games, players frequently have to change direction at various angles using different techniques (Bloomfield et al., 2007; Robinson et al., 2011). Notational analysis has been used with premier league football players, and this reveals that the players carried out 727 turns/swerves on average during a 90-minute game, and approximately eight turns/swerves were made every minute to the right or left (Bloomfield et al., 2007). Although CODs are commonly associated with non-contact ACL injuries in sport (Brophy et al., 2015; Walden et al., 2015), such manoeuvres are strongly linked to ACL tears (Boden et al., 2000; Cochrane et al., 2007). To perform a directional change, the athlete must first decelerate before redirecting the body in a new direction, and then accelerate away (Hase & Stein, 1999). Moreover, deceleration–acceleration movements combined with rapid changes in direction cause a larger knee valgus angle and moment, and lower knee flexion, which are potential risk factors for non-contact ACL ruptures (McLean et al., 2004; Olsen et al., 2004). This could result in greater loads on the knee joints and may be affected by other factors; for example, greater angles used in changing direction mean the athlete must reduce their horizontal velocity to zero, or almost zero, in order to move in a different direction. Throughout the match, high and low speeds were indicated during turns (Orendurff et al., 2010). Moreover, the turn angle plays an important role in running speeds, as athletes jog when turning in small angles, but slow their speed more, stop and then accelerate when turning in larger angles; this is more likely to put athletes at greater risk of ACL injury (Bloomfield et al., 2007; Havens & Sigward, 2015b). Thus, understanding the ideal techniques for COD and reductions to injury risk are of great concern to coaches and practitioners who work with multidirectional sports persons on initial prevention and on the return to participation.

As previously mentioned in section 2.3.6, a possible underlying mechanism for the ACL rupture is the lower-extremity motions in the sagittal, frontal and transverse planes (Boden et al., 2000). Even so, it seems that motions and forces in the frontal plane could predict an increased risk of noncontact ACL injury risk and an inciting mechanism compared with other planes (Hewett et al., 2005). These injuries have been attributed to poor technique or mechanics, as they often occur without contact from another player or object. Specifically, altered sagittal and frontal plane loading mechanics during COD manoeuvres are thought to place athletes at greater risk of ACL injuries (Markolf et al., 1995). Carrying out a COD manoeuvre requires an abducted, rather than an adducted, hip position, because greater hip abduction is needed to ensure a larger lateral foot plant distance in order to generate medial-lateral forces (Havens & Sigward, 2015b; Jones et al., 2015). However, this increases the risk of injury because of the potential to create a larger knee valgus moments as the force vector directs laterally to the centre of the knee joint (Havens & Sigward, 2015a; Jones et al., 2015). Change of direction technique (kinematics) and the resultant load (kinetics) appear to be dependent on the demands of the task (e.g., angle, speed and condition) (Dos'Santos, McBurnie, Thomas, Comfort, & Jones, 2019b). The following sections will review the COD technique in detail (section 2.5.2) and the effects of the angle (section 2.5.3), velocity (section 2.5.4) and anticipation (section 2.5.5) on change of direction biomechanics and how they can influence non-contact ACL injury.

2.5.2 Change of Direction Technique

Three different COD techniques have been primarily identified within the literature: the sidestep, crossover and split-step (Figure 2-6) (Dos'Santos et al., 2019b). Sidestep CODs are defined as a player planting their foot laterally opposite to the direction of travel (Figure 2-6) in order to create a push off impulse in a new direction (Dos'Santos et al., 2019b). The body is typically rotated towards the new direction of travel with the player accelerating to the opposite direction of the planted limb (Andrews, McLeod, Ward, & Howard, 1977). A crossover COD (Figure 2-6) involves positioning the plant foot on the same side of the new direction and then crossing the opposite limb for a new step in a different direction, whilst accelerating in the same direction as the push-off limb (Andrews et al., 1977). Finally, the split-step COD (Figure 2-6) involves the player performing a small jump, landing with both feet after which the contralateral limb is used to push-off in the new direction of travel (Dos'Santos et al., 2019b).

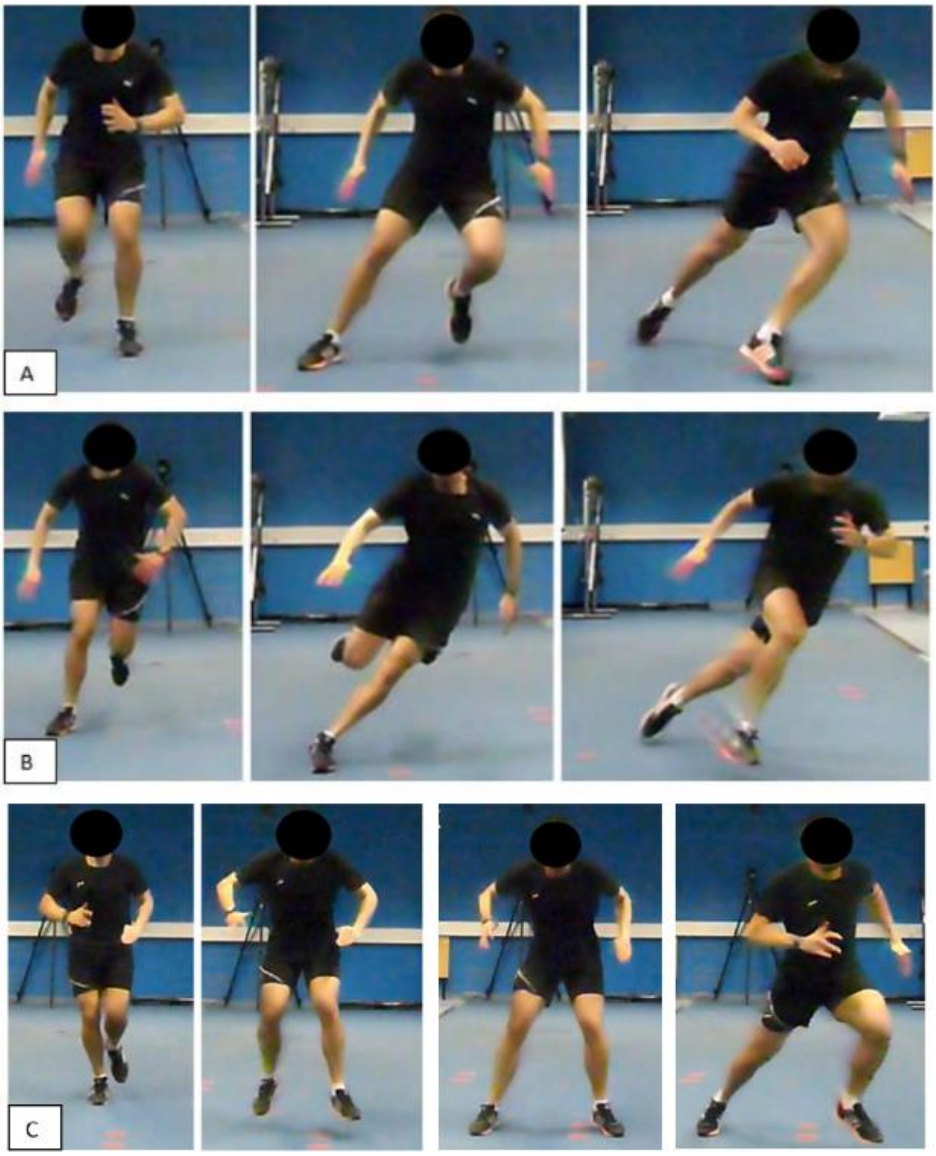


Figure 2-6 Three COD techniques: A) Sidestep, B) Crossover and C) Split-step (Dos'Santos et al., 2019b).

Researchers has compared the sidestep, crossover and split-step techniques from an injury risk point of view. Trewartha, Munro, and Steele (2007) found the split-step produced comparable lower knee joint loads to sidestepping. This could be explained by the bilateral symmetrical landing during split-steps, which distribute forces equally across both limbs, which contrasts with a unilateral landing during a sidestep. A side-step COD manoeuvre produces greater external knee abduction loading, knee flexion loading and internal rotation loading than crossover tasks (Besier et al., 2001b). Similarly, greater hip flexor, knee flexor, and knee valgus moments have been observed during sidesteps compared to crossovers (Kim et al., 2014). Consequently, the side-step COD technique seems to be a higher risk technique compared to the crossover and split-step technique; this is due to the kinetics and kinematics associated with the greater knee joint loading. Therefore, due to the higher risk of injury, the sidestep COD technique has been investigated in this thesis.

Few studies have investigated the difference in mechanics between COD manoeuvres and other manoeuvres. Jones et al. (2014) examined whether there is a link between single-legged landing and 90° COD manoeuvres in female athletes and found moderate correlations with knee valgus moments and significant correlations with peak knee valgus angles. Furthermore, Kristianslund and Krosshaug (2013) discovered moderate correlations for knee abduction angles between double-leg drop vertical jumps and 69° COD manoeuvres. The knee joint moments were higher in all three planes for the COD manoeuvre, and the knee abduction moments were six times higher for COD compared with drop jumps (Kristianslund & Krosshaug, 2013). The literature shows a poor correlation between constrained lateral cutting activities for SLL (stride land and cut; far-box land and cut; close-box land and cut) and COD manoeuvres (O'Connor, Monteiro, & Hoelker, 2009). COD is a high-energy situation requiring high speed for the change in direction, along with a single-legged stance; it therefore carries a greater risk compared to more controlled movements, for example double-leg drop jumping, landing, squatting or running, with various differences in the kinematics and kinetics involved.

2.5.3 Effect of Angle on Change of Direction Biomechanics

The magnitude of the load placed on the knee joint will be largely determined by the COD angle (Dos'Santos et al., 2019b). Many studies have examined lower-limb biomechanics during COD manoeuvres at various angles in order to gain an understanding of the associated injury risk factors. Moreover, studies have thoroughly examined planned COD manoeuvres of 45° (Havens & Sigward, 2015a, 2015b, 2015c; Schreurs et al., 2017), 90° (Havens & Sigward, 2015b, 2015c; Jones, Herrington, & Graham-Smith, 2016a; Jones et al., 2015; Schreurs et al., 2017) and 135° (Schreurs et al., 2017).

Lower hip and knee flexion angles, larger hip abduction angles and increased knee abduction moments have been found when undertaking COD manoeuvres at 90° rather than 45° in 25 healthy soccer players (comprising 13 males and 12 females) (Havens & Sigward, 2015b). In a further study involving 45 soccer athletes (25 males and 20 females)

Sigward et al. (2015) found that athletes displayed greater knee valgus moments, greater hip abduction angles and more ground reaction force when changing direction at a 110° angle compared with 45°. Generally, knee valgus moments were found to be 2.4 times greater during the 110° COD. The increased knee valgus moment may be due to the influence of the vertical ground reaction force, which was found to be 24.76 (N/kg) and 21.91 (N/kg) when comparing 110° COD with COD at 45° respectively. In addition, the increase in the knee valgus moment may be due to the more lateral placing of the vertical ground reaction force, which increases the moment arm and, as a result, could increase the knee valgus moment (Kristianslund et al., 2014).

It should be noted that the previous studies have analysed kinetic and kinematics variables at different time points. For example, Sigward et al. (2015) identified kinematic differences at the initial contact and found differences in the kinetic variables at the maximum value. Havens and Sigward (2015b) analysed kinematic variables from the initial contact to the maximum knee flexion angle and identified the maximum value for kinetic variables. It should be noted that previous studies investigating knee and hip biomechanics in dynamic movements have typically done so using discrete points (peak values). Analysing knee and hip kinematics during different phases would help to explain the different results between previous studies. Therefore, examining knee and hip joint ROM alongside peak angles during different time periods could provide a more complete analysis and additional understanding. These periods would include: (1) at the initial contact (IC); (2) from IC to peak vertical ground reaction force (PVGRF); (3) from IC to the first 60 millisecond of stance (60 ms); (4) from IC to peak external knee abduction moment (PEKAM), and (5) from IC to peak knee valgus angle (PKVA). In addition, involving the range of motion (ROM) in the knee abduction angle will help to further understand the altered knee valgus, which is at greater risk of ACL. Such an examination is warranted as the use of a discrete point analysis alone may not detect all significant asymmetries and relationships.

Schreurs et al. (2017) assessed knee kinematics and kinetics at various COD angles. The sample involved 29 healthy 18 to 27 year old team sport athletes (13 males and 16 females). The participants completed the trials in a laboratory setting over the course of one day. One trial involved a 5-metre sprint towards a force plate, followed by a COD manoeuvre on that force plate, and then a 5-metre sprint to the endpoint. Schreurs et al. (2017) examined five different conditions, namely running forward, and planned COD at 45°, 90°, 135° and 180°. The participants received verbal instructions and were asked to sprint at full speed from the start to the finish. An eight camera motion analysis system was used to carry out three-dimensional motion analysis, and two force plates were used to obtain ground reaction force data. In addition, two timing gates were used to calculate and record the task completion time. The main outcome variables of the study was: VGRF, the knee flexion angle, the knee valgus moment, and all kinematic and kinetic values of the right limb, which were assessed at the peak valgus moment. As far as the author is aware, this is the only study that has examined the differences between 90° and 135° COD manoeuvres (Schreurs et al., 2017). Nevertheless, Schreurs et al. (2017) found a reduction in the knee flexion angle with sharper CODs, which would be seen as problematic because extended knee positions increase anterior tibial shear forces (Yu, Lin, & Garrett, 2006), leading to increased ACL strain (Markolf et al., 1995). In addition, Schreurs et al. (2017) found that there were greater knee valgus moments in athletes demonstrating sharper CODs (90°, 135° and 180°) compared to 45° COD. Furthermore, in opposition to previous work Sigward et al. (2015) confirmed that VGRF magnitudes are significantly greater with sharper cuts, Schreurs et al. (2017) documented significantly greater VGRF in 45° cuts than the sharper CODs (at 90°, 135° and 180°). However, it should be noted that the task condition could be one reason for the conflicting result as Sigward et al. (2015) used unplanned COD manoeuvres while Schreurs et al. (2017) used planned COD manoeuvres.

However, it should be noted that Schreurs et al. (2017) has only investigated knee biomechanics; thus there is a lack of evidence on the differences between hip frontal, transverse and sagittal planes in COD manoeuvres. In addition, Schreurs et al. (2017) only investigated these differences at peak knee valgus moments. However, it is important to analyse kinematic and kinetic variables for hip and knee joints at multiple points throughout the entire stance phase of a COD manoeuvre, and not only one critical point. For example, this would include: the initial contact, at the peak VGRF, at the first 60 ms after initial contact, and at the peak knee valgus angle and moment. The authors knowledge that these differences have never been examined at 90° and 135° COD manoeuvres at all critical time points; such details would provide additional insight into the technique used for various directional changes and the associated risk of ACL injury.

Schreurs et al. (2017) found significant increases in the completion time for both male and female athletes as the COD angle increased (45°, 90°, 135° and 180°). The results are expected because, as the COD angle increases, a greater reduction in velocity (change in momentum) is required, which increases the demand for preliminary deceleration; furthermore, this usually happens over longer distances (Havens & Sigward, 2015c). A variety of running speeds have been examined during COD manoeuvres at different angles, namely 45°, 90° and 135°. The average speed for males was 4.7 m/s, 3.8 m/s, 3.5 m/s and 3.4 m/s, during 45°, 90°, 135° and 180° COD, respectively. For females these averages were 4.2 m/s, 3.6 m/s, 3.3 m/s and 3.2 m/s during 45°, 90°, 135° and 180° COD, respectively (Schreurs et al., 2017). However, it should be noted that it is essential to standardise the running speed of participants when comparing kinematics and kinetics, in order to ensure the same speed enables more accurate comparisons between individuals. Different speeds between COD manoeuvres may have affected the results as increased running speeds have been shown to cause changes to the kinematics and kinetics of the lower extremities (Nedergaard, Kersting, & Lake, 2014; Vanrenterghem, Venables, Pataky, & Robinson, 2012). Furthermore, the majority of researchers have confirmed that running speeds need to be standardised (Colby et al., 2000; Kadaba et al., 1989; Malinzak, Colby, Kirkendall, Yu, & Garrett, 2001; Pollard, Davis, & Hamill, 2004; Queen, Gross, & Liu, 2006). Therefore, speeds in the current study were controlled between participant and both manoeuvres. The participants used a completion time running speed of 4.2 m/s ± 0.5 for COD at 90° and 135° manoeuvres.

The magnitude of the load placed on the knee joint may be determined by the joint congruence and morphology. Joint congruence is the measurement of two opposing joint surfaces as they relate to one another considering the position and shape of each bone at their interface. For example, relating the femoral condyle to the tibial plateau. A mismatch between contacting surfaces may cause abnormal joint forces and stresses. Joint congruency has been suggested as a potential risk factor in ACL injury. Regarding ACL anatomical risk factors, there has been a focus on femoral and tibial bony morphology, especially the morphological and morphometric relationship between the lateral femoral condyle and the lateral tibial plateau (Vasta et al., 2018). A recent study involving 200 ACL-injured knees examined the impact of radiographic tibiofemoral lateral compartment bony morphology and morphometry on the risk of ACL injury (Vasta et al., 2018). The authors found that the measures that showed poor tibiofemoral congruity were linked to an increased risk of ACL injury (Vasta et al., 2018). Moreover, they also addressed the relationship between the anteroposterior distance of the lateral femoral condyle flat surface (XY) and the anteroposterior distance of the lateral tibial plateau (AB), which is the Porto ratio. Hence, the authors claim that an association between the two aforementioned osteoarticular surfaces potentially explains the divergent lateral pivot-shift outcomes for complete ACL ruptures. In other words, a disproportional Porto ratio may suggest greater severe knee instability. In addition, the morphometric ratio (XY/AB) according to the participants' radiographic measurements accurately identified individuals at a higher risk of ACL ruptures, utilising a multivariate logistic regression model, which revealed that the morphological parameters (XY and AB) are associated with ACL ruptures. Patients who had suffered an ACL rupture revealed much smaller lateral femoral condyle flat surface distances (XY) and smaller anteroposterior distances of the tibial plateaus (AB). Within this line, a flat, longer surface for the lateral femoral condyle, or a higher Porto ratio (XY/AB), is linked to a lower risk of ACL injury (Vasta et al., 2018).

While the quadriceps and hamstrings muscles believed to be essential to protecting the knee and provide stability during dynamic movements, specifically with regard to ACL injury prevention, Morgan, Donnelly, and Reinbolt (2014) also found that the gastrocnemius muscles play an important role in supporting the knee during single-leg landing tasks and therefore can potentially reduce ACL injury risk. So, measuring gastrocnemius strength is probably one other key risk factor. Thus, gastrocnemius muscles strength and coordination should be targeted in developing preventative ACL injury training protocols to reduce ACL injury risk.

In a recent systematic review assessing the impact of knee morphology on the likelihood of ACL injury this found that increased tibial slope along with poor tibiofemoral congruity, are linked to an increased risk of ACL injury (Bayer et al., 2020). A further systematic review by Andrade et al (2016) evaluating the morphological parameters of the bones of the lower extremities and the likelihood of suffering an ACL injury discovered that steeper posterior, medial and lateral tibial slopes are often associated with a higher risk of ACL injury (Andrade et al., 2016). However, other evidence regarding the tibial slope in ACL injured and non-injured populations had contrasting findings in respect to the effect of tibial plateau slope on injury risk (Wordeman, Quatman, Kaeding, & Hewett, 2012). In a case-control study involving 73 individuals with ACL-injured knees, it was found that an increase in the posterior tibial slope is related to the incidence of ACL injuries (Zeng et al., 2016). According to Sonnery-Cottet et al. (2011) and Hashemi et al. (2010), an increased posterior tibial slope presents a risk factor for ACL injuries. On the other hand, Chung, Chan, and Wong (2011) found no significant differences in the posterior tibial slope of ACL-deficient and ACL-intact knees. Furthermore, a study involving 76 ACL-injured individuals found a link between greater lateral posterior tibial slope and an increased likelihood of an ACL injury (Bojicic, Beaulieu, Imaizumi Krieger, Ashton-Miller, & Wojtys, 2017). Blanke et al. (2016) measured the lateral and medial tibial slopes of 121 non-contact ACL-injured knees, however they did not find any differences with non-injured participants. In addition, Van Diek, Wolf, Murawski, Van Eck, and Fu (2014) measured 45 ACL-injured knees and 43 healthy knees using MRI and also did not find differences in the lateral and medial tibial slopes when comparing the injured to the non-injured group.

From a biomechanical perspective, the larger the tibial slope, when there is a compressive load, the anterior shear component of the tibiofemoral reaction force generated will be higher, which causes an increase in the anterior motion of the tibia relative to the femur (Dejour & Bonnin, 1994). Due to the ACL forming the main restraint against such motion, it makes sense that an increase in the posterior tibial slope will cause a higher load on the ACL (Butler, Noyes, & Grood, 1980). McLean, Lucey, Rohrer, and Brandon (2010) have also suggested that axial compression of a knee with a higher lateral tibial plateau slope, compared with a medial tibial plateau slope, were significantly correlated with peak knee valgus angle and moment. This may cause greater anterior motion of the lateral compartment of the tibia compared with the medial compartment during a dynamic single leg landing, creating a net internal rotation of the tibia with respect to the femur, which may increase loading on the ACL. It may be also that greater relative lateral posterior tibial slope medially shifts the primary tibiofemoral contact area under impact-induced compressive loading. In this instance, the frontal plane moment arm of the ground reaction force will be larger, increasing the potential for larger external loads in this plane and subsequent abduction angles which may increase loading on the ACL.

In addition, greater knee valgus moments (Schreurs et al., 2017) and greater GRF magnitudes (Havens & Sigward, 2015c) occur with sharper CODs, although greater moment arms could occur as a result of greater hip abductions (Havens & Sigward, 2015b) which can be related to an increase in knee abduction moment (KAM), and can increase ACL strain (Markolf et al., 1995). However, the aforementioned findings present a challenge, because Hewett et al. (2005) found in their research with female adolescent athletes that KAMs can prospectively predict non-contact ACL injury. Therefore, sharper CODs place athletes at risk of greater knee joint loading and this increases the risk of injury. However, sharp CODs cannot be avoided when playing sport, and are necessary to evade opponents or pursue a ball. Therefore, athletes must have the physical capacity to cope with the subsequent knee joint loading associated with sharp directional changes and the ability to carry out these movements using optimal mechanics. Substantiating the findings of previous studies suggests that COD at larger angles are a far more risky manoeuvre and may need greater control in the hip frontal, sagittal, and transverse planes.

To reduce the knee joint load (knee valgus angle and moment), and therefore lessen the risk of ACL injury during COD manoeuvres, it is important to look at the correlation between dynamic knee-valgus variables and other lower extremity kinematics and kinetics. This is especially important for the knee and hip, and the biomechanics of the knee and hip joints during COD manoeuvres for 90° and 135° angles, which should thus be examined. The biomechanical demands of CODs are angle dependent and are critical factors that influence the technical execution of a COD and the knee joint loading. Therefore, practitioners and researchers should acknowledge and understand the implications of angle on COD biomechanics when interpreting biomechanical research. Therefore, one of the aims of this thesis is to compare the hip and knee biomechanical characteristics between COD manoeuvres performed at 90° and 135°.

2.5.4 Effect of Velocity on Change of Direction Biomechanics

It is useful to standardise the running speed of participants when comparing kinematics and kinetics, as ensuring the same speed enables a more accurate comparison between individuals, which is not affected by their speed. Increased running speeds have been shown to cause change in the kinematics and kinetics of the lower extremities (Nedergaard, Kersting, & Lake, 2014; Vanrenterghem, Venables, Pataky, & Robinson, 2012), and the majority of researchers have confirmed that running speed needs to be standardised (Colby et al., 2000; Kadaba et al., 1989; Malinzak, Colby, Kirkendall, Yu, & Garrett, 2001; Pollard, Davis, & Hamill, 2004; Queen, Gross, & Liu, 2006). The running speed prior to a manoeuvre clearly has an impact on the knee, because as the running speed increases, the stress on the knee increases. Knee valgus loads of 0.15 Nm/kg have been reported at running speeds of 3 m/s, while running speeds of 5 m/s have revealed knee valgus moment of 1.14 Nm/kg during COD 45 (Vanrenterghem et al., 2012). This result is corroborated by studies that have reported greater knee valgus moment with increased approach velocities (Kimura & Sakurai, 2013; Nedergaard et al., 2014). For loading in relation to deceleration, it may be expected that at higher speeds the load would increase, and the results of several studies confirm this, for example Landry, McKean, Hubley-Kozey, Stanish, and Deluzio (2007) found that peak valgus loading of approximately 0.30 Nm/kg ±1 at 3.5 m/s ±1, versus peak loading of up to 1.2 Nm/kg ±1 at 5.5–7.0 m/s ±1 had a major impact, and approach speeds of 5.5–7.0m/s ±1 present a risk due to inducing knee valgus loading, which could result in an ACL tear, or reduced performance. . In a second study, Sigward and Powers (2007) found high knee valgus loads of 1.2 Nm/kg ±1, yet on average, only approach speeds of 5.15 m/s ±1 were reached, and there was similar loading at higher speeds. This may be because the participants decided to aim for a balance between achieving the manoeuvre and mechanical loading due to awareness that higher shear forces will be generated with increased running speed. The speed-loading relationship is not a linear one, and questions remain concerning whether there is a minimum speed threshold for deceleration that causes knee loading, and at which level it is more likely to cause structural damage, as well as whether there is a maximum speed for COD manoeuvres.

A variety of running speeds during COD manoeuvres, with different angles of 45°, 90° and 135°, have been examined. For a COD angle of 45°, the speed ranges between 3.5-4.5 m/s (Dempsey et al., 2007; Schreurs, Benjaminse, & Lemmink, 2017). For a COD angle of 90°, the speed ranged between 3.8-4.7 m/s (Havens & Sigward, 2015b; Jones, Herrington, Munro, & Graham-Smith, 2014; Jones, Herrington, & Graham-Smith, 2015; Schreurs et al., 2017). For COD angles of 135°, the speed was found to be 3.5 m/s (Schreurs et al., 2017), and for 180°, speeds ranged between 3.4-3.9 m/s (Jones et al., 2014; Schreurs et al., 2017). It is essential to ensure sufficient loading as well as protecting the safety of participants; therefore, some kind of trade-off is necessary between achieving a manoeuvre and loading, which has led to a progression speed of 4.2 m/s ±0.5 being suggested as the optimum speed for investigating lower-limb loading mechanisms with COD manoeuvres of 90° and 135°.

2.5.5 Effect of Anticipation on Change of Direction Biomechanics

Research has been conducted to analyse the impact from a task anticipation status (planned vs. unplanned) on the mechanics of the lower extremity. Table 2.2 shows more detail about the effect of anticipation on biomechanical risk factors for ACL injuries during COD manoeuvres that consider the types of task and population. Conflicting results have been found when the mechanics of the lower extremities are examined in the sagittal, frontal and transverse planes (Brown, Brughelli, & Hume, 2014a). For example, in 37 male middle school soccer players, Kim et al. (2014) found a significant increase in the peak knee abduction angle and moment when tasks involved unplanned COD manoeuvres at 45° compared with planned COD at 45°. Also, significant increases in the peak knee abduction moment were discovered when tasks involved unplanned COD manoeuvres at 60° and 45° in comparison to planned COD at 60° and 45° (Besier, Lloyd, Ackland, & Cochrane, 2001a; Lee, Lloyd, Lay, Bourke, & Alderson, 2013). Weir, van Emmerik, Jewell, & Hamill (2019) found a significant increase in knee abduction moments from 23 to 36% of a stance when tasks involved unplanned COD manoeuvres at 45° compared with planned

COD at 45°. However, significant differences between planned and unplanned were not found at the time when peak knee abduction moments occurred (around 10% of the stance). The time, from 23 to 36% of stance, fell into the mid-stance phase; however, ACL injury usually occurs at an early stance phase in which the estimated time of ACL injury ranged from 17 to 60 milliseconds after the initial contact (IC) (Bates et al., 2020; Koga et al., 2010; Krosshaug et al., 2007b). In contrast, Cortes, Blount, Ringleb, and Onate (2011) found significant increases in peak knee abduction moments amongst 13 female soccer players when tasks involved planned COD manoeuvres at 45° compared with unplanned COD manoeuvres at 45°. However, four other studies, as shown in Table 2.2, have not found any significant differences in the peak knee abduction angle and the moment between planned and unplanned COD manoeuvres at 30°, 45° and 60° (Cochrane et al., 2010; Dempsey et al., 2009; Donnelly et al., 2012; Weinhandl et al., 2013). In addition, it was discovered that the COD performance under unplanned conditions led to greater hip (Kim et al., 2014; Weinhandl et al., 2013) and knee flexion (Besier et al., 2001a; Cortes et al., 2011; Dempsey et al., 2009; Donnelly et al., 2012; Kim et al., 2014), as well as ankle dorsiflexion (Weinhandl et al., 2013), when compared with planned conditions. However, in contrast, studies have also found no impact on knee sagittal plane kinematics (Cochrane et al., 2010; Lee, Lloyd, Lay, Bourke, & Alderson, 2013; Weinhandl et al., 2013). Furthermore, the impact on the mechanics of the knee in the transverse plane in unplanned tasks have been analysed. Kim et al. (2014) found a statistically significant increase in peak knee internal rotation angles during unplanned COD manoeuvres. In contrast, some researchers claim that anticipation did not affect the transverse plane kinematics of the knee (Cochrane et al., 2010; Dempsey et al., 2009; Donnelly et al., 2012; Weinhandl et al., 2013). It should be noted that the aforementioned studies only investigated the anticipation effect during small COD angles at 30°, 45° and 60°. However, the effect of anticipation on biomechanical risk factors for ACL injuries during 90° and 135° COD manoeuvres have not been tested to date.

Table 2-2 The effect of anticipation on biomechanical risk factors for ACL injuries during COD

Study and participants	Activity	Effect of Anticipation	Magnitude of differences	Statistical value
Kim et al., (2014)	45° COD manoeuvre.	**Kinematic findings**		
		Increased peak knee flexion angles (°)	ANT 45.7 ± 7.5 vs UNA 57.8 ± 7.6	$P < 0.001$
37 male	Biomechanical variables	Increased knee varus angles (°)	ANT 0.7 ± 6.8 vs UNA -0.7 ± 9.6	$P = 0.011$
Adolescent soccer	examined over 100% of	Increased peak knee internal rotation (°)	ANT 10.9 ± 11.1 vs UNA 13.6 ± 10.6	$P = 0.011$
players	stance phase.	Increased peak hip flexion (°)	ANT 39.7 ± 8.0 vs UNA 48.4 ± 7.8	$P < 0.001$
		Kinetic findings		
		Greater peak knee abductor moment (Nm.kg-1)	ANT 0.10 ± 1.00 vs UNA 1.44 ± 1.16	$P < 0.001$
		Approach speed	Not significant	
Cortes et al., (2011)	45° COD manoeuvre.	**Kinematic findings**		
		Increased peak knee flexion angles (°)	ANT: 45.2 ± 4.5 vs UNA: 52.4 ± 5.6	$P < 0.001$
13 female	Biomechanical variables	Increased knee valgus angles (°)	ANT: -4.0 ± 5.3 vs UNA: -7.2 ± 5.3	$P < 0.001$
Soccer players	examined during the first 50	Knee transverse plane	Not significant	
	% of stance.	**Kinetic findings**		
		Smaller peak knee abductor moment (Nm.kg-1)	ANT: 0.52 ± 0.40 vs UNA: 0.37 ± 0.36	$P = 0.035$
		Approach speed (ms-1)	ANT: 4.4 vs UNA: 3.7	$p < 0.001$
Demspey et al., (2009)	45° COD manoeuvre.	**Kinematic findings**		
		Increased mean knee flexion (°)	ANT: 29.7 ± 4.8 vs UNA: 32.1 ± 2.8	$P = 0.038$
9 male	Biomechanical variables	**Kinetic findings**		
Nonelite team sport	examined during weight	knee abductor moment (Nm.kg-1)	Not significant	
	acceptance phase.	**Approach speed (**ms-1)	ANT: 5.7 vs UNA: 5.1	$p < 0.05$
Weinhandl et al., (2013)	45° COD manoeuvre.	**Kinematic findings**		
		Increased hip flexion (°)	ANT: 36.3 ± 9.0 vs UNA: 38.6 ± 8.0	$P = 0.006$
20 female	Biomechanical variables	Increased ankle dorsiflexion (°)	ANT: 3.5 ± 4.4 vs UNA: 5.3 ± 3.0	$P = 0.015$
Recreational athletes	reported at peak ACL	Knee sagittal plane	Not significant	
	loading.	Knee transverse plane	Not significant	
		Kinetic findings		
		Knee abductor moment (Nm.kg-1)	Not significant	
		Approach speed (ms-1)	Not significant	

Besier et al., (2001) 11 male	30° and 60° COD manoeuvre Biomechanical variables reported during weight acceptance phase.	**Kinematic findings** Increased knee flexion (°) Knee transverse and frontal plane not investigated **Kinetic findings** Greater knee abductor moment (Nm.kg-1) **Approach speed (**ms-1)	 30: ANT 31.9 vs UNA 35.2 60: ANT 32.3 vs UNA 34.3 30: Not significant 60: ANT 0.02 vs UNA 0.3 30: ANT 2.7 vs UNA 2.5 60: ANT 2.4 vs UNA 2.2	 P < 0.001 P = 0.005 p < 0.05 p < 0.05 p < 0.05
Cochrane et al., (2010) 50 male Australian rules footballers	30° and 60° COD manoeuvre Biomechanical variables reported during weight acceptance phase.	**Kinematic findings** Knee frontal plane Knee transverse plane Knee sagittal plane **Kinetic findings** Knee abductor moment (Nm.kg-1) **Approach speed (**ms-1)	 Not significant Not significant Not significant Not significant Not significant	
Donnelly et al., (2012) 34 male Australian rules footballers	45° COD manoeuvre. Biomechanical variables examined during weight acceptance phase.	**Kinematic findings** Increased knee flexion (°) Knee frontal plane Knee transverse plane **Kinetic findings** Knee abductor moment (Nm.kg-1) **Approach speed (**ms-1)	 ANT: 33.0 ± 6.2 vs UNA: 35.3 ± 6.4 Not significant Not significant Not significant Not significant	 P < 0.01
Lee et al., (2013) 15 high level male and 15 low level male soccer players	45° COD manoeuvre. Biomechanical variables examined during weight acceptance phase.	**Kinematic findings** Decreased hip flexion (°) Knee sagittal, transverse and frontal plane not investigated **Kinetic Variables** Greater peak knee abductor moment (Nm.kg-1) **Approach speed (**ms-1)	 ANT 47.5 ± 7.3 vs UNA 41.4 ± 6.6 ANT 0.39 ± 0.27 vs UNA 0.66 ± 0.30 Not significant	 P < 0.001 P < 0.01
Weir et al., (2019) 22 male Collegiate team sport athletes	45° COD manoeuvre. Biomechanical variables examined over 100% of stance phase.	**Kinematic findings** Knee sagittal, transverse and frontal plane not investigated **Kinetic Variables** Knee abductor moment (Nm.kg-1) **Approach speed (**ms-1)	 ANT 0.05 vs UNA 0.15 from 23 to 36% of stance Not significant	 P= 0.001

There are some possible reasons for the contrasting results across the investigations. For example, the conflicting results for hip and knee kinematics and kinetic variables could be attributed to the different COD assessment methods used. Whilst most studies used a fairly straightforward stimulus, such as alternating colours or directional arrows, to instruct and direct movement (Besier et al., 2001a; Weinhandl et al., 2013; Weir et al., 2019) others incorporated different stimuli by using an “opponent” video to indicate the requisite movement direction, which may more closely mimic participation in sport (Cortes et al., 2011; Lee et al., 2013). Lee et al. (2013) used a traditional arrow stimulus over a stimulus that required participants to respond to a video of a soccer defender. They report that both significantly influenced knee mechanics, but the soccer simulation video had a greater impact. It is also notable that the simple unplanned condition used for COD tests, such as responding to a light system or directional arrows, is not an ecologically valid stimulus, and such generic stimulus types do not sufficiently resemble the situations faced by athletes during sport to tell elite and sub-elite performers apart (Sheppard & Young, 2006). This is because more elite athletes can use their extensive game knowledge to quickly anticipate situations according to phase of play sequences, before reacting to the kinematic cues shown among their opponents (Abernethy & Russell, 1987; Farrow, Chivers, Hardingham, & Sachse, 1998). Hence, tests involving generic cues, such as light stimulus or directional arrows, are likely to have limited potential for assessing transferrable sport-specific abilities among athletes, as these abilities involve perception-action coupling, along with decision making, which enable the accomplished performance of a COD task. Therefore, not using realistic scenarios is likely to limit our understanding of the effects of real-life unplanned action and the risk of an ACL injury. Although the research so far has been useful and informative, the methods used to assess the impact from anticipation have been relatively controlled compared with the realistic demands of a sporting environment. This is because most studies have only included two or three choices. However, the temporal control of the video directional stimuli of opponents that required participants to react with a directional change is questionable as milliseconds may alter ACL injury risk factors (Stephenson, Zhu, & Dai, 2016). In parallel to the stimuli presented, researchers also utilised different time delays between the presentation of the directional stimuli and the impact with the ground. Some investigations did not specify a delay (Kim et al., 2014), whilst others adjusted the delay per participant (Besier et al., 2001a), and a selection of literature specified a temporal delay ranging from 350 to 850 milliseconds (Cortes et al., 2011; Lee et al., 2013; Weinhandl et al., 2013). It is possible that temporal delays may allow the athlete to completely implement a new motor plan in response to the directional stimuli. Notably, no study that specified a precise time delay provided methodological details, validation, or indications of variability in these time points.

The conflicting results in the hip and knee kinematic and kinetic variables between planned and unplanned COD manoeuvres may be attributed to the fact that some studies did not control for similar approach speeds between planned and unplanned COD manoeuvres, as shown in Table 2.2 (Besier et al., 2001a; Cortes et al., 2011). For example, in the study by Besier et al. (2001a), the approach speed was significantly different between the planned and unplanned COD manoeuvres, as the unplanned manoeuvres were performed ~0.15 ms-1 slower than the planned COD manoeuvres ($p < 0.05$). Furthermore, in the study by Cortes et al., (2011), the participants had an approach speed of 3.7 ± 0.2 m/s for unplanned COD manoeuvres, and 4.4 ± 0.5 m/s for planned COD manoeuvres. Thus, there was a significant difference between the conditions for approaching speeds ($p < 0.001$). It is useful to standardise the running speed of participants when comparing the kinematics and kinetics between planned and unplanned COD manoeuvres, as ensuring the same speed enables more valid comparisons between individuals, which are not affected by speed. Increased running speeds have been shown to cause change to the kinematics and kinetics of the lower extremities (Vanrenterghem et al., 2012), and thus running speeds need to be standardised and controlled. Furthermore, the approach speeds in these studies were slower than the recommended approach speed (4 m/s) for meaningful knee loading, and the minimisation of task failure (Vanrenterghem et al., 2012), which were 3.0 m/s (Besier et al., 2001a) and 3.5 m/s (Kim et al., 2014).

Conflicting results may also be contributed to the analysis of variables at different pre-selected time points, such as initial contact, during the weight acceptance phase (Besier et al., 2001a; Dempsey et al., 2009; Donnelly et al., 2012; Lee et al., 2013), at peak ACL loading (Weinhandl et al., 2013), during the first 50% of stance (Cortes et al., 2011) and during 100% stance (Kim et al., 2014). Analysing the kinematic and kinetic variables during different phases, would explain the difference between the results from previous studies. In addition, analysing the kinematic and kinetic variables at the first 50% of stance (Cortes et al., 2011) or during 100% of stance (Kim et al., 2014; Weir et al., 2019) may not be relevant to the point at which the ACL injury may occur, as the estimated time of injury ranged between 17 and 60 milliseconds after the initial contact (IC) (Bates et al., 2020; Koga et al., 2010; Krosshaug et al., 2007b). Therefore, in this study kinematic and kinetic variables were investigated during the following time periods: (1) at initial contact (IC), (2) from IC to the peak vertical ground reaction force (PVGRF), (3) from IC to the first 60 millisecond of stance (60 ms), (4) from IC to peak external knee abduction moment (PEKAM), and (5) from IC to the peak knee valgus angle (PKVA). These time periods are essential as they could offer more comprehensive opportunities for analysis and an additional understanding that may help to further comprehend the true alteration to the knee valgus which is at greater risk of ACL.

In addition, these investigations utilised different samples; some relied on males (Besier et al., 2001a; Kim et al., 2014; Lee et al., 2013) whilst others on females (Cortes et al., 2011; Weinhandl et al., 2013). They also utilised different levels of experience, ranging from healthy active (Weinhandl et al., 2013), recreational (Besier et al., 2001a), National Collegiate Athletic Association (NCAA) athletes (Cortes et al., 2011) and collegiate team sport athletes (Weir et al., 2019). Across the spectrum of expertise, some investigations controlled for sport specialisation (basketball, soccer, and/or volleyball, with soccer the most common). No investigation explored the effects of sport specialisation on anticipation effects. It could be concluded that our understanding of the effect of a task's anticipation status (planned vs. unplanned) on the mechanics of the lower extremity is limited by contrasting results from the studies (Table 2.2).

2.6 Hip motion and loading

Several studies have investigated the relationship between hip kinematics and strengths and ACL injury risk factors, dynamic knee valgus moments and angles. Associations have been found between peak knee valgus moments and angles, and initial hip internal rotation, hip abduction and hip flexion angles (Havens & Sigward, 2015a; McLean, Huang, & van den Bogert, 2005; Sigward & Powers, 2007) throughout COD manoeuvres at 45° and 90° angles. However, Imwalle et al. (2009)) reported hip adduction angle is a significant predictor of knee valgus angle during jump with a double leg landing followed by COD manoeuvres at 45° and 90° angles. In addition, a number of researchers have discovered a link between reduced hip muscle strength and greater knee valgus angles (Gehring, Melnyk, & Gollhofer, 2009; Hollman et al., 2009; Jacobs, Uhl, Mattacola, Shapiro, & Rayens, 2007; Willson, Ireland, & Davis, 2006), as well as knee valgus moments (Lawrence, Kernozek, Miller, Torry, & Reuteman, 2008). Isometric abduction and external hip rotation strength has been reported to independently predict ACL injuries (Khayambashi, Ghoddosi, Straub, & Powers, 2016), indicating that weakness in hip abductor and external rotator muscles is a modifiable risk factor. These findings are concerning because greater knee valgus angles are linked to increase knee valgus moments (Jones et al., 2015; Kristianslund, Faul, Bahr, Myklebust, & Krosshaug, 2014; Sigward, Cesar, & Havens, 2015), which can increase ACL strain (Markolf et al., 1995; Markolf et al., 1990). In addition, greater knee valgus angles have mechanisms and characteristics associated with ACL injuries (Koga et al., 2010; Montgomery et al., 2018; Walden et al., 2015). The proximal control of the femur is an area of interest, because excessive hip motion while performing closed chain activities, for example landing and COD, may affect the frontal plane knee biomechanics (Powers, 2010). Control of hip frontal and transvers plane motion are typically viewed as being important as they are linked to greater knee valgus angles throughout movement (Dempsey et al., 2007; Frank et al., 2013; Havens & Sigward, 2015a; Hewett et al., 2005; Imwalle et al., 2009; Jones et al., 2015; Kristianslund et al., 2014; McLean et al., 2005; Powers, 2003; Sigward et al., 2015; Sigward & Powers, 2007). Clinically, the most regularly targeted areas are the two hip muscles, and to improve the control of the thigh, the gluteus maximus and medius are considered. The gluteus medius is the primary hip abductor and the gluteus maximus is the primary hip extensor and external rotator.

Some research has found internal hip rotation and dynamic knee valgus to be related (Havens & Sigward, 2015a; McLean et al., 2005; Powers, 2010). It is likely that internal hip rotation has an impact on dynamic knee valgus, because the rotation of the hip could place strain on the ligaments working to stabilise the knee while in the frontal plane position; however, while it has been associated with knee abduction, internal hip rotation is an independent predictor of ACL injury (Paterno et al., 2010). Internal hip rotation can result in frontal plane knee loading injuries. McLean et al. (2005) carried out a study with twenty basketball players from the national collegiate athletic association division one, and requested them to perform COD to 45°. They found that initial hip internal rotation is an important predictor of knee valgus angle and should be taken into consideration to minimise ACL injuries. Sigward and Powers (2007) also discovered that the initial hip internal rotation angle is related to peak knee valgus angle in their study carried out with 61 female football players involving a 45° COD manoeuvre. Havens and Sigward (2015a) also claim that peak knee valgus moment is directly linked to internal hip rotation positions according to their research with twenty-five male and female football players who were asked to perform 45° and 90° COD manoeuvres. However, (Jones et al., 2015) found that for 90° COD, internal hip rotation angles did not correlate with peak knee valgus moments in their study conducted with twenty-six elite and sub-elite female football players. The external rotator muscles are involved in hip extensor functioning, and the movement of the external and internal rotation of a joint takes place in the transverse plane. It is important to consider rotation as a factor in ACL injuries due to the position of no return, which occurs as the hip internally rotates (Ireland, 1999). Landing with the hip in internal rotation and adduction is one of the risk factors for knee valgus (Hewett et al., 2005), and McLean et al. (2005) explain that to mitigate this risk, it is advisable for the neuromuscular control of the lower extremities, particularly hip strength, is addressed in order to reduce the risk of ACL injuries. Leetun, Ireland, Willson, Ballantyne, and Davis (2004) claim that hip abductor and external rotator strength are essential to prevent injury, as they discovered that the strength of the hip external rotators can predict injury, and that injury-free athletes tend to have strong hip abductors and external rotators.

Research has been carried out to investigate the correlation between knee abduction angle and hip abduction throughout COD manoeuvres for different angles, and this has produced three different results. Sigward and Powers (2007) found initial hip abduction angle to be associated with peak knee valgus moment in their study carried out with 61 female football players doing a 45° COD manoeuvre. In addition, Kristianslund et al. (2014) examined 123 female handball players carrying out a COD manoeuvre (angles ranged between 60° and 80°), and they discovered that COD width (the angle between a line from centre of pressure to the centre of mass) is an important predictor of peak knee valgus moment. Frank et al. (2013) found hip adduction moments to be linked to increased knee valgus moments for 60° single-leg jump-cuts. In contrast, Imwalle et al. (2009) carried out a study with 19 female football players, and for 45° and 90° COD manoeuvres, noted that hip adduction angle is a significant predictor of knee valgus angle ($R = 0.49$). Finally, Havens and Sigward (2015a) carried out a study with 25 football players and found that initial hip abduction angle not to be significantly correlated with peak knee abduction moments during two COD manoeuvres (to 45° and 90°). Jones et al. (2015) analysed 26 elite and sub-elite female football players performing a COD to 90° and claim that hip abduction angles and peak knee valgus moments did not correlate significantly, but lateral leg plant distance did.

The link between hip muscle strength and knee valgus angle and moment must be considered, and this is controlled by two muscles- the gluteus maximus and gluteus medius. The gluteus maximus extends and rotates the hip externally, and the gluteus medius abducts and assists with internal rotation, creating force in the opposite direction to counter valgus collapse (Hollman et al., 2009). Several studies have examined the relationship between hip muscle strengths and ACL injury risk factors, Knee valgus angle and moment. Decreased hip muscle strength is associated with greater knee valgus angles (Gehring et al., 2009; Hollman et al., 2009; Jacobs et al., 2007; Willson et al., 2006) and greater knee valgus moments (Lawrence et al., 2008). All these studies have investigated the relationship between hip muscle strengths and ACL injury risk factors only during landing or single leg squat or double leg landing. However, almost 60-70 % of non-contact ACL injuries occurred while a player performed a COD manoeuvre (Johnston et al., 2018; Montgomery et al., 2018). Furthermore, COD have greater risk of ACL injury compared to landing from a drop jump (Chinnasee, Weir, Sasimontonkul, Alderson, & Donnelly, 2018). Knee valgus moment was 6 times higher in COD compared to the drop jump landing (1.58 ± 0.60 Nm/kg vs 0.25 ± 0.16 Nm/kg) (Kristianslund & Krosshaug, 2013). Therefore, COD manoeuvre would produce larger magnitudes of knee abduction motion which would be more appropriate to look for, especially when change direction to sharper angle 90° and 135°. However, there has been no published research correlating the hip abductor, extensor, and external rotator strength on frontal plane knee biomechanics during 90° and 135° COD manoeuvres. It is important to understanding the lower extremity biomechanics involved in COD manoeuvres to prevent injury as a result of the mixture of deceleration and change in direction required for COD, which has been shown to potentially lead to injury, especially ACL injuries. Thus, more investigation is important to understand what the role of hip frontal, sagittal and transvers planes mechanics on the mechanism of ACL injuries during COD to 90° and during 135° angles. Furthermore, hip abductors and extensors muscle strength are important point of focus and may decrease the risk of an ACL injury. Therefore, other aim for this thesis was to explore the relationships between ACL injury risk factors (dynamic knee valgus angle and moment) and hip biomechanics and muscles strength.

2.7 **Conclusion**

There are serious short term consequences which can occur as a result of an ACL injury, as it may take months to recover and engage in rehabilitation before a return to sport is possible. In addition, ACL injury greatly increases the likelihood of early-onset OA. Despite a great deal of research into the factors that lead to such injuries, the rate of injuries has not reduced. Increased frontal plane knee loading places the ACL under great strain, and past research has focused on finding ways to reduce the knee valgus angle and moment during COD manoeuvres. Thus, the proximal control of femoral motion in relation to hip abductors, extensors, and external rotators is currently an important area of research. Although players frequently change direction at degrees greater than 90°, limited knowledge is available on the kinematic and kinetic factors occurring at 90° and 135° COD angles. Moreover, the relationship between hip biomechanics and ACL risk factors during COD at 90° or 135° angles have not been discovered. No published studies have identified the hip kinematics and the relationship to ACL injury risk factors throughout 90° or 135° COD manoeuvres. Thus, it is essential to understand such relationships to mitigate injury risk and for the development of successful ACL injury prevention programs. Therefore, one aim of this thesis is to examine knee and hip kinematics and find their relationship to ACL injury risk factors during COD manoeuvres. In addition, it is essential to assess the lower limb biomechanics at various interval times, and not only consider the IC and peaks. For example, an investigation into the lower limb biomechanics regarding the risk of ACL injury over IC, the peak knee flexion angle, peak vertical GRF and peak knee abduction angle, peak external knee abduction moment and 60 milliseconds after initial contact, may provide additional understanding. Furthermore, to the author's knowledge, no published studies have investigated the relationship of knee and hip kinematics over multi-phases throughout COD manoeuvres at 90° and 135° angles. Therefore, the aim of this thesis is to examine knee and hip biomechanics and ACL risk factors over multi-phases during COD manoeuvres at 90° and 135° angles.

Moreover, all the previous studies examined one side and have not compared between two sides (preferred and non-preferred limbs). Therefore, more investigation is necessary to decide whether strength and biomechanical asymmetries exist between the preferred and non-preferred lower limbs during COD manoeuvres at 90° and 135° angles. However, no studies have examined the side-to-side differences in hip and knee kinematics and kinetics during COD manoeuvres at 90° and 135° angles. Therefore, other aim for this thesis was to determine whether asymmetry in knee and hip biomechanics kinematics and kinetics between preferred and non-preferred limbs during COD manoeuvres at 90° and 135° angles exists. The relationship between knee valgus angle and moment with hip muscle strength during COD manoeuvres still unknown. Currently, there has been no published research correlating the hip abductor, extensor, and external rotator strength on frontal plane hip and knee biomechanics during 90° and 135° COD manoeuvres, which could be expected to relate to each other. Therefore, other aim for this thesis was to explore the relationships between ACL injury risk factors (dynamic knee valgus angle and moment) and hip muscles strength.

In summary, the purpose of this thesis was to investigate the kinematic and kinetic characteristics of the hip and knee joints and find the relationship to the non-contact ACL injury risk factors over multi-phases during 90° and 135° COD manoeuvres. Other aim was to determine whether asymmetry in knee and hip biomechanics kinematics and kinetics between preferred and non-preferred limbs during COD manoeuvres at 90° and 135° angles exists. Last aim of this thesis was to explore the relationships between ACL injury risk factors (dynamic knee valgus angle and moment) during 90° and 135° COD manoeuvres and hip muscles strength.

2.8 Research questions:

The thesis answers the following questions:

(RQ1) What are the differences in the biomechanical variables between limbs during 90° and 135° COD manoeuvres?

(RQ2) What are the differences in the isometric hip abductors, extensors and external rotator strengths between limbs?

(RQ3) What are the differences in the biomechanical variables between 90° and 135° COD manoeuvres?

(RQ4) What is the relationship between hip and knee biomechanical variables during 90° and 135° COD manoeuvres?

(RQ5) What are the relationships between isometric hip abductors, extensors and external rotator strengths, and knee biomechanical variables during 90° and 135° COD manoeuvres?

2.9 Research null hypotheses:

1. The following null hypotheses will be assessed in the thesis:
2. There are no significant differences in the biomechanical variables between limbs during 90° and 135° COD manoeuvres.
3. There are no significant differences in the isometric hip abductors, extensors and external rotator strengths between limbs?
4. There are no significant differences in the biomechanical variables between 90° and 135° COD manoeuvres.
5. There are significant relationships between hip kinematics and knee valgus angles and moments during 90° and 135° COD manoeuvres.

6. There are significant relationships between isometric hip abductors, extensors and external rotator strengths and knee valgus angles and moments during 90° and 135° COD manoeuvres?

3 Chapter 3: Methodology

In this chapter, the general methodological approaches which have been used in this thesis will be discussed incorporating the biomechanical and strength assessment procedures. The reliability studies are then presented which ensures that before investigating the project's main goal, appropriate measurement procedures that give consistent and reproducible values with small measurement errors are understood.

3.1 Research Environment

The motion analysis work was completed in the motion analysis laboratories of the University of Salford in the United Kingdom and the Imam Abdulrahman Bin Faisal University in Saudi Arabia. At these laboratories, two different motion capture systems are in situ and thus the calibration of these systems is presented below. Calibration enables the system to collect kinematic and kinetic data. The accuracy of the calibration process during motion analysis plays a key role in determining the accuracy of the data, which means it should be administered carefully and according to the manufacturer's guidelines (Richards, 2008).

3.1.1 The University of Salford

A ten camera Qualisys Oqus 700+ motion analysis system sampling at 250Hz (Qualisys, Gothenburg, Sweden) operating through Qualisys Track Manager software (version 2.16) and three embedded force platforms sampling at 1000Hz (AMTI BP400600, USA) were used to collect the kinematic and kinetic lower limb data.

Two pieces of equipment are required to carry out the calibration process: a reference object, which is an L-shaped metal frame with four markers placed on it (Figure 3-1). This is placed in the corner parallel to the Y and X axis of force platform. The distance between the origin of force platform coordinate system and the markers is defined beforehand and automatically calculated before being passed over to the software (Winter, 2009). The frame enables a definition of the laboratory co-ordinate system, along with the X, Y, and Z axis, which form the medial/lateral, anterior/posterior, and vertical, respectively. The second piece of equipment is a wand, which is T-shaped and has two markers on it (Figure 3-2). The wand is randomly moved in all planes around the testing area with the L-shaped frame on the force platform in order to note the position and orientation of ten cameras in relation to the coordinate system (Payton, Bartlett, Sport, & Sciences, 2008).

The calibration process takes just one minute, and once completed, the calibration residual bar will indicate if the procedure was successful or not. For the calibration to be accepted all of the residual volume results should be below 1mm for each camera; if they were more than 1mm, the calibration was repeated until the successful calibration is achieved.

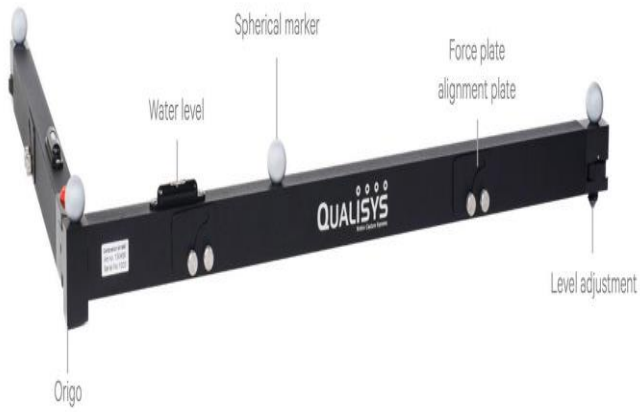


Figure 3-1 L-shaped calibration frame

Figure 3-2 T-shaped calibration wand

3.1.2 Imam Abdulrahman Bin Faisal University in Saudi Arabia

The 3D motion analysis laboratory at Imam Abdulrahman Bin Faisal University has a ten-camera Vicon motion analysis system (Vicon-Bonita cameras, UK) operating through Nexus software (Version 2.6.1) sampling at 250 Hz, along with two embedded force platforms (Kistler force plate Type 9286AA, Winterthur, Switzerland) sampling at 1000 Hz was used to collect the kinematic and kinetic data.

For the Vicon system calibration, one piece of equipment is required to carry out the process. This equipment is a wand, which is T-shaped frame with five LED markers placed on it (Figure 3-3). It is used for calibrating and setting the origin and orientation of the reference system. To start the calibration process, the wand is randomly moved around the testing area for one minute. Once completed, the T-shaped frame is placed on the force platform in order to note the position and orientation of ten cameras in relation to the coordinate system (Payton et al., 2008). It is put in a corner parallel to the Y and X axis of force platform. The distance between the origin of force platform coordinate system and the markers is defined beforehand and automatically calculated before being passed over to the software (Winter, 2009). The frame enables a definition of the laboratory co-ordinate system, along with the X, Y, and Z axis, which form the anterior/posterior, medial/lateral and vertical, respectively. In this thesis the X axis anterior is forward, the Y axis medially to the lift and Z axis vertical to upward.

Figure 3-3 T-shaped calibration wand for the Vicon system

3.1.3 Three-dimensional motion capture and markers placement

The marker placement was identical at both laboratories. At the start of the procedure, reflective markers (14 mm) were placed on anatomical landmarks of the lower limbs' of the participants using hypo-allergic adhesive tape. Forty reflective markers were used as static markers on the following sites:

- Two on the anterior superior iliac spines.
- Two on the posterior superior iliac spines.
- Two on the Iliac crest.
- Two on the greater trochanters.
- Four on the medial and lateral femoral condyles.
- Four on the medial and lateral malleoli.

- Two on the posterior calcanei (markers were placed on the standard training shoe).
- Six markers on the head of the first, second, and fifth metatarsals for both limbs (markers were placed on the standard training shoe).
- Four rigid plates, each consisting of four reflective markers, were attached to the antero-lateral aspect of the thigh and shank both limbs.

Following a satisfactory capture of all the static markers during the static trial, the anatomical markers were detached, keeping only 28 as tracking markers (16 markers over 4 cluster plates, 8 markers attached to standard shoes, and 4 markers on ASISs & PSISs). Both static and tracking markers are illustrated in figure 3-4 and 3-5 respectively.

Figure 3-4 Static Trial Markers

Figure 3-5 Tracking Markers

Markers were used to define the anatomical reference frame and the centre of joint rotation (Orishimo & Kremenic, 2006), and the movement of each segment was determined using the Calibrated Anatomical System Technique (CAST) (Cappozzo, Catani, Croce, & Leardini, 1995). During the trial, all reflective markers were in view of the cameras. The CAST technique has been used for dynamic manoeuvres to define the six degrees of freedom of movement for each segment of movement (Cappozzo et al., 1995); this has been proven to be reliable for biomechanical data collection (Benedetti, Catani, Leardini, Pignotti, & Giannini, 1998; Reinschmidt, Van den Bogert, Nigg, Lundberg, & Murphy, 1997). CAST has the advantage of offering improved anatomical relevance, compared with the modified Helen Hayes marker set. Furthermore, it attempts to reduce STA by attaching cluster markers (containing four markers) to the centre of segments rather than single markers on the joints, as in the Helen Hayes model (Cereatti, Camomilla, Vannozzi, & Cappozzo, 2007; Collins, Ghoussayni, Ewins, & Kent, 2009). The best practical solution was to have four markers per cluster attached to the skin where STAs are globally minimised. This is usually achieved if one avoids the placement of markers in correspondence with bone prominences where slipping effects are particularly evident. Each individual marker would have its own STA and thus a solid plate allows only the STA to be resolved, which would reduce the overall STA. Therefore, one would hope that the repeatability of the knee adduction moment would be improved due to the reduction in STA (Kadaba et al. 1989). In addition, the misplacement of retroreflective markers produce the greatest errors in 3D motion analysis (Ford et al., 2007; Malfait et al., 2014); this includes mistakes in identifying marker locations, which affects the calculations used to determine the positions of joint centres, resulting in errors in joint kinematic and kinetic calculations (Baker, 2006). One way to reduce these errors is to place markers on rigid plates fixed to the thigh and shank, as this has been demonstrated to result in less STA than when applied directly to the skin (Manal, McClay, Stanhope, Richards, & Galinat, 2000). Moreover, one examiner can attach all markers in all trials of the study, which can control this error. However, one limitation for using a cluster could be the restricted movement from the straps. In addition, they could slip; however, care was taken to ensure this did not happen as the cluster was attached to the thigh and leg using double-sided adhesive tape and secured with a strap.

3.1.4 Data processing

The raw marker trajectory data from the 3D motion analysis systems in both labs were reconstructed in Qualisys Track Manager Software (version 2.16) or Vicon Nexus software (Version 2.6.1). In these programmes, markers were labelled and then exported as a C3D file to Visual3D (Version 6.00.16, C-Motion Inc., Rockville, MD, USA) (Figure 3-6 and 3-7).

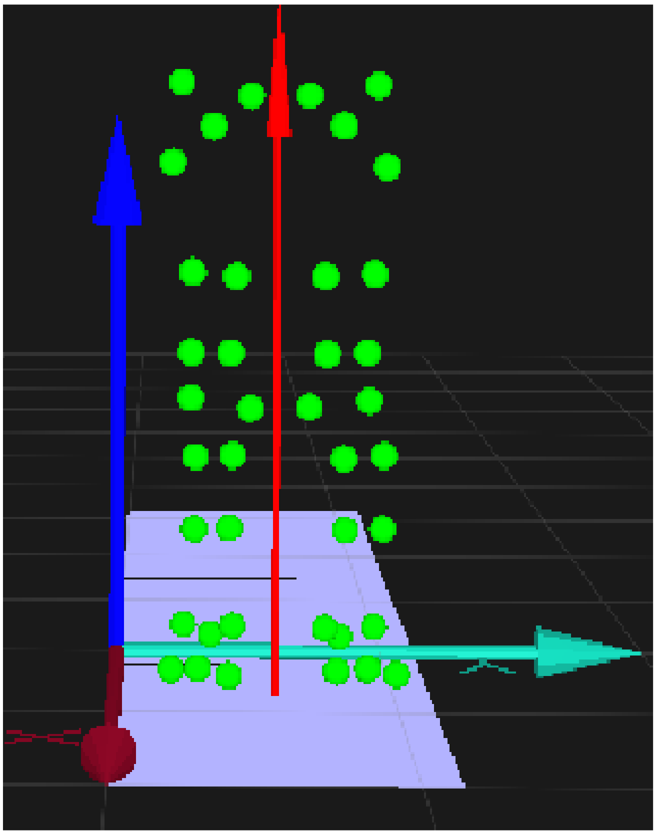


Figure 3-6 QTM™ static models

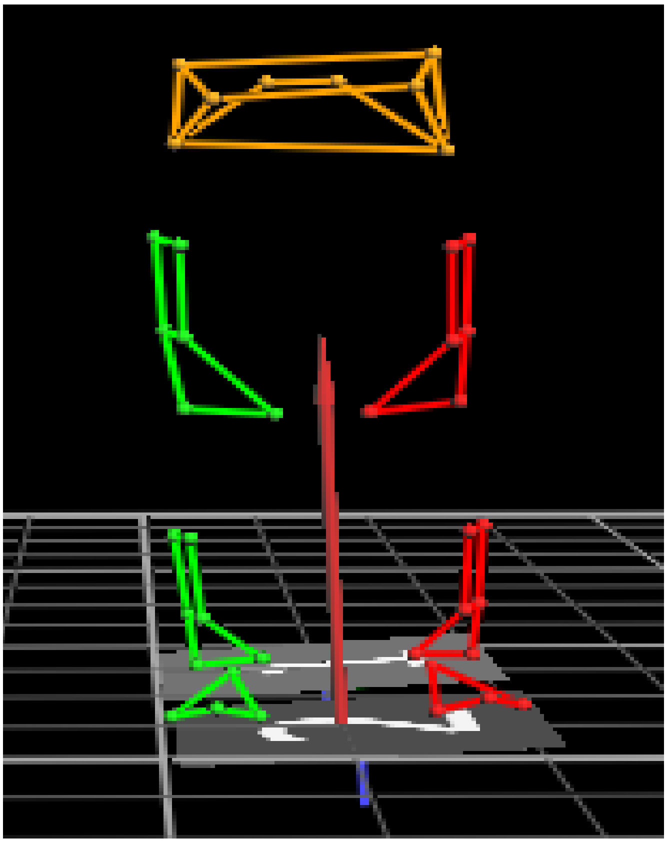


Figure 3-7 Vicon Nexus static models

This study used a model made up of seven rigid segments attached to the joint (Figure 3-8). Each segment has six variables for describing its position in 3D space: three variables for the position of the origin, and three variables for the rotation. Three of the variables describe the segment translation along three perpendicular axes (vertical, medial-lateral and anterior-posterior), and three variables describe the sagittal, frontal and transverse rotation around each axis of the segment. The details entered into the software for use in kinetic calculations were each participant's body mass in kilogrammes and their height in metres. Every segment of the pelvis, thigh, shank and foot were modelled to discover the proximal and distal joint/radius, and the hip-joint centre was calculated automatically using posterior and anterior superior iliac spine markers with the regression equation introduced by Bell, Oates, Clark, and Padua (2013) as descripted in table 3-1.

Table 3-1 Visual3d segments model.

Segment	Proximal markers location	Distal markers location	Tracking markers
Pelvic	Right and left anterior superior iliac spine markers	Right and left posterior superior iliac spine markers	4 markers in pelvic cluster built
Thigh	Hip joint center	Medial and lateral epicondyle markers	4 markers in thigh cluster
Shank	Medial and lateral epicondyle markers	Medial and lateral malleolus	4 markers in shank cluster
Foot	Medial and lateral malleolus markers	1st and 5th metatarsal head markers	Heel marker, 1st metatarsal head marker, 2nd metatarsal head marker and 5th metatarsal head marker

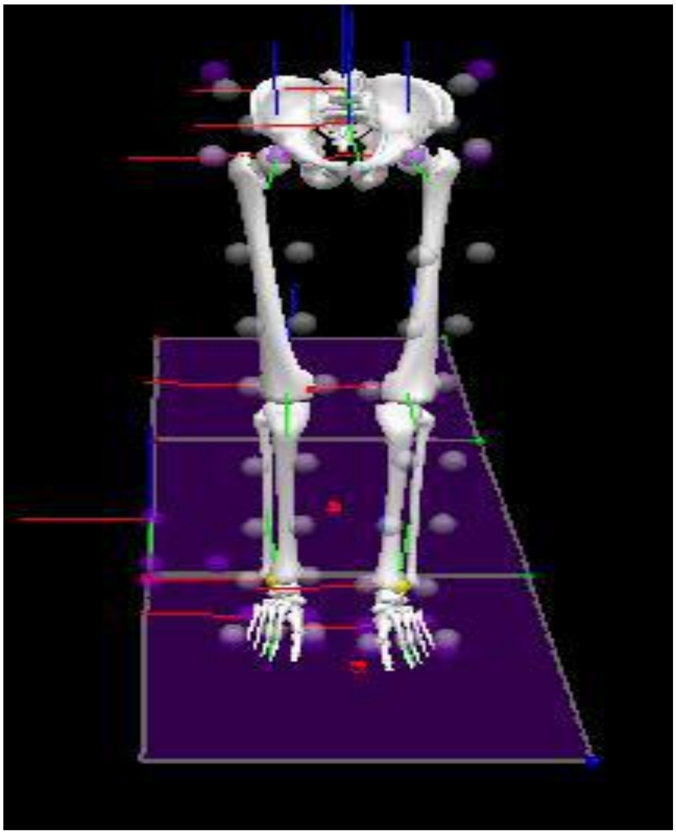


Figure 3-8 Visual 3D™ bone model

All kinematic data were interpolated with a maximum gap-fill of ten frames and filtered with a low-pass filter. Motion and force plate data were filtered using a Butterworth 4th order bi-directional low-pass filter with cut-off frequencies of 12 Hz for raw marker coordinate data and 25 Hz for data from the force platform. This was based on a priori residual analysis by Winter (2009), and a visual inspection of motion data and recommendations by Roewer, Ford, Myer, and Hewett (2014). Determining the appropriate low-pass marker cut-off value is achieved through residual analysis methods (Winter, 2005). For level walking and more dynamic tasks, such as landing and cutting, motion data are typically filtered using a cut-off frequency lower than 20 Hz (Ford, Myer, & Hewett, 2007; Ford, Myer, Toms, & Hewett, 2005; Hewett et al., 2005). Applying the same low cut-off frequency to GRF data will attenuate the signal and remove the true GRF impact peak that appears during rapid, high-impact movements like a landing or a sidestep cut (Roewer et al., 2014). This GRF impact peak contains large forces that are quickly transmitted from the ground through the foot, and must be absorbed, redirected and converted by the active (muscles) and passive (bones, ligaments and tendons) structures in the body (Collins, 1989). Even though these impact peaks may lead to joint moments that appear quite large, these are real forces that the body acts upon and one should be extremely cautious before filtering them out using a low GRF cut-off frequency to produce ‘smoother’ joint moment curves. However, Bezodis, Salo, and Trewartha (2011) suggest that the kinetic data should be filtered at the same cut-off frequency as kinematic data. This will likely yield the most realistic representation of the true resultant joint moments, as the peak resultant joint moments remain unaffected and artificial fluctuations are likely soon after contact by 'un-matched' higher frequency kinetic input data, which are avoided.

Kristianslund, Krosshaug, and van den Bogert (2012), indicated that significantly larger peak knee abduction moments were observed during a sidestep cut when different cut-off frequencies were applied to marker and GRF data (10 and 50 Hz, respectively) compared to when the same cut-off frequencies were applied (10 and 10 Hz, respectively). However, it should be noted that significant differences were observed when the marker and force data were filtered at large different cut-off frequencies (10-50 Hz). This may not stand true if the marker and force data were filtered at different small cut-off frequencies (12-25 Hz). Moreover, Roewer et al. (2014) investigated the effects of using the same low cut-off frequencies versus different cut-off frequencies to filter marker and GRF on joint moment magnitudes during a vertical drop jump. Marker and force data were low-pass filtered at the same low cut-off frequency (10, 12 and 15 Hz) and at different cut-off frequencies (10–50, 12–50 and 15–50 Hz, respectively). The analyses indicated differences between the peak knee abduction moment, which were computed when using marker and GRF data filtered with the same low cut-off frequencies and peak knee abduction moments that were computed using data filtered with different cut-off frequencies. However, athletes with the largest peak knee abduction moment values when the same cut-off frequencies were applied also had the largest peak when different cut-off frequencies were applied. In addition, in the current study, using the same or different cut-off frequencies will not affect the results. For example, when comparing between tasks or limbs, the same cut-off frequency will be applied to both tasks and both limbs; thus, all peak resultant joint moments will be affected the same which will not affect the final results.

Joint kinematics were calculated using an X-Y-Z Euler rotation sequence, where X equal flexion-extension, Y abduction-adduction/valgus-varus and Z internal-external rotation, as depicted in figure 3-9. Joint kinetic data were calculated using three-dimensional inverse dynamics, and the joint moment data were normalized to body mass and presented as external moments. The sign conventions which were used in this thesis are presented in the following table (Table 3-2).

Table 3-2 sign conventions used in current thesis

	Plane	Hip	Knee
Joint angles	Sagittal	Flexion +ve	Flexion +ve
	Frontal	Adduction +ve	Adduction +ve
	Transverse	Internal rotation +ve	
Joint moment	Frontal		Abduction +ve

+ve = positive

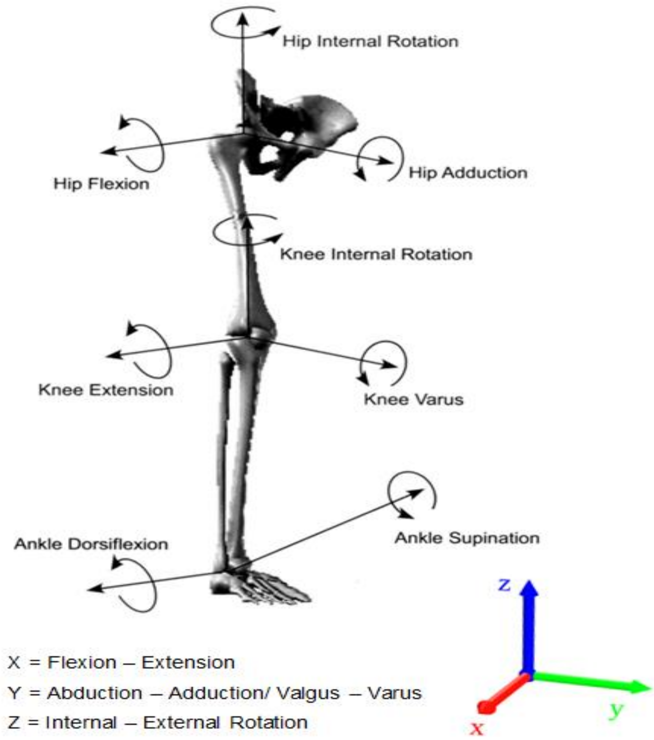


Figure 3-9 Lower extremity segment and joint rotation denotations (McLean et al., 2004)

Then, the defined starting and stopping motion events were created after the construction of a six degrees of freedom model. Firstly, initial contact (IC) and toe off (TO) were determined based on the force plate data. The IC was defined when the vertical ground reaction force (VGRF) first exceeded 10 Newtons and TO was defined when VGRF fell below 10 Newtons. The IC and TO events were created automatically based on the kinematic data of the first stance phase using automatic pipeline throughout using pattern recognition technique. The kinematic, kinetic and GRF data were investigated in the whole stance phase which is defined from initial contact to toe off. Kinematic and Kinetic data were time normalised to 100% of the right and left limb contact phase. The individual's maxima and minima of each trial was used to calculate the mean of the five trials to calculate each interested outcome in the kinematic, kinetic and GRF data. The primary outcomes during stance phase were calculated for the following variables: peaks of hip x angle, hip y angle, hip Z angle, knee x angle, knee y angle, knee y moment and VGRF. Biomechanical variables were exported as spreadsheets into Microsoft Excel to conduct the analysis and construct a graphical illustration.

3.1.5 Study Procedure

When the participants arrived at the laboratory, demographic characteristics (mass and height) and past medical history were collected. Participants were also asked if they have read and understood the information sheet and if they have any questions. They were also asked to sign a consent form. Body mass and height were measured using an electronic personal floor scale (MARSDEN, charder, model: M-420), and height measure (Seca, United Kingdom). Prior to testing, participants were asked to wear shorts and standard shoes. They were asked to perform five minutes of low intensity warm-up exercises using a cycle ergometer (Monark, Ergomedic 874 E), as advocated by Bell et al. (2013), to lessen the risk of physical discomfort and help to avoid any injuries during the tests (Woods, Bishop, & Jones, 2007). Three dimensional (3D), 3D ground reaction force (GRF) and Biodex System 4 dynamometer (Biodex Medical Systems, Inc, Shirley, New York, USA) were used to measure knee and hip joint kinematics and kinetics and isometric hip muscle strength. All data were collected for both the right and left limb during the performance of a pre-planned COD manoeuvres at 90º and 135 º to the opposite side of the contact (plant) limb and MVIC.

3.1.5.1 Change of direction manoeuvre (COD)

The participants were allowed to practice the manoeuvres until they were familiar and comfortable with it, which typically involved two to three attempts (Phillips & van Deursen, 2008). Following familiarisation, a total of 40 reflective markers were placed on the participants' lower limb using hypo-allergic adhesive tape (see section 3.1.3). After replacing the reflective markers, the calibration process was done (see section 3.1.1). Then, a static trial was collected as the participants stand with neutral alignment over the force plate with weight distributed equally over both lower limbs. Following a satisfactory capture of all the static markers, the anatomical markers were detached, keeping only 28 as tracking markers, details of the static and tracking markers are on section 3.1.3.

Participants were then requested to perform a minimum of five successful trials for COD manoeuvres at 90º and 135º with both limbs (Ross, Guskiewicz, & Yu, 2005); 30 seconds rest were provided between trials to reduce the effect of fatigue and two minutes between manoeuvres. Each participant was requested to run for 5 m, and then asked to contact the force platform using their foot (preferred or non-preferred), before immediately changing direction 90° and 135° to the opposite side of the contact limb and running 3 m in that direction. The course incorporated strategically placed cones to ensure that any angles were set at either an angle of 90° or 135°, also ensuring that all participants followed the same movement processes as the cones guide the participants to COD at an angle of 90° and 135°, as depicted in figure 3-10 and 3-11. A Brower Timing Gate System (TC-Timing System, USA) was placed at the start and end of the 8 m path to measure the manoeuvre's completion time and average running speed during the manoeuvre after each trial. These were set at approximately hip height for all participants, so that only one body part, for example the lower torso, breaks the beam (Yeadon, Kato, & Kerwin, 1999). In order to compare the findings with the literature and to control for performance differences between limbs and tasks, the participants used an average time to complete speeds during the task, at 4.2 m/s ± 0.5 for COD at 90° and 135° manoeuvres, as mentioned in Chapter 2, section 2.5.4. However, using a timing Gate System at the start and end of the 8 m path to measure the manoeuvre's completion time and average running speed may be not accurate as the subject could approach slowly and then exit fast or approach fast and then exit slowly. Another way to measure the approach speed accurately is to calculate the average linear velocity of a marker, the average COM velocity in the original direction of progression, the horizontal velocity in the direction of motion of the hip joint centre, and the timing gates in the original direction of the run. However, in order for this to be able to be calculated accurately, there would need to be a large enough capture volume pre- and post-movement. Unfortunately, the data collected were too limited in the capture volume (e.g., 1m before force platform) to calculate the approach speed from the COM velocity. This is a future direction of work, namely ensuring that there is a larger capture area approach speed and exit speed for measurement. Ultimately, in this study the subject needs to complete the task in a specific time and have been told to keep the same speed before and after the force plate. This is more in line with rehabilitation and training goals where an individual has to complete a task in a known time which, in this study, was consistent. However, the total time to complete the task was used to monitor and control the performance between trials and participants. It should be noted that one of the study limitations is that we could not accurately measure the approach speed and exit speed (addressed under the study limitations on page 185).

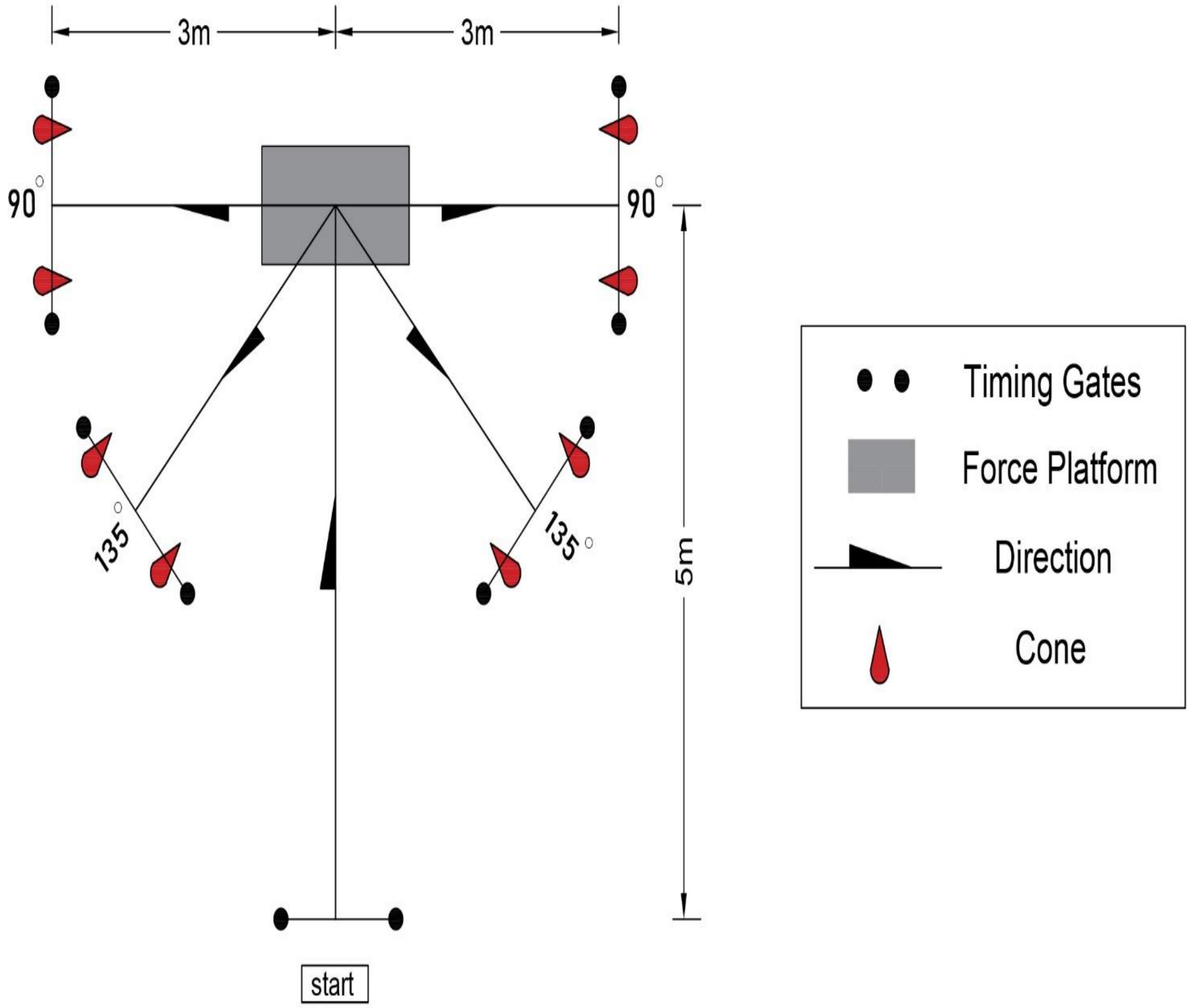


Figure 3-10 A plan view of the experimental set up at Imam Abdulrahman Bin Faisal University lab.

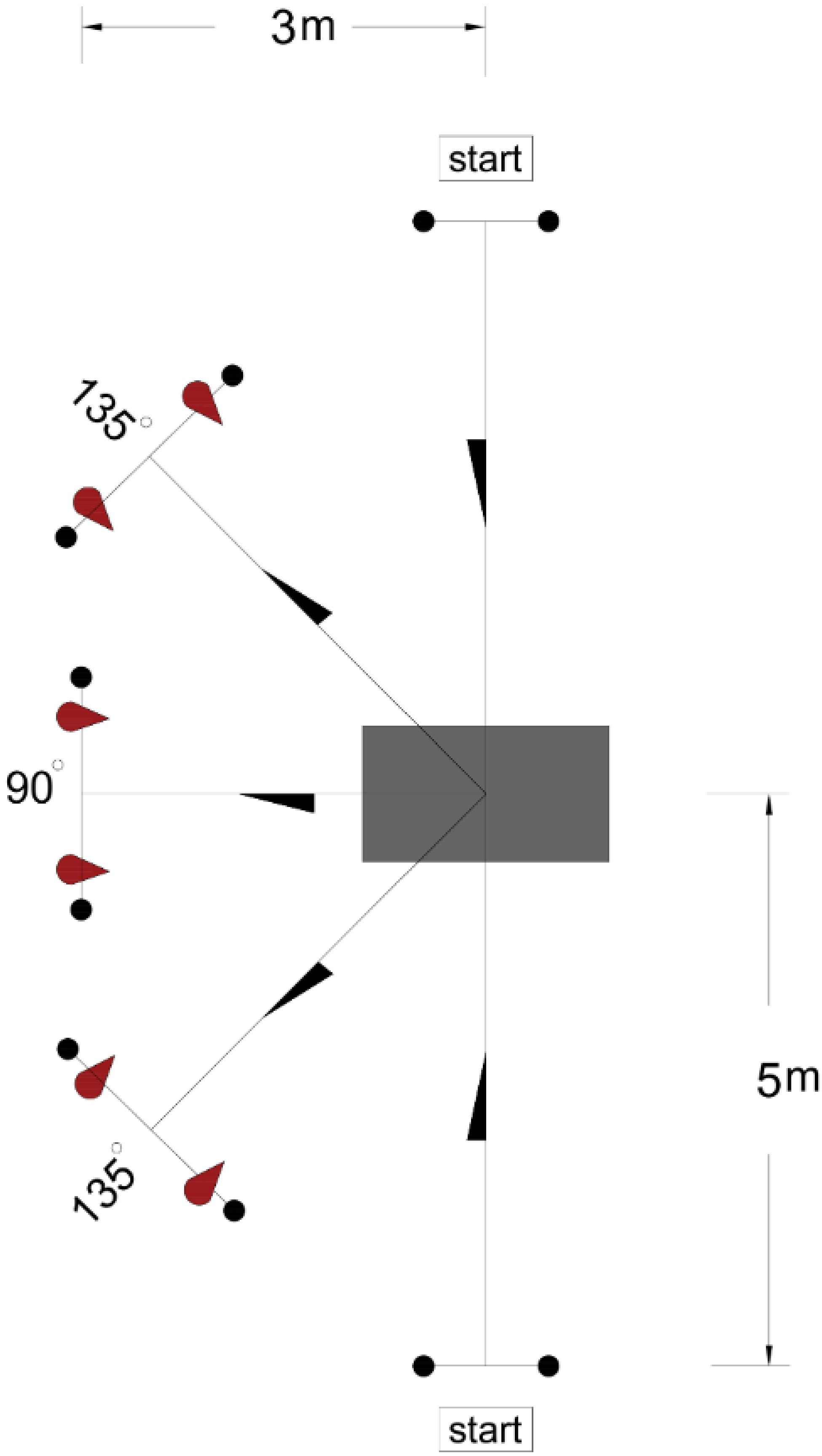


Figure 3-11 A plan view of the experimental set up at Human Performance Lab.

All data were collected for both the preferred and non-preferred limbs during the performance of a pre-planned COD manoeuvres at 90º and 135 º to the opposite side of the contact limb. Players may not typically use their dominant limb (the preferred limb to kick the ball with) as a push-off limb during COD manoeuvres; instead, they may use the other limb during COD manoeuvres rather than dominant limb. Therefore, in the current study the preferred limb was determined by two different methods. The participants were allowed to practice the COD manoeuvres until they were familiar and comfortable with it with let them to perform the task as they normally do it and that later show which limb they preferred to use during practice session. In addition, preferred limb was also determined by asking participants which limb they would prefer to use as the push-off limb during COD manoeuvre. Thus, the preferred limb was determined by ask and watch the participants. The data collection for preferred and non-preferred limb was collected in a random order using randomisation blocks via randomization.com. All participants were required to carry out minimum of five successful COD trials of preferred and non-preferred limbs to 90° and 135°. The trial was successful, if the contact phase of the movement on force plate, and all reflective markers were in view of the cameras; any unsuccessful trial was noted but was not included in the results (Phillips & van Deursen, 2008). After a satisfactory capture of all the requested trials, all the reflective markers were detached. Participants were then requested to perform five MVIC of the hip abductors, extensors and external rotators. 30 minutes rest were provided between COD manoeuvres and MVIC tests to reduce the effect of fatigue.

3.1.5.2 Isokinetic strength assessment

A Biodex System 4 dynamometer (Biodex Medical Systems, Inc, Shirley, New York, USA) was used to measure isometric hip muscle strength in both labs. The calibration of the Biodex dynamometer was applied according to the specifications outlined by the manufacturer’s service manual, before starting the tests. Studies emphasise the need for standardisation of testing protocols in order to target specific muscles and inhibit movements during strength measurements. Standardised tests positions across all participants allowed for more accurate comparison across participants. Generally, hip muscles strength should be measured in consistent positions across all participants.

Widler et al. (2009) carried out a comparison between various positions for isometric hip strength measures; they found side-lying to be the most valid and reliable method for measuring the hip abductor’s torque compared with the standing and supine positions. Similar finding has been confirmed by Meyer et al. (2013) who recommended the use of the side-lying body position whenever hip abductor strength is assessed. Isometric hip external rotator strength is most commonly measured at 90° of hip flexion in seated position. Prins and van der Wurff (2009) measured external rotator strength while seated, with the hip and knee at 90° angles. In addition, external rotation torque was not changed with increases in hip flexion (Johnson and Hoffman, 2010). Therefore, testing isometric hip external rotator strength at 90° of hip flexion would be valid.

For hip extension testing, participants were asked to lie prone on the testing table with their limbs off the end of the table and the non-tested limb on the ground while the tested limb was positioned at a 30° hip flexion while actively maintaining the knee at 90° flexion (Cronin, Johnson, Chang, Pollard, & Norcross, 2016; Worrell et al., 2002). This test aimed to assess the strength of the hip extensors, in a position mimicking that of the COD manoeuvres. The hip flexion angle ranged from 30° to 50° during COD manoeuvres. Havens and Sigward (2015a) found 46° and 32° of hip flexion at initial contact during a 45° and 90° COD manoeuvres respectively. Also, during COD manoeuvres at 45°, Sigward and Powers, 2007 discovered on average 46° in hip flexion. This shows that performing a COD manoeuvres needs a flexed hip position. Furthermore, half prone hip extension testing has been shown to be reliable in assessing hip extensors muscle strength, in which test-retest reliability ICC for this position was 0.92 (Lue et al., 2009). In addition, this position is a more stable measurement method, the hip joint is in a flexed position, subjects may be able to produce a more stable maximum force. The subjects stood in a comfortable and stable position without control of the joint angle.

The data collection for preferred and non-preferred limb was collected in a random order using randomisation blocks via randomization.com. The participants had the chance to practice every test two to three attempts until they were familiar and comfortable with it, which should involve two to three attempts. Also, to prevent fatigue, up to five minutes was the time given between different muscle group tests. Five maximal voluntary isometric contractions (MVIC) of the hip abductors, extensors, and external rotators were performed, and participants were requested to generate torque as hard as possible for five seconds with 30 second rest period between trials (Jacobs et al., 2007; Widler et al., 2009). A two minutes rest period was included between each of the three strength tests. Five completed MVIC trials were conducted for each muscle group. The torque-time curve from each MVIC trial was evaluated straight away to discover the initial countermovement and a plateau on the curve. Trials revealing a countermovement, and/or no plateau were discarded, and the trials were repeated until a maximum of five trials is reached. All data were collected for both the right and left limb. The procedure for the tests were as follows:

3.1.5.2.1 Hip abduction test (Figure 3-12)

The participant was placed lying on the non-tested side with the hip centre of rotation in alignment with the dynamometer's axis of rotation. The hip joint centre was defined as the intersection of two lines directed inferiorly from the anterior superior iliac spine and medially from the greater trochanter of the femur (Jacobs & Mattacola, 2005). A strap over the iliac crest was used to stabilise the hip, and a resistance pad for the moment arm was placed just above the lateral epicondyle. The participant was also strapped for the test limb in zero degrees of hip flexion, abduction and external rotation. They were asked to push their limb straight up towards the resistance pad.

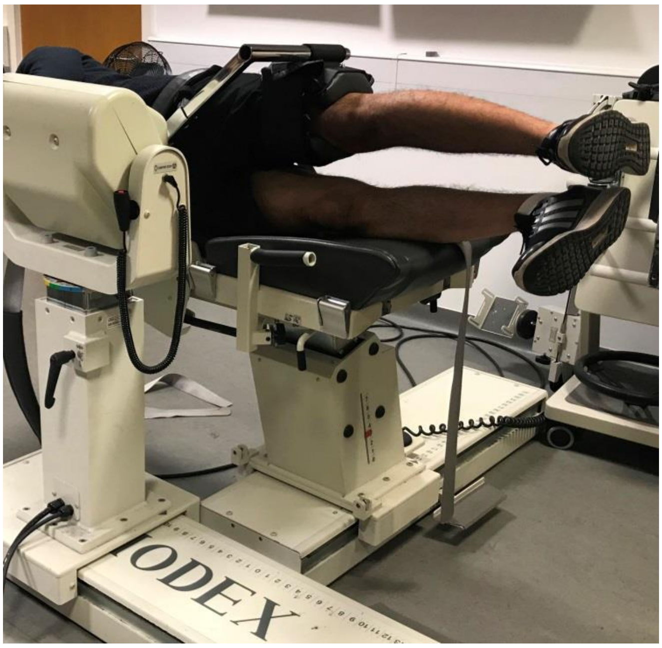


Figure 3-12 Participant position for MVIC test of the hip abductors

3.1.5.2.2 Hip extension test (Figure 3-13)

The participant was asked to lie prone with the limbs off the end of the Biodex, and the moment arm was positioned just above the popliteal fossa. The greater trochanter was aligned with the dynamometer axis of rotation. A goniometer was used for the test limb and placed in 30° of hip flexion and 90° of knee flexion. The participant lower back was strapped down. The participant was requested to push their thigh straight backwards while keeping 90° of knee flexion.

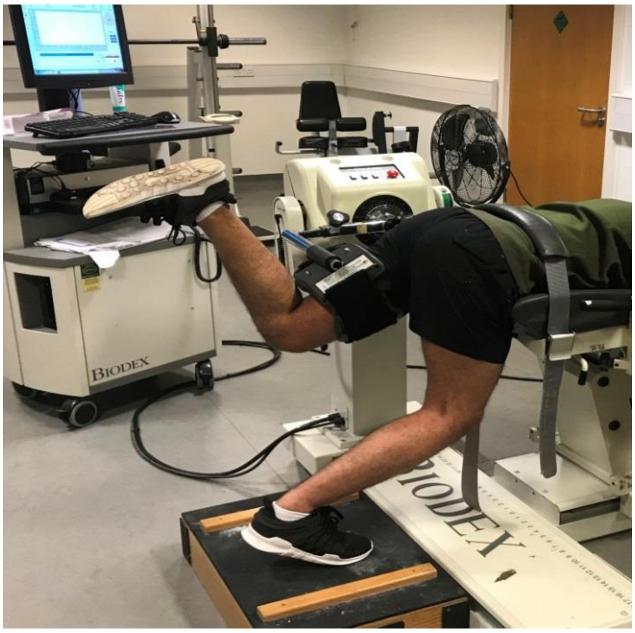


Figure 3-13 Participant position for MVIC test of the hip extensors

3.1.5.2.3 Hip external rotation test (Figure 3-14)

The participant was seated with limbs off the end of the Biodex. The moment arm was positioned just 5 cm above the medial malleolus, and the middle of the patellar tendon was aligned with the dynamometer axis of rotation. A goniometer was used, and the test limb was positioned at 90° of hip flexion and 90° of knee flexion. The test limb and hip were strapped down and the arms placed across the chest to ensure that no other hip movement occurred, including contraction of the hip adductors affecting the strength measure. The participant was requested to push their lower limb into the dynamometer towards the opposite lower limb.

Figure 3-14 Participant position for MVIC test of the hip external rotators

All measurements were carried out by one examiner and torque was corrected automatically for gravity by Biodex software. Gravity correction was incorporated by having the limb of the subject weighed prior testing (Meftah, Mahir, Lmidmani, & Elfatimi, 2016). The limb was weighed by the subject raising the limb slightly, holding the limb in that position, and the subject relaxing the limb completely while the dynamometer measured the limb weight. This gravity correction eliminates the error of weight that could be caused by the dynamometer arm and weight of the body segment (Anderson, Nussbaum, & Madigan, 2010). All muscle torque values were collected in Newton meter (Nm) and later normalised to each participant's body mass (Nm/kg). Normalisation to body mass allowed for more accurate comparison across participants and to the literature. Prior to testing, participants were asked to wear training clothes and testing was carried out at two different times, one test on the same day and the same test after one week. A maximum of 45 minutes was the time needed for testing. The participants were asked to perform five minutes of low intensity warm-up exercises using a cycle ergometer (Monark, Ergomedic 874 E), as advocated by Bell et al. (2013), to lessen the risk of physical discomfort and help to avoid any injuries during the tests (Woods et al., 2007).

3.1.6 Recruitment

The participants were recruited by advertising on posters, which were placed on notice boards around both universities.

3.1.7 Inclusion and exclusion criteria

Participants were physically active (attend at least 30 minutes of physical activity 3 times a week on a regular basis over the last 6 months), recreational healthy soccer players. Soccer is considered one of the most popular sports in the world (Junge & Dvorak, 2015). In the Kingdom of Saudi Arabia, soccer has grown and Saudis of all ages are participating. Participants were required to be free from lower extremity injury for at least six months prior to testing and have no history of lower extremity surgery. Injury was defined as any musculoskeletal complaint which stopped the participant from undertaking their normal exercise routine. Participants were also frequently performing COD task to 90° and 135° during their own sport. Before testing, each participant read and signed a written informed consent statement which approved by Research, Innovation and Academic Engagement Ethical Approval Panel at the University of Salford (Ethical number HSCR16-88). The age was limited from 18 – 35 years as it is the expected age for most of the athletes in most of sports, they are more prone to injury, and they are mostly those who our study would be applicable to (Griffin et al., 2000).

When the participants arrived at the laboratory, their demographic characteristics and past medical history were collected. Participants were also asked if they have read and understood the information sheet and if they have any questions. They were also asked to sign a consent form. Body mass and height were measured using an electronic personal floor scale (MARSDEN, charder, model: M-420), and height measure (Seca, United Kingdom).

3.1.8 Relationship between labs

As two different labs were used to collect the data in this thesis, it was expected that each lab would provide similar kinetics and kinematics results as the camera systems and also force platforms have been used in previous experiments. However, it was necessary to ensure this. For that, two subjects attended the two labs (Human performance lab (HPL) and Imam Abdulrahman Bin Faisal University lab (IABFUL)) where one investigator placed all markers. The participants performed five successful COD trials at 90º and 135 º angle of preferred and non-preferred limbs in each lab to determine whether there were any differences in the primary outcome measures (Knee valgus angle and moment, knee flexion angle, hip flexion, abduction and external rotation angles and vertical ground reaction force) between labs. The exact same procedures as undertaken in this thesis were performed. As can be seen from the table 3-3 and 3-4, both labs provided similar outcome measures values. Furthermore, the frontal, sagittal and transverse planes knee and hip angles were similar across labs as well as vertical ground reaction force (VGRF) and external knee abduction moment (EKAM), which supports that all labs provide similar kinetics and kinematics values during COD manoeuvres, regarding the type and location of the system. Although two labs were used in the current thesis, each participant visited one lab so they were their own control and thus any differences between the labs would not have an influence on this.

Although SEM is lower than the differences in some variables, some explanations could be offered to explain this. First, it should be noted that these data are only for two people and, when combining the two subjects' data there is less difference. It is true that there are some differences between the labs for only two subjects, but the reliability study is based on 10 subjects together. Another explanation could be due to variation in performance as the standard deviation in some variables is high. We cannot justify a significant difference between labs because we have only a small sample size (two subjects), which is a limitation and too underpowered. In addition, the actual SEM used is a combination from 10 individuals based in two different sessions. However, this data is only from one subject each who only made one visit. Furthermore, it could be suggested that there is a significant difference between labs, although any difference could just be measurement noise attributable to the instrument.

Table 3-3 Hip and knee joint angles, EKAM and VGRF of preferred and non-preferred limb during COD to 90º and 135º in the two labs for the first participant.

Variables	Preferred limb		Non-preferred limb	
	HPL	IABFUL	HPL	IABFUL
COD to 90º				
Hip Flexion (°)	48	52	54	50
Hip Abduction (°)	-19	-17	-18	-16
Hip Internal Rotation (°)	8	10	7	11
Knee Flexion (°)	67	65	64	66
Knee Abduction (°)	-7	-8	-5	-6
Knee Abduction moment (Nm/Kg)	0.98	1.1	0.8	.90
VGRF (*BW)	2	1.9	1.8	1.7
COD to 135º				
Hip Flexion (°)	53	54	57	56
Hip Abduction (°)	-24	-21	-22	-19
Hip Internal Rotation (°)	10	12	8	10
Knee Flexion (°)	73	72	69	70
Knee Abduction (°)	-8	-10	-7	-8
Knee Abduction moment (Nm/Kg)	1.5	1.4	1.6	1.8
VGRF (*BW)	2	2	1.7	1.8

Angle (°); newton meter per kilogram (Nm/Kg); body weight (BW); Human performance lab (HPL); Imam Abdulrahman Bin Faisal University lab (IABFUL).

Table 3-4 Hip and knee joint angles, EKAM and VGRF of preferred and non-preferred limb during COD to 90º and 135º in the two labs for the second participant.

Variables	Preferred limb		Non-preferred limb	
Angle (°)	HPL	IABFUL	HPL	IABFUL
COD to 90º				
Hip Flexion (°)	64	66	66	67
Hip Abduction (°)	-20	-20	-21	-19
Hip Internal Rotation (°)	11	9	12	13
Knee Flexion (°)	63	60	67	65
Knee Abduction (°)	-4	-4	-5	-5
Knee Abduction moment (Nm/Kg)	1.1	1.05	1.1	1.07
VGRF (*BW)	2.4	2.1	2	2.1
COD to 135º				
Hip Flexion (°)	67	69	68	70
Hip Abduction (°)	-25	-23	-24	-26
Hip Internal Rotation (°)	13	15	14	12
Knee Flexion (°)	65	69	72	69
Knee Abduction (°)	-6	-6	-7	-5
Knee Abduction moment (Nm/Kg)	1.7	1.5	1.6	1.7
VGRF (*BW)	2.3	2.2	2.3	2.2

Angle (°); newton meter per kilogram (Nm/Kg); body weight (BW); Human performance lab (HPL); Imam Abdulrahman Bin Faisal University lab (IABFUL).

3.2 The reliability of lower limb biomechanical variables during a change of direction to 90- and 135-degree manoeuvres.

Change of direction (COD) manoeuvres are crucial for many field sports, however they are unfortunately associated with non-contact anterior cruciate ligament (ACL) injury (Benis et al., 2018; Chinnasee et al., 2018; Johnston et al., 2018; Montgomery et al., 2018). The cause of non-contact ACL injuries is multifactorial and have been referred to poor mechanics. Abnormal lower limb biomechanics during activity has been postulated as a risk factor of non-contact ACL injury (Hughes, 2014). Research studies have found that changes in the knee valgus angle (abduction) and knee valgus moment (external abduction moment) are thought to increase risk of non-contact ACL injury (Hewett et al., 2005; Krosshaug et al., 2007; Pollard et al., 2015). Stress on the ACL is the greatest with an extended knee combined with knee valgus angle and moment and with internal tibial rotation (Montgomery et al., 2016; Myer et al., 2015). In addition, high knee valgus angle and moment during COD manoeuvres is associated with joint positions including increased hip flexion, abduction and internal rotation angles (Havens & Sigward, 2015a; Imwalle et al., 2009; Sigward & Powers, 2007). Therefore, it is important to assess non-contact ACL injury biomechanical risk factors, kinematics and kinetics, during COD manoeuvres.

Most of the studies of lower-limb biomechanics have utilised three-dimensional (3D) motion-analysis systems (Alenezi, Herrington, Jones, & Jones, 2016; Mok, Bahr, & Krosshaug, 2018). In fact, the gold standard for examining lower limb biomechanics is 3D motion analysis system and allows researchers to calculate motion in all three planes during dynamic manoeuvres (Meldrum, Shouldice, Conroy, Jones, & Forward, 2014; Munro, Herrington, & Carolan, 2012). Few studies have evaluated the reliability of lower limb biomechanical measurements during COD manoeuvres at 45° (Mok et al., 2018; Sankey et al., 2015) and at 90° (Alenezi et al., 2016). However, limited knowledge is available on the kinetic and kinematic data reliability of lower limb during COD at 90° and 135° angle.

Using reflective markers placed on certain anatomical landmarks, allows the skeletal system to be focused on and biomechanical features to be easily recorded and measured throughout functional manoeuvres. Accurate and reliable 3D lower extremity measurements during the performance of different sporting manoeuvres have been obtained using high speed motion analysis technologies (Ford et al., 2003; Gao, Cordova, & Zheng, 2012; Sled, Khoja, Deluzio, Olney, & Culham, 2010; Zeller, McCrory, Kibler, &

Uhl, 2003), and their use has assisted in the screening and rehabilitation of injuries related to these manoeuvres. 3D motion analysis studies are usually carried out under controlled laboratory conditions, they are useful for replicating specific high-risk postures that can arise during injury, and they enable scientists to study joint biomechanics, kinematics and kinetics, during these high-risk activities by replicating motion patterns and dynamic body control. However, misplacement of retroreflective marker produces the greatest errors in 3D motion analysis (Ford, Myer, & Hewett, 2007; Malfait et al., 2014), such as mistakes in identifying marker locations, which affects the calculations used to determine the positions of joint centres, resulting in errors in joint kinematic and kinetic calculations (Baker, 2006). One way in which these errors can be reduced is to place markers on rigid plates fixed to the thigh and shank, as this has been demonstrated to result in less movement than those applied directly to the skin (Manal, McClay, Stanhope, Richards, & Galinat, 2000). Moreover, one examiner attached all the markers in all trials of the study, it would control this error. In addition, Calibration Anatomical Systems Technique (CAST) were used to determine each segment's movement during the trial (Cappozzo, Catani, Croce, & Leardini, 1995) to reduce the skin movement artefact by attaching the markers in centre of the segment rather than close to the joints. However, there is still the anatomical placement of the local coordinate system markers that can be variable, and each study should examine the investigators reliability in placing these markers before embarking on larger scale studies. Reliability of the measurements depends on various important issues, such as: daily calibration of the cameras; accuracy of marker placement; good and effective training of the examiner; and system updates (McGinley, Baker, Wolfe, & Morris, 2009).

There is another way in which joints and segments can be defined. The Joint Coordinate System (JCS) is another approach to calculate the centre of the knee joint. This functional approach is where the axis of the two body segments (shank and thigh, for example) is used to calculate the centre of the knee joint. Schwartz and Rozumalski (2005), who describe a new method for joint parameter estimation by using functional approach, suggest that the functional method is objective, precise, and practical. However, this study did not use the functional method, and instead two markers were placed in either knee joint. It should be noted that, based on their characteristics, the subjects of this study are recreational athletics and thus, identifying palpatable anatomical landmarks is easier. This approach has been used in biomechanic labs worldwide. Therefore, it is important to maintain consistency with the literature. In addition, a reliability study was undertaken to ensure that the author could accurately and repeatedly place the markers in the same place. This was seen in the repeatability study, thus confidence was gained and the functional knee joint axis was not required. Therefore, it is important to examine the between-day reliability of lower-limb kinematic and kinetic variables during COD manoeuvres at 90° and 135° using a 3D motion analysis system.

3.2.1 Study aim:

To investigate the between-days reliability of lower limb kinematic and kinetic variables during a COD manoeuvre at 90° and 135° using 3D motion analysis.

3.2.2 Null hypothesis:

The outcome measures are not reliable and significant differences occur during testing sessions between days.

3.2.3 Methods:

The reliability study was carried out in the Human Performance Lab at the University of Salford.

For the details on the motion capture please refer to the method chapter (section 3.1).

For the details on the three-dimensional motion capture and markers placement please refer to the method chapter (section 3.1.3).

For the details on the data processing please refer to the method chapter (section 3.1.4).

For the details on the study procedure please refer to the method chapter (section 3.1.5.1).

For the details on the recruitment please refer to the method chapter (section 3.1.6).

For the details on the inclusion and exclusion criteria please refer to the method chapter (section 3.1.7).

3.2.3.1 Repeatability

Testing was carried out during two separate testing sessions. The same testing procedures were used with each participant during two different sessions separated with one week apart. The participants were assessed at similar times of the day to minimise the impact of diurnal variation.

3.2.3.2 Main outcome measures

The following discrete variables were calculated during stance phase for each trial:

- Peak VGRF.
- Peak external knee abduction moments.
- Peak joint angles (hip and knee in sagittal, frontal and transverse planes).

3.2.3.3 Statistical analysis

The statistical analyses were carried out using Statistical Package for Social Sciences software (version 21, SPSS Statistics 20.Ink). The mean of five trials from the first and second sessions were used to calculate the between day reliability. Intra class correlation (ICC) was utilised to assess the consistency and conformity of the measurements (Field, 2013). ICC model two-way mixed effects was used to calculate the reliability since only principle investigator conduct all measurements. This ICC model was chosen based on recent study which gave a guideline on how to choose the right ICC model for reliabilities studies (Koo & Li, 2016). ICC values were interpreted according to (Coppieters, Stappaerts, Janssens, & Jull, 2002): Poor < 0.40, Fair 0.40 to 0.70, Good 0.70 to 0.90, Excellent ≥ 0.90. The ICC is a unitless measurement with value ranging from 0 to 1 with higher value indicate better result. Confidence intervals (CI) and standard deviation (SD) were also calculated and presented in the results. Although ICC, as a reliability measurement, appeared easily interpreted, it does not provide the amount of disagreement between the measurement and is an arbitrary value. Therefore, to gain the full picture, the Standard Error of Measurement (SEM) was calculated, which provides the error value in the same unit as the measurement. SEM is an important measurement as it provides an estimated prediction for any measurement, which gives the range for where the true value of any measurement is likely to lie (Denegar & Ball, 1993). Knowing such information allows for an accurate evaluation between tests changes and thus, determines whether the change is real or due to measurement error (Carter, Lubinsky, & Domholdt, 2013; Munro et al., 2012). The SEM was computed to determine absolute reliability, with low values representing good reliability. This was used to help the researcher estimate the real change over the measurement error (Baumgartner, 1989).

SEM was calculated based on the following formula.

SD (pooled) * (√ (1-ICC)) (Thomas, Silverman, & Nelson, 2015).

3.2.4 Result:

10 recreational healthy soccer players were recruited to take part in the study. All the participants were male (age 22 ± 4 years; height 1.73 ± 0.05 m; and mass 66 ± 10 kg). Table 3-5 contains the ICC values for (95% CI), day 1 and day 2 means, SEM, and SD for hip and knee kinematics and kinetics variables for COD at 90°. The ICC values for all variables ranged between 0.88 and 0.98, reporting good to excellent reliability. The SEM

value ranged between 0.48 to 1.20 degree for angles, 0.03 Nm/kg for knee abduction moments and 0.01 *BW for GRF.

Table 3-5 COD at 90° Intra-class Correlations (ICC), Confidence Intervals (CI), SEM, Mean and SD

Variables	ICC (95%CI)	Day 1 Mean (SD)	Day 2 Mean (SD)	SEM
Hip Flexion (°)	0.92 (0.68-0.98)	49 (6)	48 (7)	0.59
Hip Adduction (°)	0.92 (0.66-0.98)	-12 (6)	-12 (5)	0.61
Hip Internal Rotation (°)	0.90 (0.56-0.97)	8 (7)	8 (7)	1.14
Knee Flexion (°)	0.88 (0.50-0.97)	62 (7)	64 (9)	1.20
Knee Abduction (°)	0.89 (0.57-0.97)	-6 (3)	-7 (4)	0.48
Knee Abduction moment (Nm/Kg)	0.91 (0.65-0.97)	0.82 (0.28)	0.83 (0.22)	0.03
VGRF (*BW)	0.98 (0.92-0.99)	2.27 (0.45)	2.30 (0.44)	0.01

Intra-class Correlations (ICC); Confidence Intervals (CI), standard error of measurement (SEM); standard deviation (SD); Angle (°); newton meter per kilogram (Nm/Kg); body weight (BW).

Table 3-6 contains the ICC values for (95% CI), day 1 and day 2 means, SEM and SD for hip and knee kinematics and kinetics variables for COD at 135°. The ICC values for all variables ranged between 0.85 and 0.95, reporting good to excellent reliability. The SEM value ranged between 0.44 to 1.68 degree for angles, 0.10 Nm/kg for knee abduction moments and 0.05 *BW for GRF.

Table 3-6 COD at 135° Intra-class Correlations (ICC), Confidence Intervals (CI), SEM, Mean and SD

Variables	ICC (95%CI)	Day 1 Mean (SD)	Day 2 Mean (SD)	SEM
Hip Flexion (°)	0.90 (0.58-0.97)	52 (5)	51 (7)	0.63
Hip Adduction (°)	0.92 (0.68-0.98)	-15 (6)	-16 (5)	0.66
Hip Internal Rotation (°)	0.85 (0.41-0.96)	10 (9)	7 (7)	1.68
Knee Flexion (°)	0.92 (0.67-0.98)	67 (9)	68 (10)	0.89
Knee Abduction (°)	0.90 (0.58-0.97)	-7 (3)	-7 (3)	0.44
Knee Abduction moment (Nm/Kg)	0.90 (0.58-0.97)	0.85 (0.24)	0.81 (0.24)	0.10
VGRF (*BW)	0.95 (0.78-0.99)	2.18 (0.48)	2.14 (0.50)	0.05

Intra-class Correlations (ICC); Confidence Intervals (CI), standard error of measurement (SEM); standard deviation (SD); Angle (°); newton meter per kilogram (Nm/Kg); body weight (BW).

3.2.5 Discussion:

The purpose of this study was to examine the between-day reliability of the lower limb biomechanical variables during CODs at 90° and 135° in recreational healthy soccer players. The between-days ICC values and SEM were calculated for the kinematic, kinetic and GRF variables during COD at 90° and 135° degree.

The results of this study show that all 3D variables in both manoeuvres reported good to excellent between day reliability, ranging in ICC values between 0.85 and 0.98 for both manoeuvres COD at 90° (Table 3) and 135° (Table 4). The COD manoeuvres investigated in the current study showed similar between-day knee kinematics and kinetics and GRF reliability characteristics in comparison to previous work with previous studies (Alenezi et al., 2016; Mok et al., 2018). However, with COD at 90° (Alenezi et al., 2016) and 45° (Mok et al., 2018), the between-day hip kinematics is more reliable in COD manoeuvres at 90° and 135° degree (ICC= 0.90-0.92) in this study, than COD manoeuvre at 90° degree (ICC= 0.51-0.75) and 45° degree (ICC= 0.66-0.74). This may because that when placing markers, the distance was measured by tape to get better results. In addition, participants in the current study are frequently performing COD manoeuvre at 90° and 135° during their own sport so can do manoeuvres more consistency. In addition, the findings of this study in both manoeuvres showed that GRF reported the highest ICC values (between 0.95 and 0.98) compared to other kinematic and kinetic variables, which is consist with previous literature (Alenezi et al., 2016; Mok et al., 2018). These findings can be attributed to GRF being the sum of segmental acceleration, mass and gravitational vector. Therefore, no marker is needed to calculate the GRF and it does not suffer from marker placement error, therefore the assumption can be made that GRF is more repeatable.

SEM is an important measurement as it provides an estimation of prediction for any measurement which gives the range for where the true value for any measurement is likely to lie (Denegar & Ball, 1993). Knowing such information about any measurement allows for accurate evaluation for between tests changes and thus, determines whether the change is a real change or due to measurement error (Carter, Lubinsky, & Domholdt, 2013; Munro et al., 2012). In this study, the standard error of measurement for all joint angle values ranged from 0.44 to 1.68 degree (Table 3 and 4). The majority of the previous studies' SEM less than 5 degrees, which is consistent with this study's values (Alenezi et al., 2016; McGinley et al., 2009). Error less than 5 degrees is likely to be considered reasonable for clinical situation use (McGinley et al., 2009). However, the SEM for the hip internal rotation was greater than 10% of the mean value. This was due to the greater standard deviation seen on day 1 and day 2, indicating a greater variation in performance across individuals. However, it is reliable in an ICC (0.90, 95%CI (0.56-0.97)). The knee abduction moment for 90° and 135° angle COD manoeuvres showed SEM values 0.03 and 0.10. The GRF for 90° and 135° angle COD manoeuvres showed SEM values between 0.01 and 0.05.

The result of this study is subject to some limitations. First, the generalisability of result is restricted to similar laboratory setting, the model and the researcher ability to apply markers. Second, the participants in this study were examined using standardised shoes and on Mondo running surface. However, the interaction between such shoe and surface may not reflect the actual interaction for some sport such as football or other sports that are played on grass. Hence, there is a need to examine participants wearing real sport shoes on a grass surface. Finally, the testing was carried out on recreational healthy soccer players. Therefore, it is applicable only for the same population. Other populations need to be examined.

3.2.6 Conclusion

All the biomechanical variables examined achieved good to excellent between-day reliability. The result of this study leads to the suggestion that COD manoeuvre at 90° and 135° angles are a reliable manoeuvre. The current results support the use of these methods in the following study to examine the lower limb during 90° and 135° COD manoeuvres.

3.3 The reliability of isometric strength testing of hip Abductor, Extensor and External rotation muscles using isokinetic dynamometer.

Hip muscles play an important role in sport for several fundamental skills such as kicking, accelerating and change of direction (COD) (Thorborg, Couppe, Petersen, Magnusson, & Holmich, 2011) and in normal function of the lower limb (Marshall, Patel, & Callaghan, 2011). Literature shows that isometric hip muscle weakness has been reported to independently predict anterior cruciate ligament (ACL) injuries (Khayambashi et al., 2016), indicating that weakness in hip muscles is a modifiable risk factor. In addition, several investigators have reported that decreased hip muscle strength is associated with greater knee valgus angles (Gehring et al., 2009; Hollman et al., 2009; Jacobs et al., 2007; Willson et al., 2006) and knee valgus moments (Lawrence et al., 2008). These findings are concerning because greater knee valgus angles have been identified as mechanisms and characteristics linked to ACL injuries (Koga et al., 2010; Montgomery et al., 2018; Walden et al., 2015) and are linked to increase knee valgus moment (Jones et al., 2015; Kristianslund et al., 2014; Sigward et al., 2015), which can increase ACL strain (Markolf et al., 1995; Markolf et al., 1990). These findings, therefore, point towards the importance to evaluate hip muscle strength and the calculation of bilateral torque asymmetry, which are often used in sports medicine to assess the strength of the hip joint and to monitor potential hip-related injuries.

The assessment of isometric hip muscle strength can be done using different measurement tools such as: manual muscle testing, hand-held dynamometer (HHD) and isokinetic dynamometer. The manual muscle testing, rating muscle strength from 0 to 5, is considered a subjective tool with low reliability and insensitive (Andersen & Jakobsen, 1997; Frese, Brown, & Norton, 1987; Wadsworth, Krishnan, Sear, Harrold, & Nielsen, 1987). The other two methods are more objective tools. HHD has the advantage of being user-friendly with suitable size and low cost with an established reliability (Thorborg, Bandholm, & Holmich, 2013). However, a major limitation of HHD is the lack of standardization of the participants' starting position and the placement of the assessor and strength imbalance between the participant and the assessor. The isokinetic dynamometer has been considered as a gold standard measurement tool for assessing isometric hip muscle strength and become more popular in sport, research and clinical setting (Campos Jara et al., 2014; Kemp, Schache, Makdissi, Sims, & Crossley, 2013). It has good reliability for measuring muscle strength mainly because the standardisation and also the results are not influenced by strength imbalance between the participant and the assessor (Martin et al., 2006; Meyer et al., 2013). However, very few studies have focused on the between-day reliability of isometric hip strength assessment using the isokinetic dynamometer (Arokoski et al., 2002; Meyer et al., 2013; Widler et al., 2009).

A comparison between various positions for isometric hip strength measures was carried out by Widler et al. (2009), who found side-lying to be the most valid and reliable method for measuring hip abductors torque and a similar position has been supported by Meyer et al. (2013). In addition, Prins and van der Wurff (2009) measured external rotator strength while seated, with the hip and knee at 90° angles. For hip extension testing, participants were asked to lie prone on the testing table with the limbs off the end of the table and the non-tested limb on the ground while the tested limb was positioned at 30° of hip flexion while actively maintaining the knee in 90° of flexion (Cronin, Johnson, Chang, Pollard, & Norcross, 2016; Stearns, Keim, & Powers, 2013; Teng & Powers, 2016; Worrell et al., 2002).

The reliability of isometric assessments could be influenced by a number of issues when using an isokinetic dynamometer, including the testing process, positional specifics used in the assessment, motivation and learning effect (Verma, Juneja, Verma, Dhyani, & Khanna, 2010), not forgetting the instructions provided to participants during testing (Sahaly, Vandewalle, Driss, & Monod, 2001) and motivation, learning effect and age (Svensson, Waling, & Hager-Ross, 2008). In order to ensure the effective application of torques during the isometric testing, instructions must include stating “as hard as possible” to make sure that maximal effort is exerted throughout (Sahaly et al., 2001). The reliable evaluation of hip abduction, extension and external rotation muscle strength is fundamental for accurate monitoring of strength, training planning, and injury prevention/rehabilitation. Therefore, the aims of this study were to evaluate the between-days reliability for assessment of isometric peak torque of hip abductor, extensor and external rotator muscles using isokinetic dynamometer (Biodex System 4 dynamometer).

3.3.1 Study Aim:

To investigate the between-days reliability of isometric peak torque of the hip abductors, extensors, and external rotators muscles using isokinetic dynamometer (Biodex System 4 dynamometer).

3.3.2 Null hypothesis:

The outcome measures are not reliable and significant differences occur during testing sessions.

3.3.3 Methods:

For the details on the study procedure please refer to the method chapter (section 3.1.5.2).
For the details on the recruitment please refer to the method chapter (section 3.1.6).
For the details on the inclusion and exclusion criteria please refer to the method chapter (section 3.1.7).

3.3.3.1 Repeatability

Testing was carried out during two separate testing sessions. The same testing procedures were used with each participant during two different sessions separated with one week apart. The participants were assessed at similar times of the day to minimise the impact of diurnal variation.

3.3.3.2 Main outcome measures

After recording the results of the five trials for each participant, the mean value over the five trials for both limbs was calculated and reported. The main outcome measure is normalised peak torque of hip abductors, extensors, and external rotators muscles.

3.3.3.3 Statistical analysis

The statistical analyses were carried out using Statistical Package for Social Sciences software (version 21, SPSS Statistics 20.Ink). The normalised mean peak torque of five trials from the first and second sessions were used for between day reliability. Intra class correlation coefficient (ICC) was utilised to assess the consistency and conformity of the measurements (Field, 2013). ICC model two-way mixed effects was used to calculate the reliability since only principle investigator conduct all measurements. This ICC model was chosen based on recent study which gave a guideline on how to choose the right ICC model for reliabilities studies (Koo & Li, 2016). ICC values were interpreted according to (Coppieters et al., 2002): Poor < 0.40, Fair 0.40 to 0.70, Good 0.70 to 0.90, Excellent ≥ 0.90. The ICC is a unitless measurement with value ranging from 0 to 1 with higher value indicate better result. Confidence intervals (CI) and standard deviation (SD) were also calculated and presented in the results. Although ICC as a reliability measurement appear to be easy interpreted, it does not provide the amount of disagreement between the measurement. Therefore, to gain the full picture standard error of measurement (SEM) was calculated which provide the amount error in the same unit. SEM was calculated based on the following formula.

SD(pooled) * (√ (1-ICC)) (Thomas et al., 2015)

3.3.4 Result:

10 recreational healthy soccer players were recruited to take part in the study. All the participants were male (age 21 ± 4 years; height 1.73 ± 0.05 m; and mass 66 ± 10 kg). Table 3-7 contains the ICC values for (95% CI), day 1 and day 2 means, SEM, and SD of the normalised peak torque of the hip abductors, extensors, and external rotator muscles. The ICC values for normalised peak torque of the hip abductors, extensors, and external rotators muscles were excellent, ranged between 0.92 and 0.95. The SEM values for normalised peak torque of hip abductors ranged from 0.01 to 0.03 Nm/kg.

Table 3-7 Between session Intraclass Correlation Coefficient (ICC), 95% Confidence Intervals (CI), Means, SEM and SD for normalised peak torque.

Variables Nm/Kg	ICC (CI)	Day 1 Mean (SD)	Day 2 Mean (SD)	SEM
Hip Abductors	0.92 (0.70-0.98)	1.42 (0.25)	1.41 (0.25)	0.03
Hip Extensors	0.92 (0.67-0.98)	1.68 (0.33)	1.64 (0.40)	0.03
Hip External Rotators	0.95 (0.80-0.99)	0.76 (0.17)	0.77 (0.06)	0.01

Intra-class Correlations (ICC); Confidence Intervals (CI), standard error of measurement (SEM); standard deviation (SD); newton meter per kilogram (Nm/Kg).

3.3.5 Discussion:

The main objective of the study was to investigate the between-days reliability of isometric normalised peak torque of the hip abductors, extensors, and external rotators muscles using isokinetic dynamometer (Biodex System 4 dynamometer) in recreational healthy participants. In the present investigation, the normalised peak torque showed excellent reliability. The ICC values ranged between 0.92 and 0.95 along with low SEM. For all measurements, SEM ranged from 0.01 to 0.03 Nm/kg, which are less than two percent of the mean.

Generally, the sport science literature shows that ICC is typically reported when checking if a measurement is reliable (Atkinson & Nevill, 1998; Morrow & Jackson, 1993). This method is usually used to examine the reliability of the peak torque during the isometric test. In this study, the peak torque was found to have excellent reliablility based on range of ICC = 0.92 – 0.95. Similar results were found in the literature (Arokoski et al., 2002; Meyer et al., 2013; Widler et al., 2009). Meyer et al. (2013) assessed hip extensor and abductor peak torque intraclass correlation coefficient (ICC), which found that isometric hip abduction was found to be highly reliable (ICC = 0.91), whereas, hip extension was moderate reliable (ICC = 0.77). Widler et al. (2009) showed similar findings for hip abduction (ICC = 0.90). Regarding normalised peak torque of the hip abductors, the ICC value of 0.92 was found which is relatively higher than the one reported by Arokoski et al. (2002) (ICC=0.84) who measured torque in a supine position. However, comparing results for hip abduction in side-lying position from current study to results from other studies is rather difficult since testing position has been under debate and only few studies were dedicated to side-lying testing (Meyer et al., 2013; Widler et al., 2009) versus supine position (Arokoski et al., 2002).

The study was not without limitations. First, the generalisability of result is restricted to similar laboratory setting, the participant's position and the researcher ability to apply the method protocol. Second, the testing was carried out on recreational healthy participants. Therefore, it is applicable only for the same population. Other populations, such as injured people need to be examined.

3.3.6 Conclusion

The current study has demonstrated that normalised peak torque for hip muscles show excellent reliability, along with low standard error of measurement for recreational healthy participants. These results are relevant to those undertaking hip strength measurements. The current results support the use of these methods in the following study.

4 Chapter 4

Biomechanical and strength asymmetry in preferred and non-preferred limbs during 90° and 135° change of direction manoeuvres and MVIC test.

Presented at the Scandinavian Sports Medicine Congress, 31 January - 2 February, 2019 in Copenhagen, Denmark.

4.1 Introduction

Change of direction (COD) manoeuvres are crucial for many field sports, however they are unfortunately a key action associated with non-contact anterior cruciate ligament (ACL) injuries (Benis et al., 2018; Boden et al., 2000; Brophy et al., 2015; Chinnasee et al., 2018; Cochrane, Lloyd, Buttfield, Seward, & McGivern, 2007; Johnston et al., 2018; Koga et al., 2010; Montgomery et al., 2016, 2018; Olsen et al., 2004; Waldén et al., 2015). Players using both limbs allow for variety when performing COD manoeuvres. Hand dominance is straightforward defined by the preferred hand to write; however, limb dominance is sometimes more ambiguous. Limb dominance is traditionally described as the preferred limb to kick a soccer ball (Peters, 1988). However, players may not usually use their dominant limb as a push-off limb during COD manoeuvres, they may use the preferred limb during COD manoeuvres rather than dominant limb. The influence of limb asymmetry on ACL injury risk is important issue. During match play, the athlete will be challenged to perform the COD manoeuvre in both directions at different angles, with both the preferred and non-preferred limb as the push-off limb. However, Hewett et al. (2005) carried out an important prospective study of ACL injury. The authors found that there was a significantly larger limb-to- limb difference in knee abduction moment in athletes that subsequently experienced an ACL injury, compared to those who remained un-injured. Furthermore, mechanical differences between the two limbs (asymmetry) may be a precursor to injury (Kyritsis et al., 2016). In addition, non-contact ACL injuries may not always happen on the dominant or preferred limb. The ACL injury rates for non-dominant limb range from 43% to 67% (Brophy et al., 2010; Goerger et al., 2014; Matava et al., 2002; Negrete et al., 2007). However, whether or not biomechanical and muscular limb asymmetry has an effect on mechanical knee joint loading in healthy recreational soccer players is still inconclusive. Side-to-side differences in biomechanical and muscular lower limbs may result in a failure to maintain neutral lower extremity alignment during landing, thus increase load in one side than other so both limbs might be affected. Clinically, it is important to understand if limb asymmetry is an etiological factor for ACL tears, as this would lead to further development of targeted interventions to address asymmetries between limbs.

Very few studies have investigated biomechanical side-to-side differences during planned COD manoeuvres and those that have explored this area have only considered small angles, such as COD to 45° degrees (Bencke et al., 2013; Brown et al., 2014b; Greska et al., 2016; Pollard et al., 2018) and 105° (Marshall et al., 2015). It should be noted that previous studies only investigated the differences between limbs during COD manoeuvres at 45° and 105°, while the current study used different COD angles (90° and 135° degree). It is essential to address limb asymmetry during 90° and 135° COD manoeuvres because the angles of such manoeuvres at the time of non-contact ACL injuries ranged between 30-180° (Montgomery et al., 2018; Waldén et al., 2015). Knee kinetics and kinematics contribute to an increased COD manoeuvre angle, as sharper angles increase the mechanical demand on the knee (Vanrenterghem et al., 2012). Surprisingly, to date, no study exists on limb asymmetry during COD manoeuvres at 90° and 135° angles.

A recent systematic review found conflicting results when literature was evaluated on the effect of limb dominance on COD biomechanics which were associated with an increased risk of ACL injury (Dos'Santos et al., 2019a). Brown et al. (2014b) discovered differences between the dominant and non-dominant limbs in healthy females during planned COD to 45° angles when considering the knee flexion angle, knee abduction angle, knee internal rotation angle and knee abductor moment. Recently, Pollard et al. (2018) also examined differences between dominant and non-dominant limbs for planned COD at 45° angles, and found that healthy individuals showed little difference in side-to-side movements, but noted significant differences between the two limbs when the peak knee internal rotation angle was greater in the non-dominant limb.

Furthermore, in terms of other joints, another study has investigated the differences between dominant and non-dominant limbs during COD to 105° (Marshall et al., 2015). They found that internal ankle rotation moments were significantly greater in the non-dominant limb. The ankle joint was also significantly more dorsiflexed on the non-dominant side. Recently, there has been an increase in studies that compare the biomechanical differences between limbs after ACLR. It has been shown that kinematic and kinetic differences between limbs are apparent during planned 90° COD tests at nine months post ACLR (King et al., 2018), involving lower knee flexion angles, external ankle rotation moments, external knee rotation moments and knee extension moments, as well as lower knee internal rotation in the ACLR limb (King et al., 2018). In contrast, Greska et al. (2016)

studied biomechanical differences between dominant and non-dominant limbs during the performance of a COD manoeuvre at a 45° angle, and found that female collegiate soccer athletes displayed similar lower extremity biomechanics between the dominant and non-dominant limbs when considering the peak knee abduction angle and moment. This accords with previous research that has shown a lack of difference between dominant and non-dominant limbs while performing 45° COD manoeuvres (Bencke et al., 2013; Matava et al., 2002).

It should be mentioned that all the previous studies have defined the preferred limb as that with which individuals prefer to kick the ball. However, in the current study the preferred limb was determined by asking participants which limb they would preferred to use as the push-off limb during COD manoeuvres (more detail is provided in Chapter 2, section 2.3.7.2). Moreover, previous studies investigating biomechanical symmetry in COD manoeuvres have typically done so using discrete points (e.g. peak values). However, there are a number of limitations with this type of analysis. First, asymmetry may occur over phases that are not captured in a single data point. Secondly, the discrete points utilised may vary between studies and therefore using one discrete point analysis alone may not detect all significant asymmetries. In the current study, the biomechanical analysis symmetry was undertaken using multiphases and discrete points. More detail about limb dominance is provided in Chapter 2, section 2.3.7.2.

Ultimately, it remains questionable whether limb dominance is an ACL injury-risk with 90° and 135° COD manoeuvres (Dos'Santos et al., 2019a). Moreover, there is a lack of scientific data describing the relationship between limb preference and knee mechanics and how these variables affect risk factors for ACL injury during planned 90° and 135° COD manoeuvres (Dos'Santos et al., 2019a). Therefore, more investigations are necessary to decide whether biomechanical asymmetries exist between preferred and non-preferred lower limbs during planned COD manoeuvres at 90° and 135° angles. Therefore, the aim of the current study was to determine whether asymmetry in knee and hip biomechanics kinematics and kinetics and hip muscle strength between preferred and non-preferred limbs during COD manoeuvres at 90° and 135° angles and MVIC test exists.

4.2 Research questions:

- Are there any differences in the biomechanical variables between preferred and non-preferred limbs during 90° and 135° COD manoeuvres?
- Are there any differences in the isometric hip abductors, extensors and external rotators strength between preferred and non-preferred limbs?

4.3 Null hypotheses

- There are no significant differences in hip and knee biomechanical variables between limbs during 90° and 135° COD manoeuvres.
- There are no significant differences in the hip muscles normalised peak torque between limbs during MVIC test.

4.4 Methodology

For the details on the motion capture at both of the laboratories please refer to the method chapter (section 3.1.1 and 3.1.2).
For the details on the markers placement please refer to the method chapter (section 3.1.3).
For the details on the study procedure please refer to the method chapter (section 3.1.5).
For the details on the recruitment please refer to the method chapter (section 3.1.6).
For the details on the inclusion and exclusion criteria please refer to the method chapter (section 3.1.7).

4.4.1 Data processing

The raw marker trajectory data from the 3D motion analysis systems in both labs were reconstructed in Qualisys Track Manager Software (version 2.16) or Vicon Nexus software (Version 2.6.1). For the details on the data processing please refer to the method chapter (section 3.1.4). Then, the seven motion events were created after the construction of a six degrees of freedom model, as displayed in figure 4-1. Firstly, initial contact (IC) and toe off (TO) were determined based on the force plate data. The IC was defined when vertical ground reaction force (VGRF) first exceeded 10 Newton and TO was defined when VGRF fell below 10 Newton. Other four events were created named: peak vertical ground reaction force (PVGRF), peak external knee abduction moment (PEKAM), first 60 millisecond of stance (60 ms) and peak knee valgus angle (PKVA).

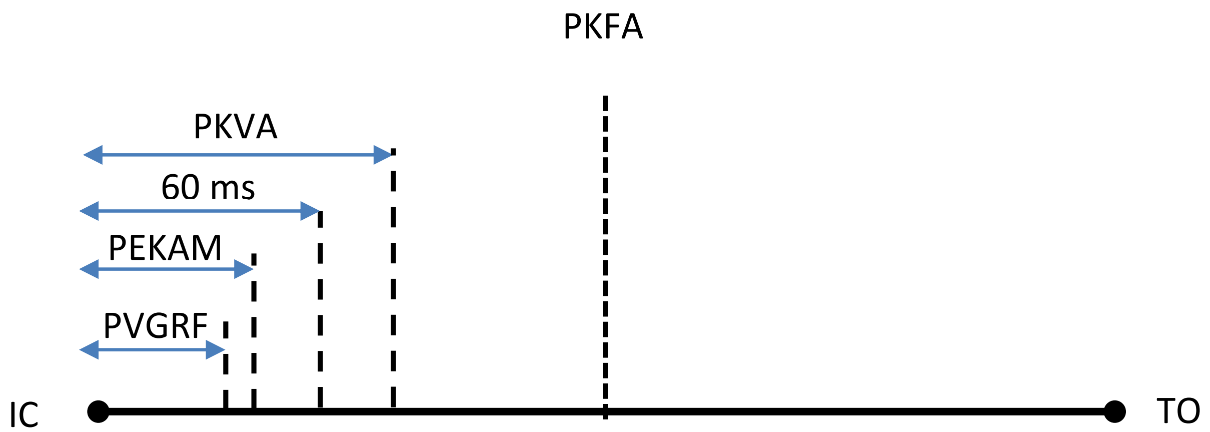


Figure 4-1 Sub-phases during COD manoeuvre contact phase

The kinematic, kinetic and GRF data were recorded in the stance phase. The curve in the kinematics of the knee sagittal and frontal planes were divided into two phases defined as the following: the initial contact phase (0%) and first peak (abduction). The curves in the kinetics data in knee y moment and GRF were divided into two phases: IC and first peak. The curve of kinematics in the hip sagittal, frontal and transverse planes were divided into five phases defined as the following: (1) at initial contact (IC), (2) IC to first peak vertical ground reaction force (PVGRF), (3) IC to first peak external knee abduction moment (PEKAM), (4) IC to the first 60 millisecond of stance (60 ms) and (5) IC to first peak knee valgus angle (PKVA). 3D variables were analysed during those phases for several reasons. Those phases usually happened at first 50% of stance phase and non-contact ACL injuries often occur during this period of stance (Boden et al., 2000; Chaudhari, Hearn, & Andriacchi, 2005; Jamison, Pan, & Chaudhari, 2012; Leppanen et al., 2017). Specifically, the estimated time of injury ranged between 17 and 60 milliseconds after initial contact (Bates et al., 2020; Koga et al., 2010; Krosshaug et al., 2007). Furthermore, PKVA, PEKAM and PVGRF during dynamic manoeuvres are thought to increased risk for noncontact ACL injuries (Hewett et al., 2005; Malinzak et al., 2001; McLean et al., 2004; McLean et al., 2005; Sigward & Powers, 2006; Yu, Lin, & Garrett, 2006). However, there is a lack of evidence on the role of hip frontal, transverse and sagittal planes on the knee frontal plane loading and motion during those phases. Focusing on those phases could provide an additional understanding of the hip and knee kinematic and kinetic and the greater risk to ACL injury. The joint range of motion (ROM) was calculated by subtracting the minimum value from the maximum value throughout the interested phases. Therefore, examining knee and hip joint displacements alongside peak angles during different time periods: could provide more complete analysis and additional understanding of the hip functions and their relationship to ACL injury mechanism. Kinematic and kinetic data were time normalised to 100% of the right and left limbs contact phase. The mean of five trials was used to calculate each interested outcome in the kinematic, kinetics and GRF data. The peak across the stance phase were calculated for the following outcomes: hip x angle, hip y angle, hip z angle, knee x angle, knee y angle and knee y moment. Biomechanical variables were exported as spreadsheets into Microsoft Excel to conduct the analysis and construct a graphical illustration.

Hip muscles peak torque was processed and analysed using Biodex System 4 with associated Software (Biodex Medical Systems, Inc, Shirley, New York, USA). After recording the results of the five trials for each participant, the mean value over the five trials for both limbs was calculated and reported. All muscle torque values were collected in Newtons meter (Nm) and later normalised to each participant's body mass (Nm/kg). Normalisation to body mass allowed for more accurate comparison across participants and to the literature.

4.4.2 Main outcome measures

The following discrete variables were calculated during stance phase for each trial:

- Peak vertical ground reaction force (PVGRF).
- Peak external knee abduction moment (PEKAM).
- Knee valgus angle and flexion at initial contact.
- Peak knee valgus angle (PKVA) and flexion (PKFA).
- Knee valgus and flexion ROM angles between initial contact and peak values.
- Peak hip angles (in sagittal, frontal and transverse planes) in the six phases IC, PVGRF, PEKAM, 60 ms, PKVA and PKFA.
- Hip ROM angles (in sagittal, frontal and transverse planes) between IC and PVGRF, IC and PEKAM, IC and 60 ms, IC and PKVA and IC and PKFA.

4.4.3 Statistical analysis

The statistical analyses were carried out using Statistical Package for Social Sciences software (version 21, SPSS Statistics 20.Ink). Normality for each variable was checked with a Shapiro-Wilk test and histograms to check whether the data were normally distributed or not (parametric or non-parametric). Studies have shown that the Shapiro-Wilk test is the best option for testing normality due to its high power (Field, 2013; Ghasemi & Zahediasl, 2012; Razali & Wah, 2011). The level of significance was set at 0.05. The mean value of five trials of each test were calculated to find the differences limbs. For parametric variables, a paired t-test was used and for non-parametric variables a Wilcoxon Rank test was used (Edwards, Steele, Cook, Purdam, & McGhee, 2012) to examine the differences between limbs (preferred and non-preferred).

When you run multiple tests, the p-values have to be adjusted to control the family-wise error rate (Type I error rate) (Dmitrienko, Tamhane, & Bretz, 2009). In order to address this issue, researchers have developed various correction methods. For instance, Bonferroni's correction method is one of the traditional methods for multiple comparison correction, and divides the significance level (e.g., $p < .05$) by the number of tests performed (Bland & Altman, 1995). Although the Bonferroni correction produces a good control of Type I errors, it has the disadvantage of being too conservative especially when many comparisons are being tested; thus, the likelihood of type II errors increases, so that truly important differences are deemed non-significant. To address this issue, many researchers use a Holm correction method, to control the family-wise error rate. However, this is less conservative than the Bonferroni correction (Chen, Feng, & Yi, 2017; Holm, 1979; Lee & Lee, 2018). Therefore, p-values were corrected using the Holm method correction ($\alpha = 0.05/ (41 \text{ comparisons} - \text{rank} + 1)$).

Also, descriptive analysis (mean and standard deviation (SD)) was done for each dependent variable in each COD manoeuvre and MVIC test. Effect sizes were determined using the Cohen δ method (Thomas et al., 2015), which defines 0.2, 0.5 and 0.8 as small, moderate and large respectively.

4.5 **Result**

36 participants took part in this study. The demographic characteristics of the participants are summarised in table 4-1. All participants completed five successful COD manoeuvres to 90° and 135° and MVIC trials using their preferred and non-preferred limb. Participants were physically active, free from lower extremity injury, and had no history of lower extremity surgery. Participants were male recreational soccer players who frequently performing COD manoeuvre to 90° and 135°. 31 participants preferred to COD using right limb while 5 subjects using left limb. Valid trials for each were collected if the participants' limb landed on force plates. The results were collected as a difference between the preferred and non-preferred limbs during 90° and 135° COD manoeuvres and MVIC test.

Table 4-1 Participant demographics.

Demographic	Mean	SD
Age (years)	24.25	6.21
Height (m)	1.72	0.06
Mass (kg)	66.41	10.83
BMI (kg/m²)	19.28	2.89

Standard deviation (SD); metre (m); kilogramme (kg); body mass index (BMI); kilogramme per square meter (kg/m²).

4.5.1 Biomechanical differences between preferred and non-preferred limbs during 90° COD manoeuvres:

The normality test (Shapiro-Wilk) for kinetic and kinematic variables revealed that all variables were normally distributed for both limbs. A paired samples t-test was used to determine whether there was a statistically significant mean difference between the preferred and non-preferred limbs when participants performed 90° COD manoeuvre.

Figure 4-2 shows sagittal, transverse and frontal planes hip joint angles, sagittal and frontal plane knee joint angles and moment and time normalised force-time curves for preferred and non-preferred limb during 90° COD manoeuvres. The results of this study indicate that no biomechanical differences exist between preferred and non-preferred limbs during 90° COD manoeuvres (Table 4-2). The joint angle profiles reveal a similar peak knee valgus angle for both preferred and non-preferred limbs during 90° COD manoeuvre. No significant difference was found for peak VGRF and external knee abduction moment between the preferred and non-preferred limbs.

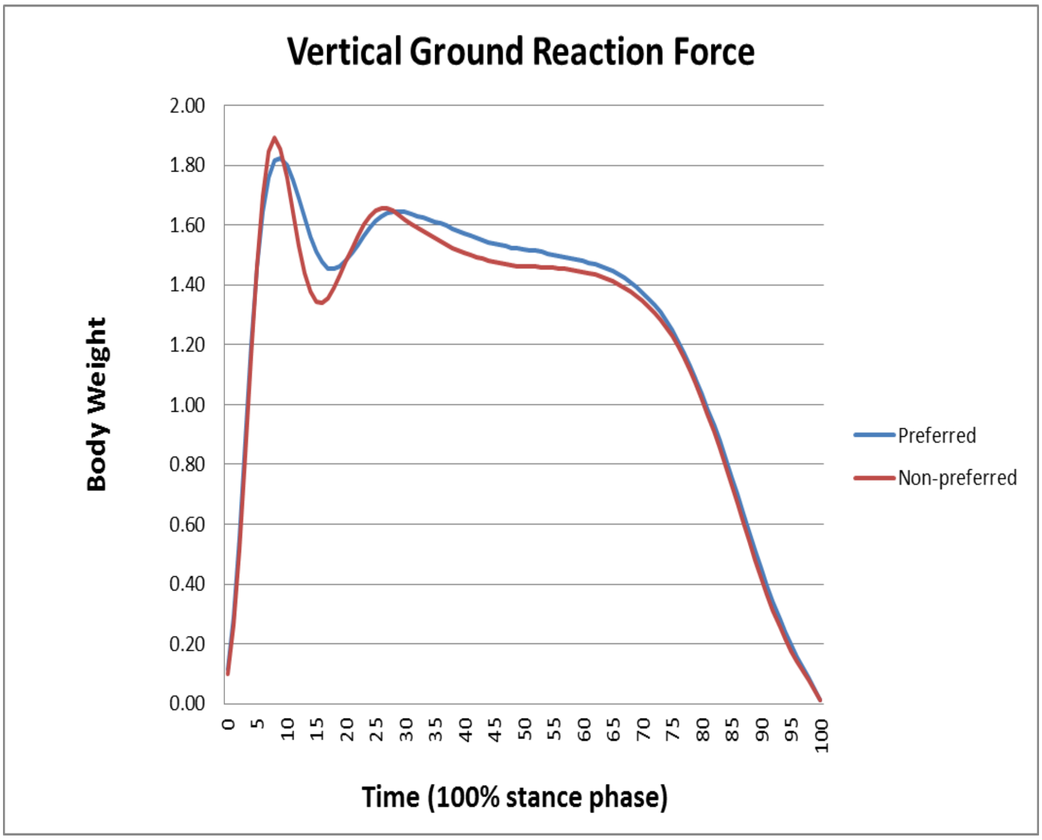
Table 4-2 Comparisons (mean ± SD) of hip and knee angles (degrees) and moments (Nm/kg) of preferred and non-preferred limb during 90° COD manoeuvre.

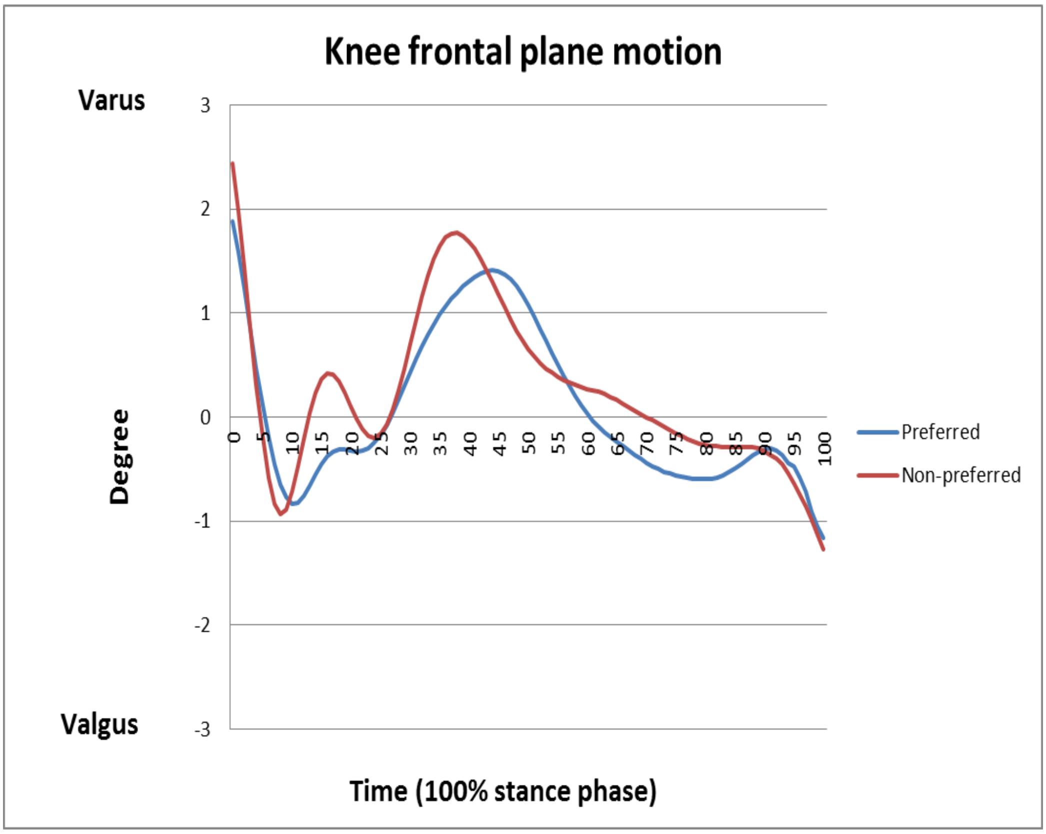
Variable	Preferred limb		Non-preferred limb		Raw p-value	Adjusted p-value[a]	ES
	Mean	SD	Mean	SD			
Kinematics							
Peak hip sagittal plane angle (°) at							
IC	39.67	7.39	42.77	8.13	0.01	0.001	0.47
PEKAM	41.44	7.44	44.10	7.97	0.02	0.001	0.41
PVGRF	42.28	7.56	45.06	8.22	0.02	0.001	0.43
PKVA	44.33	8.73	46.65	9.41	0.07	0.002	0.31
60 ms	46.06	8.36	49.29	9.81	0.02	0.001	0.42
PKFA	47.72	9.37	51.82	11.10	0.01	0.001	0.47
Hip sagittal ROM angle (°) between IC and							

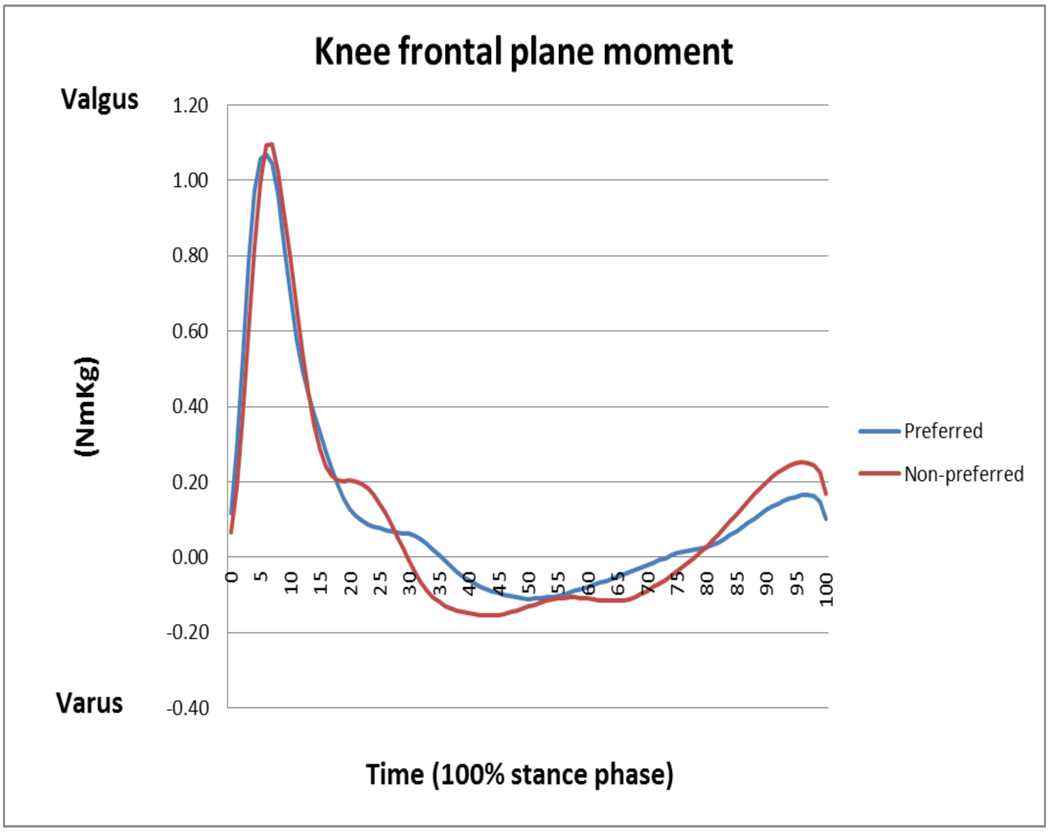
PEKAM	1.77	1.84	1.33	1.43	0.10	0.002	0.19
PVGRF	2.61	2.35	2.29	2.03	0.13	0.002	0.18
PKVA	4.66	4.45	3.88	4.56	0.40	0.006	0.10
60 ms	6.39	4.37	6.52	3.74	0.82	0.050	0.04
PKFA	8.05	5.39	9.05	5.45	0.21	0.013	0.07
Peak hip frontal angle (°) at							
IC	-20.45	6.61	-18.31	7.26	0.09	0.002	0.30
PEKAM	-19.42	6.65	-17.33	7.01	0.08	0.002	0.28
PVGRF	-19.21	6.67	-17.15	7.04	0.09	0.002	0.27
PKVA	-17.97	6.83	-16.50	6.89	0.06	0.003	0.19
60 ms	-18.09	6.89	-16.21	6.95	0.09	0.002	0.24
PKFA	-15.92	6.98	-14.32	7.09	0.42	0.003	0.21
Hip frontal ROM angle (°) between IC and							
PEKAM	1.04	1.05	0.99	1.31	0.38	0.005	0.10
PVGRF	1.24	1.24	1.16	1.54	0.44	0.007	0.09
PKVA	2.49	2.29	1.81	2.12	0.01	0.001	0.29
60 ms	2.36	1.96	2.10	2.58	0.11	0.002	0.19
PKFA	4.53	3.75	3.99	4.47	0.12	0.002	0.18
Peak hip transverse plane angle (°) at							
IC	4.83	9.18	5.94	8.60	0.48	0.010	0.12
PEKAM	-0.71	9.01	1.17	9.09	0.51	0.003	0.20
PVGRF	-1.42	9.02	0.10	9.31	0.34	0.004	0.16
PKVA	-6.82	10.72	-2.97	9.58	0.04	0.001	0.36
60 ms	-4.20	9.01	-1.91	8.83	0.15	0.002	0.24
PKFA	-11.03	9.62	-7.33	8.12	0.01	0.001	0.44
Hip transverse rotation ROM angle (°) between IC and							
PEKAM	-5.54	4.21	-4.77	3.19	0.75	0.004	0.04
PVGRF	-6.25	4.49	-5.83	3.32	0.65	0.025	0.08
PKVA	-11.65	7.24	-8.91	6.75	0.03	0.002	0.30
60 ms	-9.03	5.53	-7.85	4.62	0.30	0.003	0.18
PKFA	-15.86	6.42	-13.27	6.54	0.04	0.001	0.36
Knee frontal plane angle (°)							
IC	1.89	3.92	2.44	4.04	0.42	0.006	0.14
Peak	-4.20	5.00	-3.16	4.87	0.21	0.003	0.21
ROM	-6.09	3.84	-5.60	2.71	0.47	0.008	0.12
Knee sagittal plane angle (°)							
IC	17.87	6.32	20.06	5.82	0.01	0.001	0.47
Peak	61.61	8.09	65.94	7.09	0.0012	0.0011	0.59
ROM	43.74	7.09	45.88	6.84	0.06	0.001	0.32
Moments							
PVGRF (*BW)	2.15	0.31	2.07	0.37	0.11	0.002	0.28
PEKAM (Nm/Kg)	1.23	0.57	1.33	0.70	0.32	0.004	0.17

* The mean difference is significant.
Standard deviation (SD); effect size (ES); body weight (BW); angle (°); peak external knee abduction moment (PEKAM); newton meter per kilogram (Nm/Kg); peak vertical ground reaction force (PVGRF); initial contact (IC); peak knee valgus angle (PKVA); millisecond (ms); peak knee flexion angle (PKFA).
Sign conventions shows the position of the joints as; hip flexion (+), hip Extension (-), hip abduction (-), hip adduction (+), hip internal rotation (+) and hip external rotation (-).
[a] By holm method.

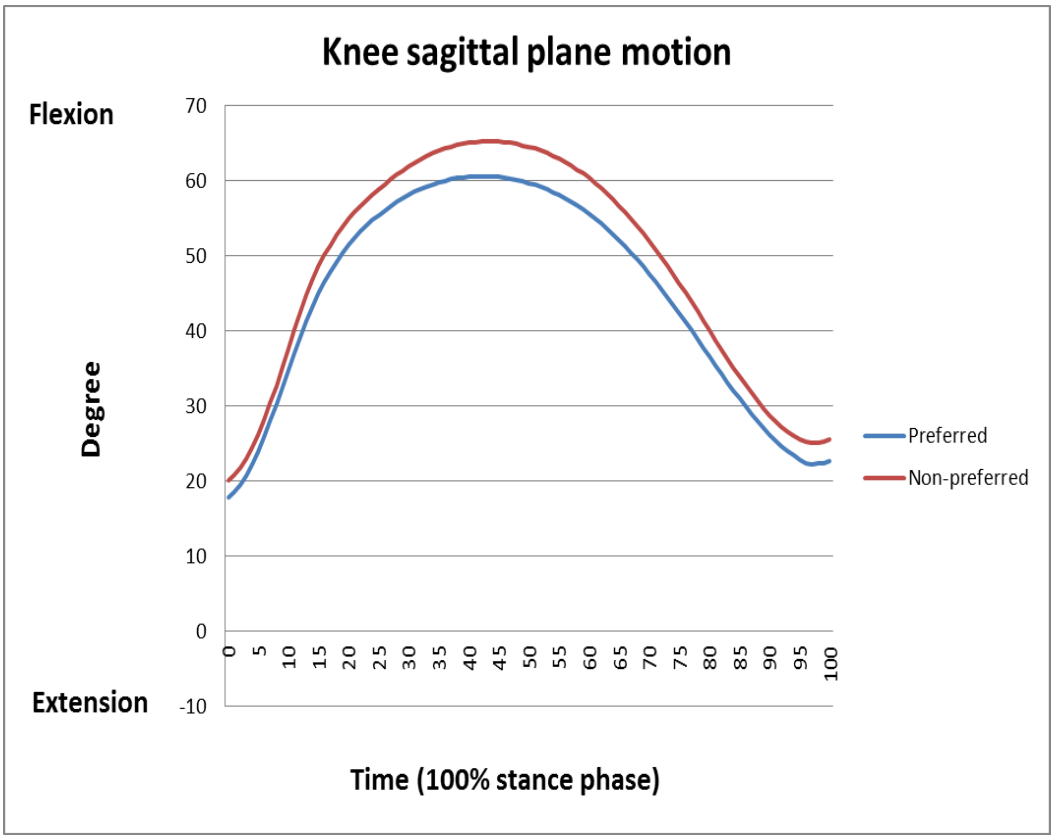
Figure 4-2 Averages of hip and knee angles, external knee abduction moment and vertical force time curves during the stance phase for preferred and non-preferred limbs during 90° COD manoeuvre (n=36), X-axis is percentage of stance phase.



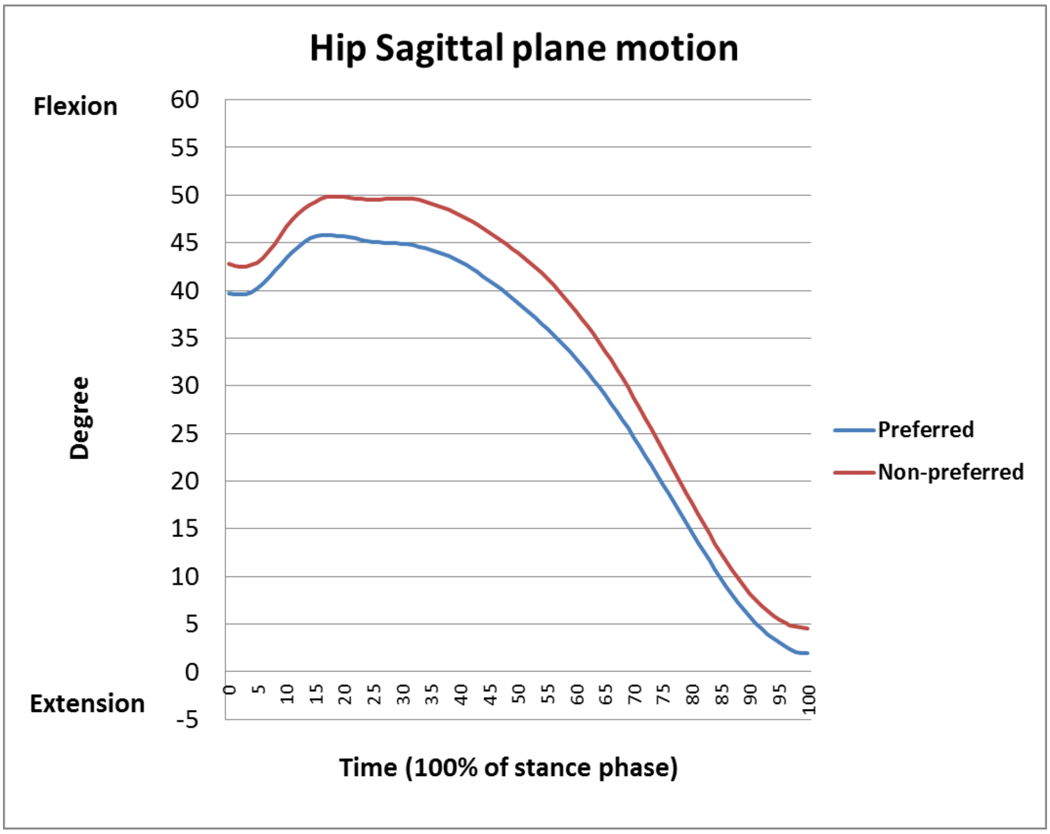


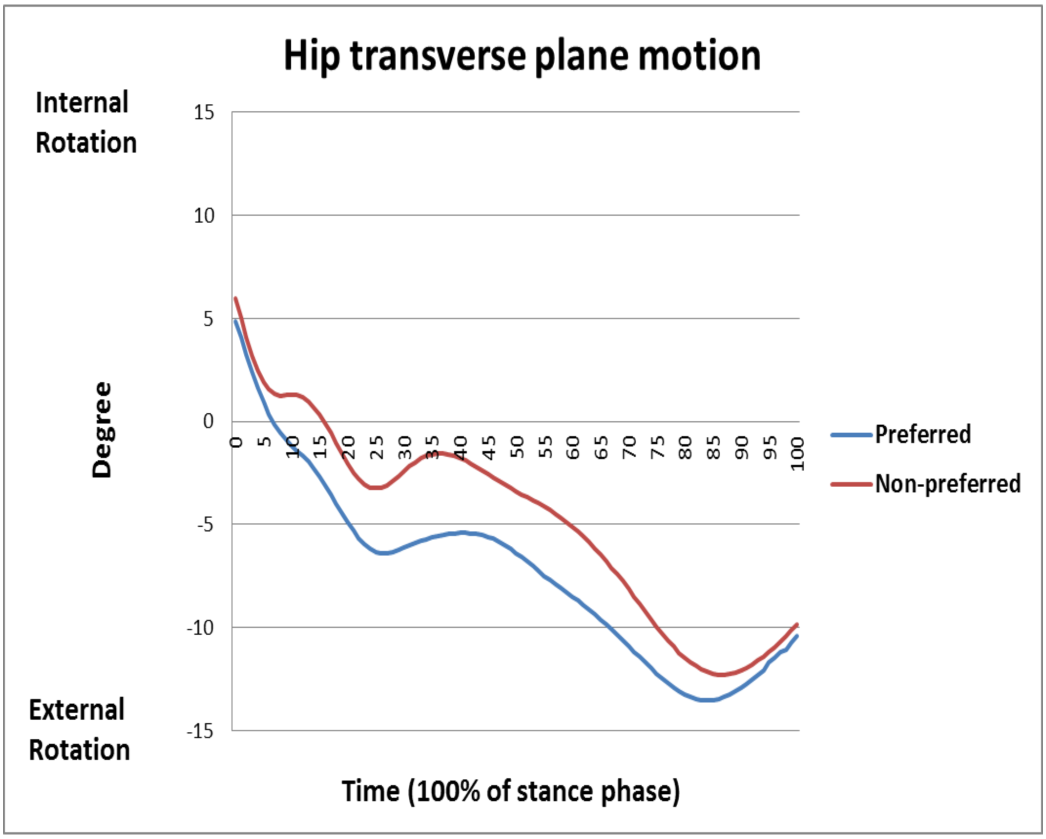


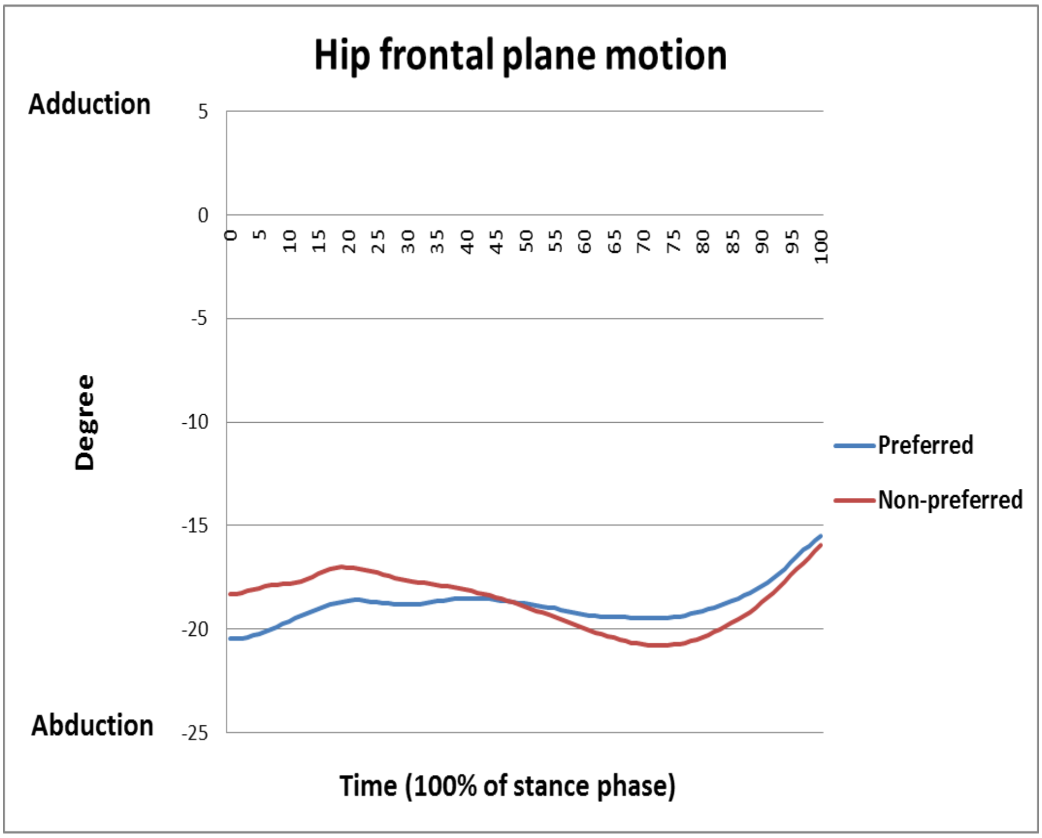
Knee frontal plane moment
Valgus
Varus
(NmKg)
1.20
1.00
0.80
0.60
0.40
0.20
0.00
-0.20
-0.40
0
5
10
15
20
25
30
35
40
45
50
55
60
65
70
75
80
85
90
95
100
Preferred
Non-preferred
Time (100% stance phase)



Knee sagittal plane motion
Flexion
Extension
Degree
70
60
50
40
30
20
10
0
-10
0
5
10
15
20
25
30
35
40
45
50
55
60
65
70
75
80
85
90
95
100
Preferred
Non-preferred
Time (100% stance phase)


Hip Sagittal plane motion
Flexion
Extension
Degree
60
55
50
45
40
35
30
25
20
15
10
5
0
-5
0 5 10 15 20 25 30 35 40 45 50 55 60 65 70 75 80 85 90 95 100
Time (100% of stance phase)
Preferred
Non-preferred


Hip transverse plane motion
Internal Rotation
External Rotation
Degree
15
10
5
0
-5
-10
-15
0 5 10 15 20 25 30 35 40 45 50 55 60 65 70 75 80 85 90 95 100
Time (100% of stance phase)
Preferred
Non-preferred



4.5.2 Biomechanical differences between preferred and non-preferred limbs during 135° COD manoeuvres:

The normality test (Shapiro-Wilk) for kinetic and kinematic variables revealed that all variables were normally distributed for both limbs. Figure 4-3 shows sagittal, transverse and frontal planes hip joint angles, sagittal and frontal plane knee joint angles and moment and time normalised force-time curves for preferred and non-preferred limb during 135° COD manoeuvres. The results of this study indicate that no biomechanical differences exist between preferred and non-preferred limbs during 135° COD manoeuvres (Table 4-3). The joint angle profiles reveal a similar peak knee valgus angle for both preferred and non-preferred limbs during 135° COD manoeuvres. The joint moment profiles reveal a similar peak VGRF and external knee abduction moment between the preferred and non-preferred limbs.

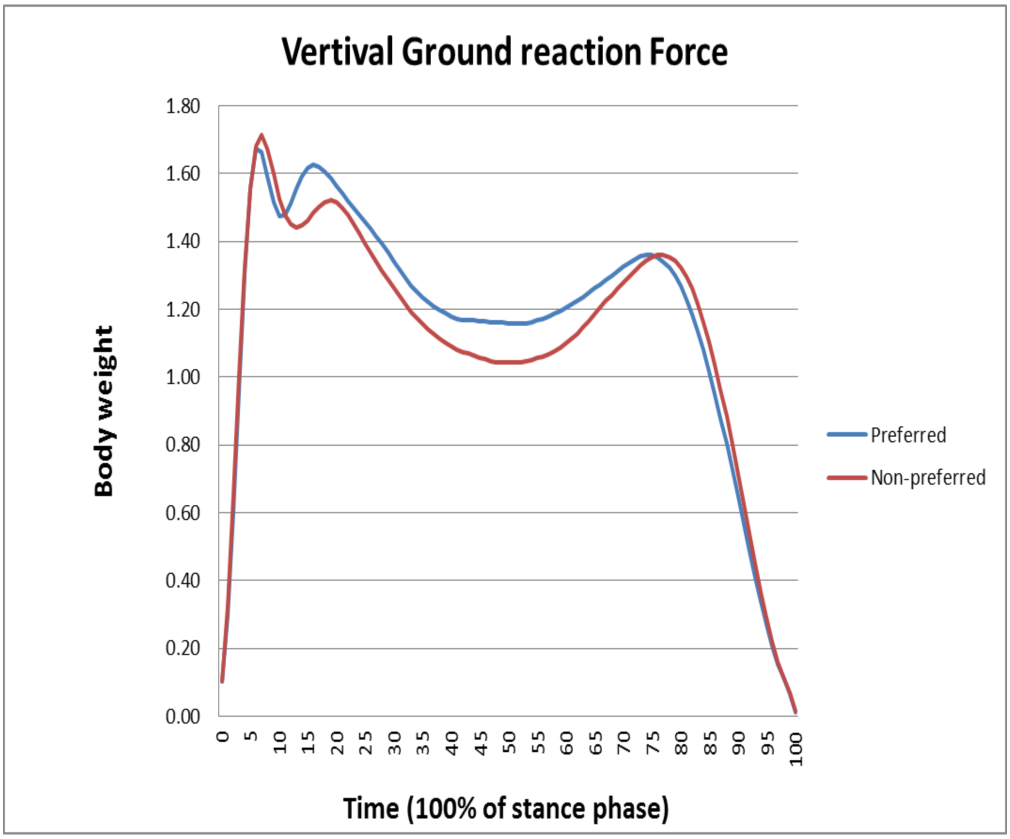
Table 4-3 Comparisons (mean ± SD) of hip and knee angles (degrees) and moments (Nm/kg) of preferred and non-preferred limb during 135° COD manoeuvre.

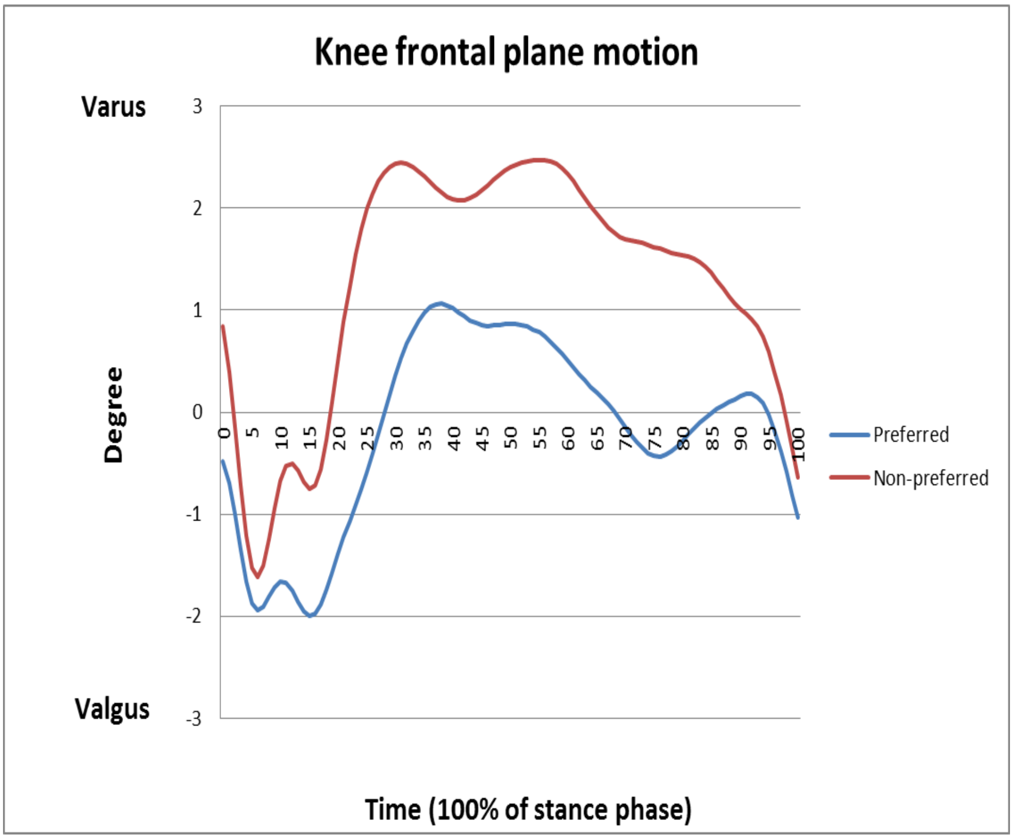
Variable	Preferred limb		Non-preferred limb		Raw p-value	Adjusted p-value[a]	ES
	Mean	SD	Mean	SD			
Kinematics							
Peak hip sagittal plane angle (°) at							
IC	37.03	8.18	40.30	7.51	0.0012	0.0011	0.59
PEKAM	39.47	8.60	42.86	7.83	0.0043	0.0012	0.51
PVGRF	39.65	8.53	43.29	7.80	0.0015	0.0011	0.57
PKVA	42.75	10.20	45.30	9.57	0.10	0.0016	0.28
60 ms	43.67	10.28	47.37	9.33	0.01	0.0013	0.46
PKFA	49.59	12.52	54.73	10.35	0.0031	0.0012	0.53
Hip sagittal ROM angle (°) between IC and							
PEKAM	2.44	2.46	2.55	2.39	1.00	0.0500	0.00
PVGRF	2.61	2.32	2.99	2.39	0.79	0.0036	0.03
PKVA	5.72	5.66	5.00	5.59	0.43	0.0019	0.09
60 ms	6.64	4.87	7.07	4.45	0.67	0.0025	0.07
PKFA	12.55	7.30	14.42	6.78	0.13	0.0017	0.26
Peak hip frontal angle (°) at							
IC	-21.94	7.56	-21.88	6.81	0.97	0.0125	0.01
PEKAM	-20.79	7.77	-20.78	6.58	1.00	0.0250	0.00
PVGRF	-20.77	7.81	-20.72	6.64	0.97	0.0100	0.01
PKVA	-19.60	8.15	-20.18	6.42	0.67	0.0026	0.07
60 ms	-20.23	7.89	-20.29	6.66	0.96	0.0083	0.01
PKFA	-17.04	7.97	-17.12	6.31	0.96	0.0063	0.01
Hip frontal ROM angle (°) between IC and							
PEKAM	1.15	1.44	1.10	1.18	0.93	0.0056	0.01
PVGRF	1.16	1.49	1.17	1.24	0.75	0.0031	0.04
PKVA	2.33	3.37	1.71	1.79	0.50	0.0019	0.08
60 ms	1.71	2.21	1.60	1.73	0.88	0.0045	0.02
PKFA	4.89	4.93	4.76	4.66	0.65	0.0022	0.05
Peak hip transverse plane angle (°) at							
IC	7.82	9.04	6.94	10.02	0.58	0.0020	0.09
PEKAM	0.13	8.71	0.06	9.93	0.96	0.0071	0.01
PVGRF	-0.17	8.68	-0.37	10.29	0.89	0.0050	0.02
PKVA	-6.73	11.02	-3.28	9.24	0.03	0.0014	0.38
60 ms	-2.30	9.43	-1.59	9.59	0.65	0.0024	0.08
PKFA	-9.98	9.23	-7.08	8.17	0.04	0.0014	0.35
Hip transverse rotation ROM angle (°) between IC and							
PEKAM	-7.69	6.30	-6.88	4.67	0.65	0.0023	0.05
PVGRF	-7.99	6.35	-7.31	4.77	0.62	0.0021	0.06
PKVA	-14.55	7.78	-10.22	6.82	0.0013	0.0011	0.58
60 ms	-10.12	6.41	-8.53	4.87	0.19	0.0018	0.22
PKFA	-17.80	7.26	-14.02	6.65	0.0049	0.0012	0.33
Knee frontal plane angle (°)							
IC	-0.48	4.26	0.84	3.90	0.08	0.0015	0.32
Peak	-5.87	5.78	-4.53	5.02	0.08	0.0015	0.30

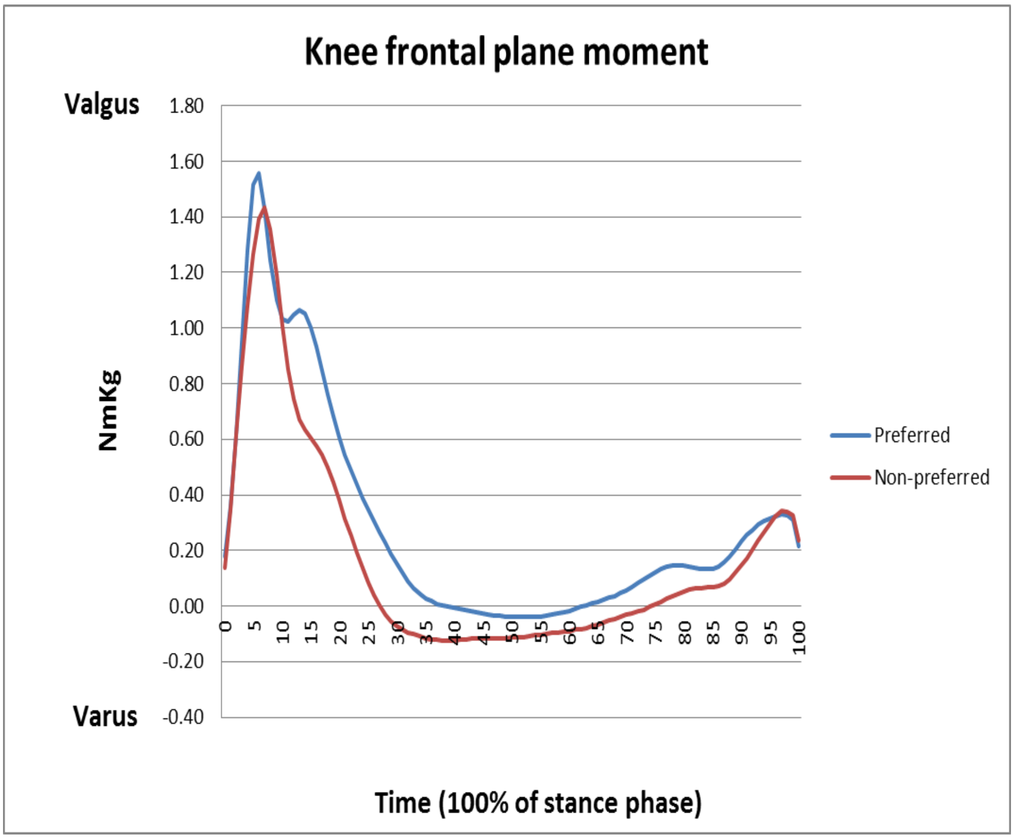
ROM	-5.39	4.07	-5.37	3.39	0.98	0.0167	0.01
Knee sagittal plane angle (°)							
IC	18.86	4.97	19.16	5.41	0.73	0.0029	0.06
Peak	63.02	8.83	67.26	7.74	0.01	0.0013	0.46
ROM	44.17	7.90	48.10	7.25	0.01	0.0013	0.45
Moments							
PVGRF (*BW)	2.16	0.35	2.07	0.36	0.09	0.0016	0.29
PEKAM (Nm/Kg)	2.34	1.11	2.04	1.11	0.03	0.0014	0.37

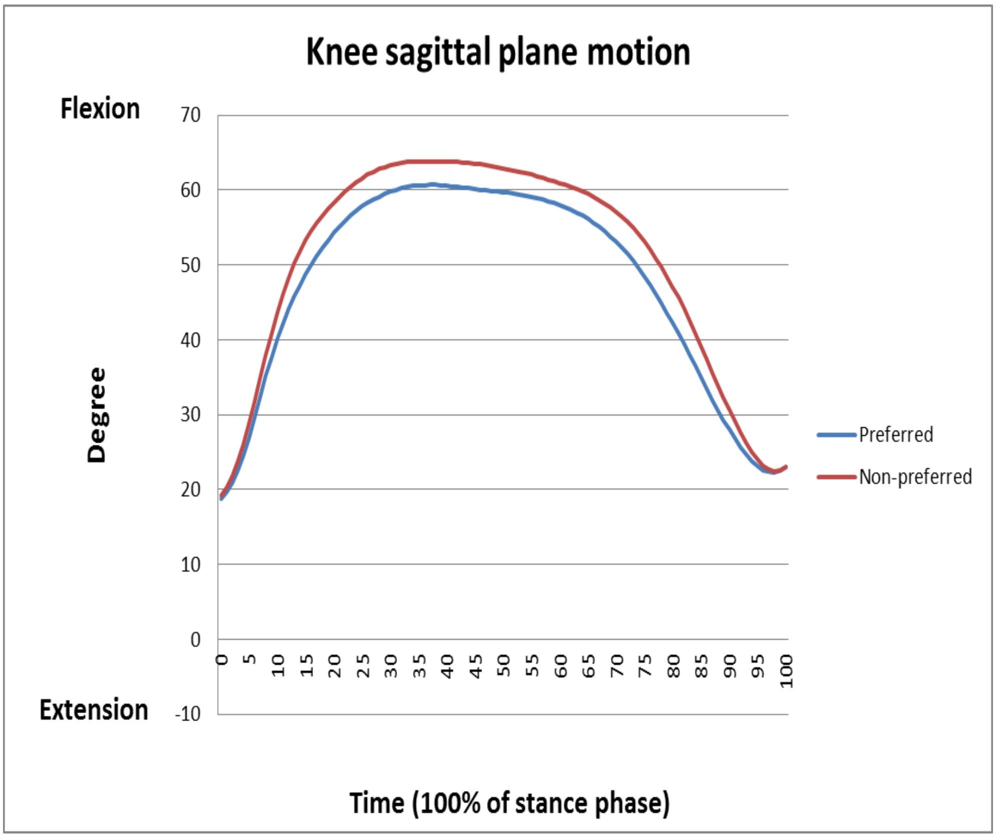
* The mean difference is significant.
Standard deviation (SD); effect size (ES); body weight (BW); angle (°); peak external knee abduction moment (PEKAM); newton meter per kilogram (Nm/Kg); peak vertical ground reaction force (PVGRF); initial contact (IC); peak knee valgus angle (PKVA); millisecond (ms); peak knee flexion angle (PKFA).
Sign conventions shows the position of the joints as; hip flexion (+), hip Extension (-), hip abduction (-), hip adduction (+), hip internal rotation (+) and hip external rotation (-).
[a] By holm method.

Figure 4-3 Averages of hip and knee angles, external knee abduction moment and vertical force time curves during the stance phase for preferred and non-preferred limbs during 135° COD manoeuvre (n=36), X-axis is percentage of stance phase.

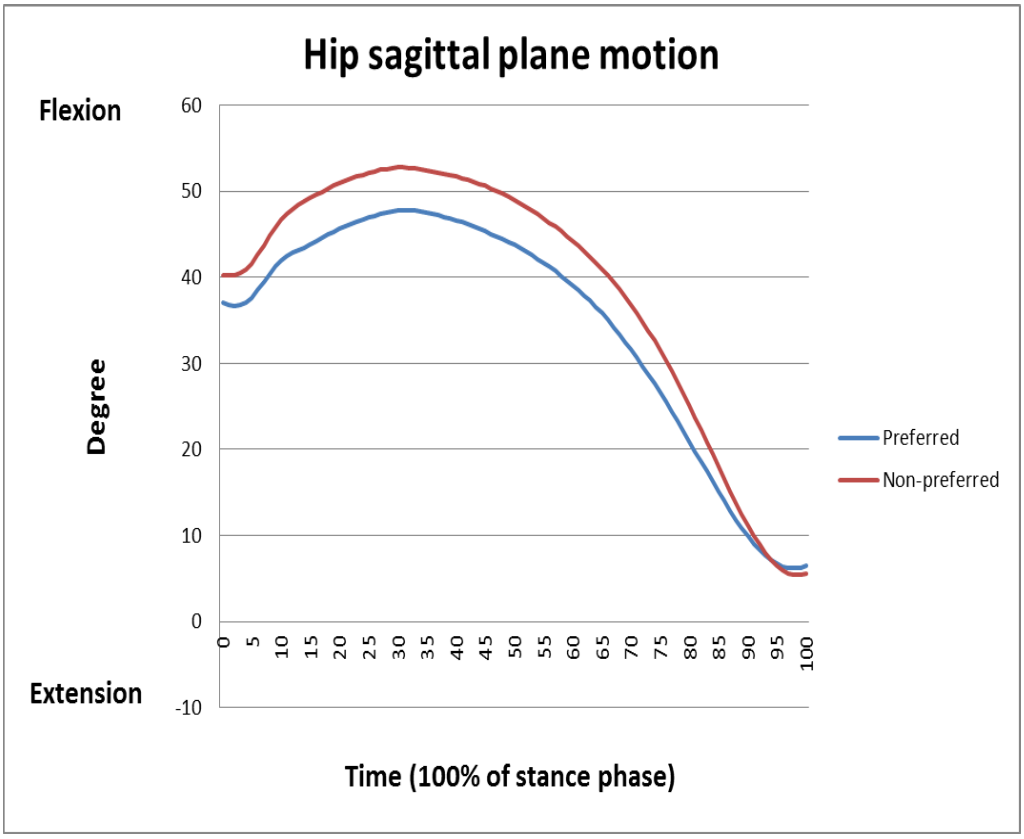



Knee frontal plane motion
Varus
Valgus
Degree
3
2
1
0
-1
-2
-3
0 5 10 15 20 25 30 35 40 45 50 55 60 65 70 75 80 85 90 95 100
Preferred
Non-preferred
Time (100% of stance phase)

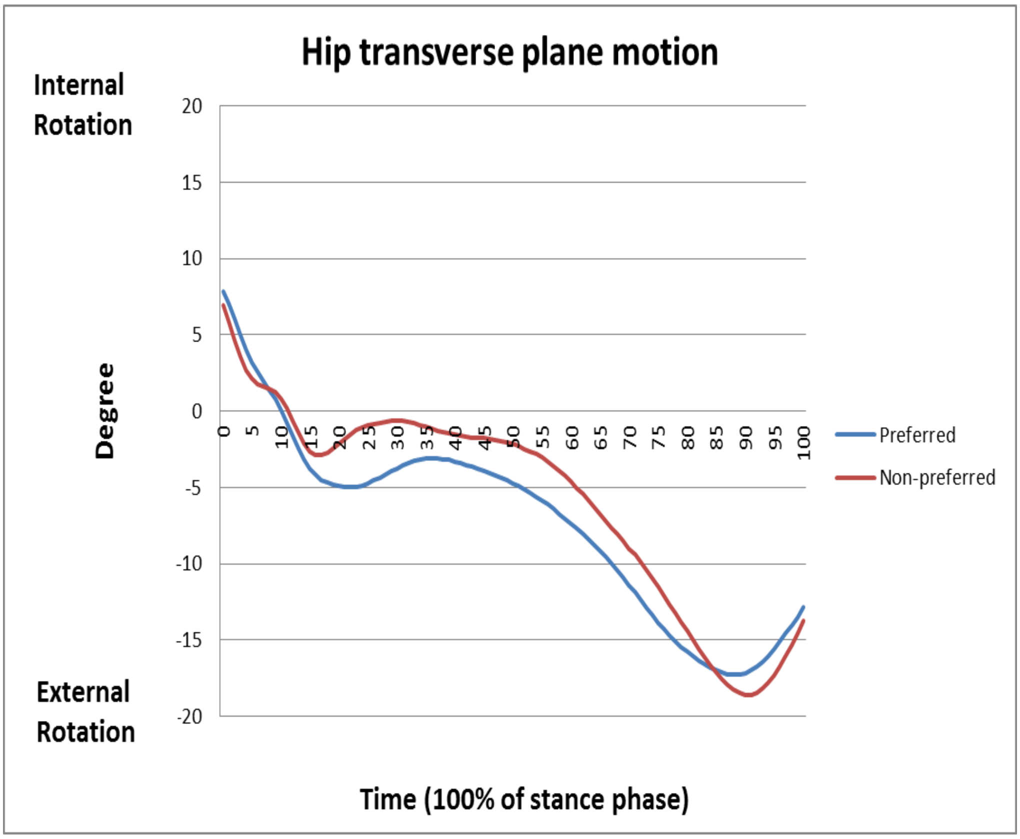

Knee frontal plane moment
Valgus
Varus
NmKg
1.80
1.60
1.40
1.20
1.00
0.80
0.60
0.40
0.20
0.00
-0.20
-0.40
0 5 10 15 20 25 30 35 40 45 50 55 60 65 70 75 80 85 90 95 100
Preferred
Non-preferred
Time (100% of stance phase)

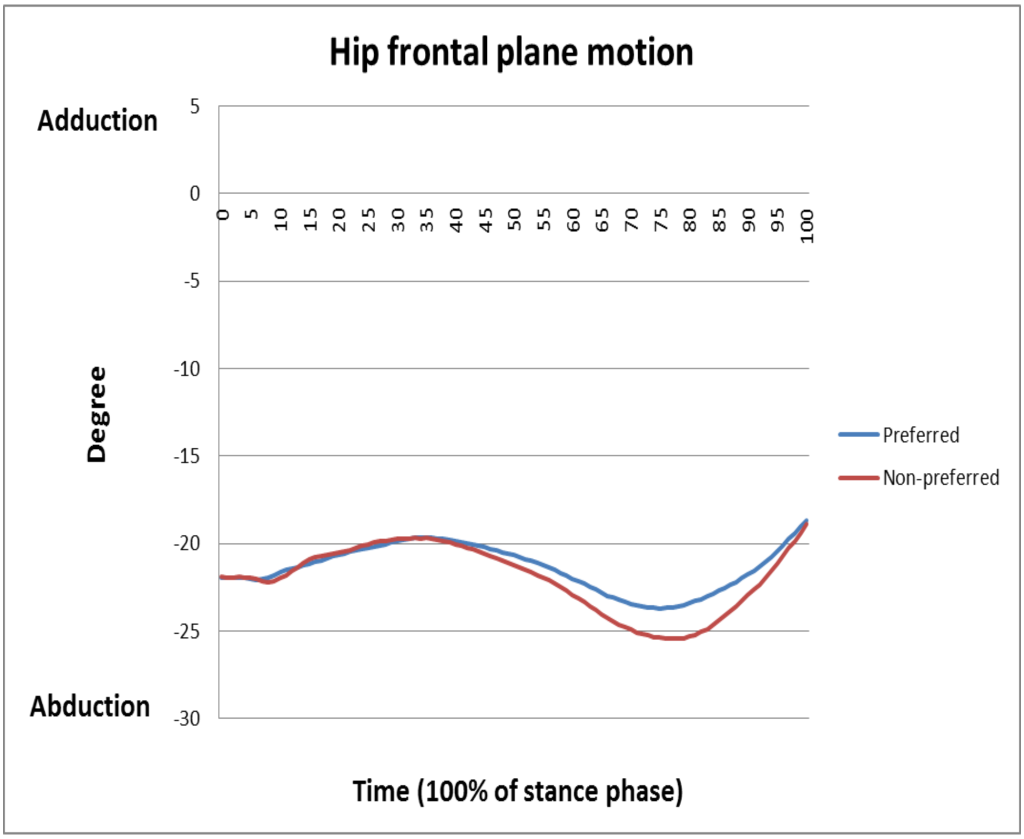


Knee sagittal plane motion
Flexion
Extension
Degree
70
60
50
40
30
20
10
0
-10
0
5
10
15
20
25
30
35
40
45
50
55
60
65
70
75
80
85
90
95
100
Preferred
Non-preferred
Time (100% of stance phase)



Hip sagittal plane motion
Flexion
Extension
Degree
60
50
40
30
20
10
0
-10
0
5
10
15
20
25
30
35
40
45
50
55
60
65
70
75
80
85
90
95
100
Preferred
Non-preferred
Time (100% of stance phase)


Hip transverse plane motion
Internal Rotation
External Rotation
Degree
20
15
10
5
0
-5
-10
-15
-20
0 5 10 15 20 25 30 35 40 45 50 55 60 65 70 75 80 85 90 95 100
Time (100% of stance phase)
Preferred
Non-preferred


Hip frontal plane motion
Adduction
Abduction
Degree
5
0
-5
-10
-15
-20
-25
-30
0 5 10 15 20 25 30 35 40 45 50 55 60 65 70 75 80 85 90 95 100
Time (100% of stance phase)
Preferred
Non-preferred

4.5.3 Hip muscle strength differences between preferred and non-preferred limbs during MVIC test:

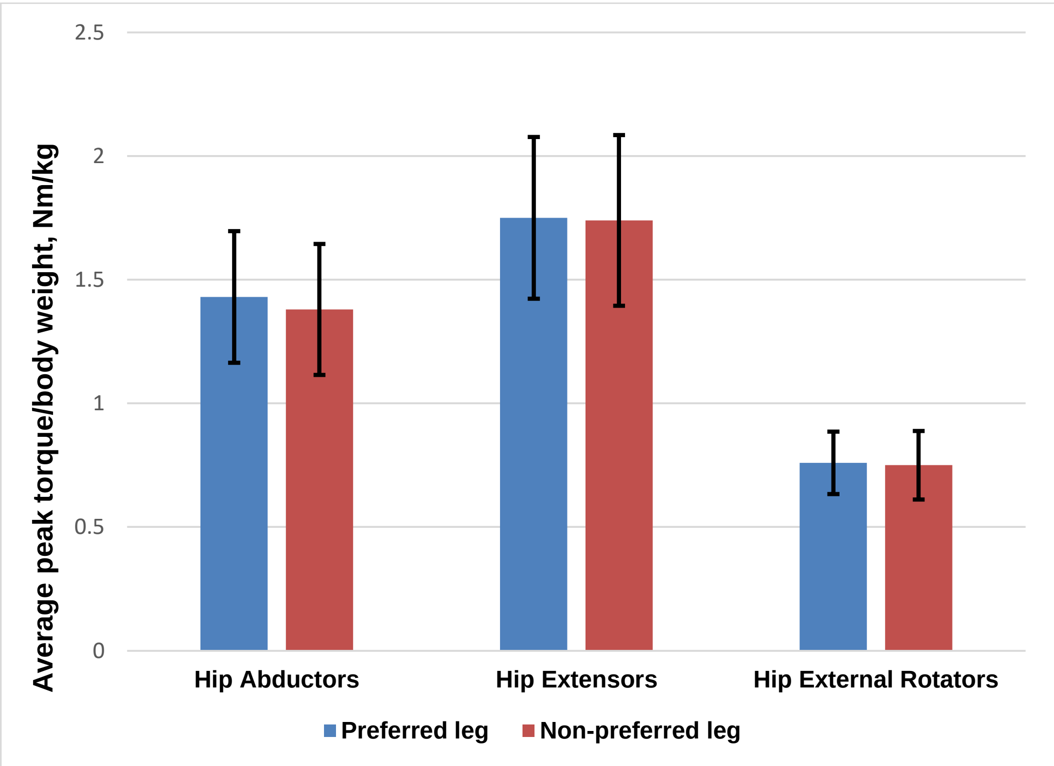
The normality test (Shapiro-Wilk) for normalised peak torque revealed that all variables were normally distributed for both limbs. Figure 4-4 shows normalised peak torque of the hip abductors, extensors, and external rotators muscles for preferred and non-preferred limb during MVIC test. The results of this study indicate that no hip strength differences exist between preferred and non-preferred limbs during MVIC test with small effect size (Table 4-4).

Table 4-4 Comparisons (mean ± SD) of normalised peak torque of hip muscles (Nm/Kg) of preferred and non-preferred limb during MVIC test.

Variable	Preferred limb		Non-preferred limb		P value	ES
Nm/Kg	Mean	SD	Mean	SD		
Hip Abductors	1.43	0.27	1.38	0.26	0.16	0.24
Hip Extensors	1.75	0.33	1.74	0.34	0.76	0.05
Hip External Rotators	0.76	0.13	0.75	0.14	0.47	0.12

* The mean difference is significant.
Standard deviation (SD); effect size (ES); newton meter per kilogram (Nm/Kg).

Figure 4-4 normalised peak torque of the hip abductors, extensors, and external rotators muscles for preferred and non-preferred limb during MVIC test (n=36).



4.6 Discussion

The main objectives of the study were to examine biomechanical and strength differences in limb preference of recreational healthy male soccer participants during COD manoeuvres at 90° and 135° angles and MVIC test. In the present investigation, the results of this study indicate that no biomechanical and strength differences exist between preferred and non-preferred limbs during 90° and 135° COD manoeuvres and MVIC test. Recreational healthy male soccer players show similar movement patterns between the preferred and non-preferred limb while performing during 90° and 135° COD manoeuvres. It was also noted that normalised peak torque of hip muscles was similar between limbs.

Understanding sports-specific biomechanics is essential to injury prevention and post-injury treatment, as 75% of ACL injury occur during COD manoeuvre (Johnston et al., 2018; Montgomery et al., 2018). Surprisingly, to date, no study has existed on the limb asymmetry during COD manoeuvre at 90° and 135° degree. While few studies have explored lower extremity biomechanical differences between dominant and non-dominant during COD manoeuvre at 45° (Bencke et al., 2013; Greska, Cortes, Ringleb, Onate, & Van Lunen, 2016; Pollard et al., 2018), this investigation is the first to examine the differences between the preferred and non-preferred limbs during COD manoeuvre at 90° and 135° degree. After analysing hip and knee kinematics and kinetics during a COD manoeuvre, almost all results support limb symmetry during this manoeuvre. Other studies have investigated differences between dominant and non-dominant limbs during COD to 45° manoeuvres (Bencke et al., 2013; Brown et al., 2014b; Greska et al., 2016; Pollard et al., 2018) and during COD to 105° (Marshall et al., 2015), of which several similarities exist when compared to the current study.

The average peak external knee abduction moment and peak knee abduction angle showed no differences between limbs during a COD manoeuvre at 90° and 135° degree. These results are consistent with previous research that demonstrates a lack of differences between dominant and non-dominant limbs (Bencke et al., 2013; Greska et al., 2016; Marshall et al., 2015; Pollard et al., 2018). Differences in peak external knee abduction moment and peak knee abduction angle were noted between the previous studies and the current study. Greska et al. (2016) found that peak Knee abduction angles were greater in the non-preferred limb compare with preferred limb (-5.9° and -4.7°, respectively). In contrast, the peak Knee abduction angles in the current study were greater in the preferred limb compare with non-preferred limb (-5.7° and -4.53°, respectively). Similary to current study, Pollard et al. (2018) and Marshall et al. (2015)

found that peak Knee abduction angles were greater in the preferred limb compare with non-preferred limb (-4.4° and -3.5°, -7.5 and -6.1 respectively). Furthermore, Greska et al. (2016) found that peak external Knee abduction moments were 0.3 and 0.2 Nm/Kg in the preferred and non-preferred limbs respectively. Bencke et al. (2013) found that peak external Knee abduction moments were 0.7 Nm/Kg in both limbs. In contrast, the peak external Knee abduction moments in the current study were 1.23 and 1.33 Nm/Kg during COD to 90° angle and 2.34 and 2.04 Nm/Kg during COD to 135° angle in the preferred and non-preferred limbs respectively. Similary to current study, Marshall et al. (2015) found that peak external Knee abduction moments were 2.5 and 2.3 Nm/Kg in the preferred and non-preferred limbs respectively during COD to 105° angle. In contrast to the current study and previous studies results, only one study discovered differences between the dominant and non-dominant limbs in healthy female during planned COD to 45° angle with regard to knee flexion angle, knee abduction angle, knee internal rotation angle and knee abductor moment (Brown et al., 2014b).

The differences between current study findings and the above mention studies can be attributed to the several factors. The previous studies investigated the differences between limbs during COD manoeuvre at 45°, while the current study used different COD angles (90° and 135° degree). In addition, the approach velocity in the previous studies was slower compare to the current study. Knee kinetics and kinematics ca contribute to increase speed and angel of the COD manoeuvre, which sharper angle and increase the velocity affect knee biomechanics (Imwalle et al., 2009; Vanrenterghem et al., 2012). In addition, the previous studies used different participants (female). Furthermore, all the previous studies have defined the preferred limb as the preferred to kick the ball with. However, in the current study the preferred limb was determined by asking participants which limb they would preferred to use as the push-off limb during COD manoeuvre. Moreover, previous studies investigating biomechanical symmetry in COD manoeuvre have typically done so using discrete points (e.g. peak values). However, there are a number of limitations with this type of analysis. First, asymmetry may occur over phases that are not captured in a single data point. Second, the discrete points utilised may vary between studies. Use a discrete point analysis alone may not detect all significant asymmetries. In contrast, in the current study the biomechanical analysis symmetry done using multiphases and discrete points.

Peak knee flexion angle was close to reaching the level of significance during the COD manoeuvre at 90°, the impact may not be important since previous studies (Krosshaug et al., 2007b;Koga et al., 2010) have shown that non-contact ACL injuries occur quickly after ground initial contact. These studies have demonstrated that a non-contact ACL injury occurs about 40 ms after initial contact (Krosshaug et al., 2007b; Koga et al., 2010). Within the current study, the peak knee flexion angle occurred at 42% of the stance phase (165 ms), for both the preferred and non-preferred limbs, with a 4.5° difference between them. It has also been demonstrated that ACL strain is greatest at knee flexion angles less than 30° (Beynnon et al., 1992). With the current difference noted between limbs occurring at greater than 60° of knee flexion, such a difference may not produce a substantial consequence relative to non-contact ACL injury risk.

It should be noted that, although the results show that no significant differences exist between limbs, the mean differences are greater than the SEM reported previously in Chapter 3. The differences are greater than the SEM so the difference is real and not due to the measurement error. However, this is still not significant as the effect size is small, which is probably because the variability is large so there is a lot of overlap. For example, the mean knee abduction angles are -4.20 (SD 5.00) and -3.16 (SD 4.87) for the preferred and non-preferred limbs respectively. The small effect size is further evidence of the fact that the differences are not true. To conclude, although the differences were greater than the SEM reported in the reliability data, these were not significant. This could be explained by large the standard deviations showing a considerable overlap between the scores, which was why we could not find significant difference between limbs.

Some of the differences in the footstrike can occur between individuals during COD manoeuvres. This may be due to slight differences in technique and the way some individuals perform the movement with their rearfoot/forefoot. Therefore, the effect of foot placement on COD biomechanics may be a factor that needs more investigation and control. However, the effect of foot strike position differences during COD manoeuvres on knee valgus angle and moment has been explored. Yoshida et al. (2016) examined the differences in knee valgus angle between rearfoot striking participants and forefoot striking participants during 60° COD manoeuvres. The result of this study suggested that there was no effect of foot strike position differences during COD manoeuvres on the knee valgus angle. In addition, David, Komnik, Peters, Funken, and Potthast (2017) identified the effect of footstrike pattern on external knee valgus moment during COD manoeuvres. The result of this study showed that both groups (14 rearfoot striking participants and 17 forefoot striking participants) generated higher external knee valgus moment during 75° COD manoeuvres. Finally, dictating to the person how to turn is not advisable. Therefore, the participants in this study were allowed to perform the COD manoeuvres in the way that they would normally do it.

To our knowledge, this is the first study to assess hip strength between the preferred and non-preferred limbs among recreational healthy male soccer participants. The finding that no differences exist between the preferred and non-preferred limbs normalised peak torque of hip muscle during MVIC test. During MVIC hip strength test, the non-preferred limb produces similar normalised peak torque value during abduction (P = 0.16; ES = 0.24), extension (P = 0.76; ES = 0.05) and external rotation (P = 0.47; ES = 0.12).

This study provides important understanding into the limb symmetry during COD manoeuvres. However, the generalisability of the results of the current study is subject to some limitations. One limitation of the study may be that the COD manoeuvre was examined in a laboratory setting so cannot create a true natural scenario in which ACL injury most often occurs. Other limitation is the relatively small number of subjects. Finally, the participants represented a recreational healthy male soccer population without lower-limb problems. Therefore, it is not possible to generalize the results obtained to very athletic, inactive or patient populations, different population have different level skill.

4.7 Conclusion

This study is the first to provide a biomechanical and strength comparison of preferred and non-preferred limb during a COD manoeuvre at 90° and 135° degree and MVIC test. The purpose of the current study was to identify biomechanical and strength differences between preferred and non-preferred limbs during COD manoeuvres at 90° and 135° angles and MVIC test. The biomechanical and strength analysis of the COD manoeuvre and MVIC in recreational healthy male soccer players showed no differences exist between preferred and non-preferred limb. Therefore, for the following research questions only the preferred limb will be examined in chapter 5 and 6.

5 Chapter 5

Biomechanical differences between 90° and 135° change of direction manoeuvres.

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5.1 Introduction

The ability to change direction is an integral component of multidirectional sports, for example basketball, football and volleyball (Alentorn-Geli et al., 2009; Bloomfield, Polman, & O'Donoghue, 2007; Brughelli, Cronin, Levin, & Chaouachi, 2008; Karcher & Buchheit, 2014; Orendurff et al., 2010). During multidirectional sport games, players frequently have to change direction to different angles (Bloomfield et al., 2007; Robinson, O'Donoghue, & Wooster, 2011; Sweeting, Aughey, Cormack, & Morgan, 2017). However, change of direction (COD) is commonly associated with non-contact anterior cruciate ligament (ACL) injuries in sport (Koga et al., 2010, Olsen et al., 2004, Walden et al., 2015, Brophy et al., 2015). To perform a directional change, the player must first decelerate before redirecting the body in the new direction, and then accelerating (Hase & Stein, 1999). Moreover, deceleration and acceleration movements with a rapid COD causes a larger knee valgus angle and moment, as well as lower knee flexion, which are potential risk factors for non-contact ACL ruptures (Brophy et al., 2015; Grassi et al., 2017; Johnston et al., 2018; Koga et al., 2010; Montgomery et al., 2018; Olsen et al., 2004; Walden et al., 2015).

70-85% of ACL injuries can be described as non-contact (not as a result of a direct blow to the knee) (Benis et al., 2018; Johnston et al., 2018; Montgomery et al., 2018). It is often possible to prevent non-contact injuries, which is essential when considering that these are the most common type of ACL injury, therefore, the mechanism of such injuries needs to be better understood to lower the risk of injury. Mechanisms of non-contact ACL injury normally involve multi-planar knee loading events (Kiapour et al., 2014; Quatman et al., 2010). Almost 60-70 % of non-contact ACL injuries occurred while a player performed a COD manoeuvre (Kiapour et al., 2014; Quatman et al., 2010). This is of concern, as premier league football players carried out 727 COD on average to different angles during a 90 minute game, approximately eight CODs every minute (Bloomfield et al., 2007). Although lower-limb biomechanical analyses of COD manoeuvres have focused mainly on COD manoeuvre to small angles (e.g., 45 degree), studies have suggested that COD manoeuvres to larger angle results in greater knee loading (Havens & Sigward, 2015b; Imwalle et al., 2009; Schreurs et al., 2017; Sigward et al., 2015). However, limited knowledge is available regarding COD manoeuvres performed to sharper angles (90 and 135 degrees).

Hip and knee internal rotation angles, hip flexion angle were increased when carrying out COD manoeuvres at 90° compared with COD at 45° (Havens & Sigward, 2015b; Imwalle et al., 2009). Sigward et al. (2015) found that athletes displayed greater knee valgus moments and GRFs when COD at a 110° angle compared to a 45° COD. Generally, knee valgus moment was found to be 2.4 times greater during the 110° COD, and this is in line with the study conducted by Havens and Sigward (2015b), which showed greater knee valgus moment during 90° COD compared to 45° COD. This increased knee valgus moment may be due to the knee valgus moment being influenced by the extent of the GRF, as well as the moment arm in the frontal plane (Kristianslund et al., 2014). In addition, greater knee valgus angles (Jones, Herrington, & Graham-Smith, 2016; Jones et al., 2015; Sigward et al., 2015) and greater GRF magnitudes (Havens & Sigward, 2015b; Jones et al., 2016; Sigward et al., 2015) occur with sharper CODs, although greater moment arms could occur as a result of greater hip abduction (Havens & Sigward, 2015b; Sigward et al., 2015) which is related to an increase in knee abduction moment, and this can increase ACL strain (Markolf et al., 1995; Markolf et al., 1990; Shin, Chaudhari, & Andriacchi, 2009; Withrow, Huston, Wojtys, & Ashton-Miller, 2006). Altogether, the aforementioned findings present a problem, because according to Hewett et al. (2005), with female adolescent athletes', knee abduction moments can prospectively predict non-contact ACL injury.

As far as the author is aware, differences between 90° and 135° COD manoeuvres have only been examined in one study (Schreurs et al., 2017). In this study, 13 male and 16 female athletes took part, which looked at five different types of movement: running forward, and planned COD at 45°, 90°, 135° and 180° using different speeds. The participants received verbal instructions and were encouraged to sprint at full speed from start till finish. The primary outcome variables for this study were VGRF, the knee flexion angle and the knee valgus moment. All the kinetic and kinematic values were analysed at the peak valgus moment. Only the kinematics and kinetics of the right limb were calculated. Schreurs et al. (2017) found a reduction in the knee flexion angle with sharper CODs. In addition, Schreurs et al. (2017) found that greater knee valgus moments in athletes demonstrated sharper CODs (90°, 135° and 180°) compared to a 45° COD. However, Schreurs et al. (2017) found that both COD manoeuvres at 90° and 135°

degrees had similar knee valgus moment magnitudes indicating that these manoeuvres may have a similar risk of injury.

However, the studies were not without limitation. First, it should be noted that Schreurs et al. (2017) have investigated only knee biomechanics for dominant limbs. The authors defined the dominant limb as that preferred for kicking the ball. However, the dominant limb may be that which they prefer to use as the push-off limb during a COD manoeuvre. Although the biomechanical analysis of the COD manoeuvre showed no differences between the preferred and non-preferred limb (in Chapter 4 in this thesis), this may not be true if the biomechanical differences were investigated between dominant and non-dominant limbs. Therefore, defining the dominant limb on the way they change is still a factor. In addition, there is a lack of evidence of the differences in the hip frontal, transverse and sagittal planes between COD manoeuvres. Second, Schreurs et al. (2017) only investigated these differences at peak knee valgus moments. However, it is important to analyse kinetic and kinematic variables for hip and knee joints at multiple points throughout the entire stance phase of a COD manoeuvre, and not only one critical time point. For example, this could include: at the initial contact, at the peak VGRF, in the first 60 ms after initial contact, and at the peak knee valgus angle. As far as the researcher knows, these differences have never been examined at 90° and 135° COD manoeuvres at all critical time points, such detail will provide additional insight into the technique used for various directional changes at the risk of ACL injury.

Third, a variety of running speeds during COD manoeuvres, with different angles of 45°, 90° and 135°, have been examined. Each type of movement was examined at a different running speed. The average speed for males was 4.7 m/s, 3.8 m/s, 3.5 m/s and 3.4 m/s, during CODs at 45°, 90°, 135° and 180° respectively. For females these averages were 4.2 m/s, 3.6 m/s, 3.3 m/s and 3.2 m/s during CODs at 45°, 90°, 135° and 180° respectively (Schreurs et al., 2017). However, it should be noted that standardising the running speed of participants when comparing kinematics and kinetics is essential, to ensure the same speed and enable a more accurate comparison between individuals, which is not affected by their speed. Different speeds between COD manoeuvres may have affected the result. Increased running speeds have been shown to cause change in the kinematics and kinetics of the lower extremities (Nedergaard et al., 2014; Vanrenterghem et al., 2012), and the majority of researchers have confirmed that running speeds need to be standardised (Colby et al., 2000; Kadaba et al., 1989; Malinzak et al., 2001; Pollard et al., 2004; Queen et al., 2006). Therefore, speeds in the current study were controlled between participant and both manoeuvres. The participants used a completion time running speed of 4.2 m/s ± 0.5 for CODs at 90° and 135°.

Therefore, it is important to analyse kinetic and kinematic variables for hip and knee joints at multiple time points throughout the entire stance phase of COD manoeuvres, not only one critical time point. For example, at initial contact, at peak vGRF, at maximum knee flexion, at first 60 ms after initial contact and at peak knee valgus angle. As far as the researcher knows these differences have never been examined at 90° and 135° COD manoeuvres in all critical time points, such details will provide additional insight into technique used for the various directional changes at the risk of ACL injury.

Many sports require players to change direction using different COD angles, but there is limited information on kinetics and kinematics during COD at 90° and 135° angles, as well as the differences between 90° and 135° COD manoeuvres. Therefore, the aim of this study was to compare the hip and knee biomechanical characteristics between COD manoeuvres performed to 90° and 135°.

5.2 Research questions

Are there any differences in the biomechanical variables between 90° and 135° COD manoeuvres?

5.3 Null hypotheses

There are no significant differences in hip and knee biomechanical variables between 90° and 135° change of direction manoeuvres.

5.4 Methods

The biomechanical and strength analysis of the COD manoeuvre and MVIC in recreational healthy male soccer players showed no difference between the preferred and non-preferred limb (as discussed in Chapter 4). Therefore, for this study only the preferred limb will be examined.

For the details on the motion capture at both of the laboratories please refer to the method chapter (section 3.1.1 and 3.1.2).
For the details on the three-dimensional motion capture and markers placement please refer to the method chapter (section 3.1.3).

For the details on the study procedure please refer to the method chapter (section 3.1.5).
For the details on the recruitment please refer to the method chapter (section 3.1.6).
For the details on the inclusion and exclusion criteria please refer to the method chapter (section 3.1.7).
For the details on the data processing please refer to the chapter 4 (section 4.4.1).
For the details on the main outcome measures please refer to the chapter 4 (section 4.4.2).

5.4.1 Statistical analysis

The statistical analyses were carried out using Statistical Package for Social Sciences software (version 21, SPSS Statistics 20.Ink). Normality for each variable was checked with a Shapiro-Wilk test and histograms to check whether the data were normally distributed or not (parametric or non-parametric). For parametric variables, a paired t-test was used and for non-parametric variables a Wilcoxon Rank test was used (Edwards et al., 2012) to examine the biomechanical differences between 90° and 135° COD manoeuvres for preferred limb, a Holm method correction, α = (0.05/(41 comparisons – rank + 1) was used to control for family-wise error. The mean and standard deviation (SD) value of five trials of each test were calculated to find the differences. Effect sizes were determined using the Cohen δ method (Thomas et al., 2015), which defines 0.2, 0.5 and 0.8 as small, moderate and large respectively.

5.5 Result

36 participants took part in this analysis. The demographic characteristics of the participants are summarised in table 5-1. All participants completed five successful COD manoeuvres to 90° and 135° using their preferred limb. Participants were physically active, free from lower extremity injury, and had no history of lower extremity surgery. Participants were male recreational soccer players who frequently performing COD manoeuvre to 90° and 135°. 31 participants preferred to COD using right limb while 5 subjects using left limb. Valid trials for each were collected if the participants' limb landed on force plates. The results were collected as a difference between 90° and 135° COD manoeuvres.

Table 5-1 Participant demographics.

Demographic	Mean	SD
Age (years)	24.25	6.21
Height (m)	1.72	0.06
Mass (kg)	66.41	10.83
BMI (kg/m²)	19.28	2.89

Standard deviation (SD); metre (m); kilogramme (kg); body mass index (BMI); kilogramme per square meter (kg/m²).

The normality test (Shapiro-Wilk) for kinetic and kinematic variables revealed that all variables were normally distributed. Figure 5-1 shows sagittal, transverse and frontal planes hip joint angles, sagittal and frontal plane knee joint angles and moments and time normalised force-time curves for 90° and 135° COD manoeuvres.

The results of this study indicate that biomechanical differences exist between 90° and 135° COD manoeuvres (Table 5-3). Peak external knee abduction moment and knee valgus angle at IC during 135° COD manoeuvres were significantly higher than 90° COD manoeuvres (p= 0.000). However, similar peak VGRF and peak knee valgus angle. In addition, COD to 135 angle shows significant greater hip flexion ROM than COD to 90 angles.

Table 5-2 Time in millisecond for all phases.

ms	90°	135°
Full Phase	395	520
IC to PEKAM	28	40
IC to PVGRF	34	44
IC to PKVA	59	70
IC to PKFA	165	189

Peak external knee abduction moment (PEKAM); peak vertical ground reaction force (PVGRF); initial contact (IC); peak knee valgus angle (PKVA); millisecond (ms); peak knee flexion angle (PKFA).

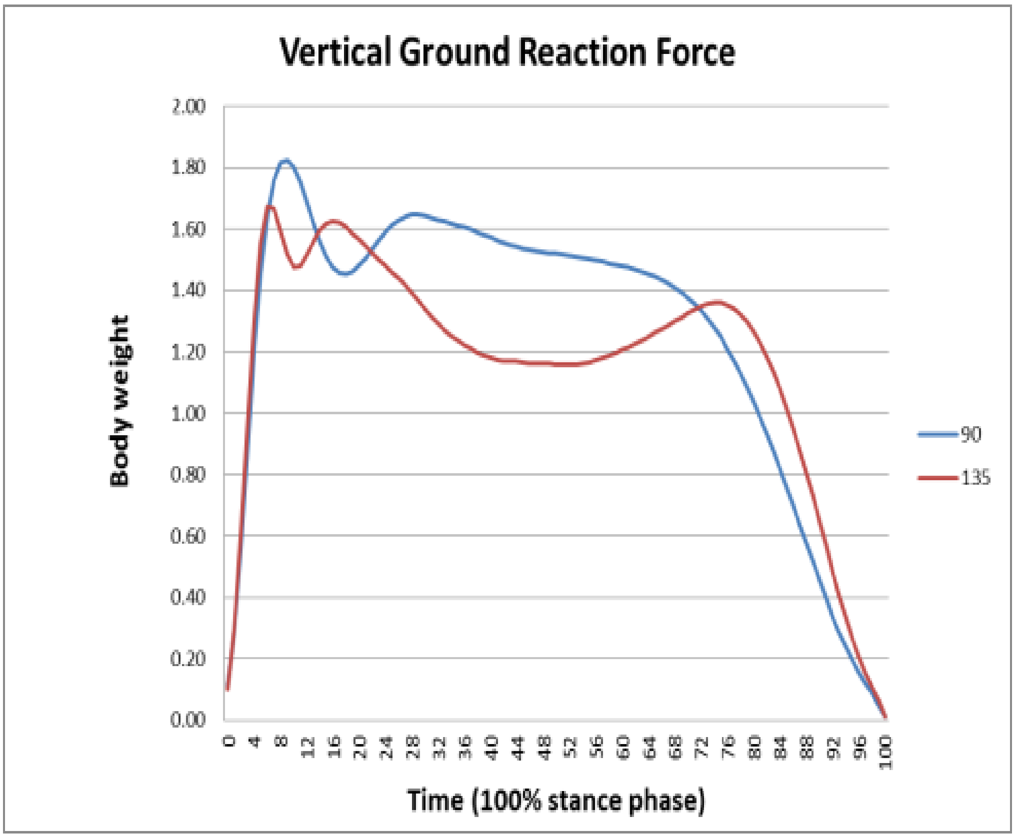
Table 5-3 Comparisons (mean ± SD) of hip and knee angles (degrees) and moments (Nm/kg) of 90° and 135° COD manoeuvres.

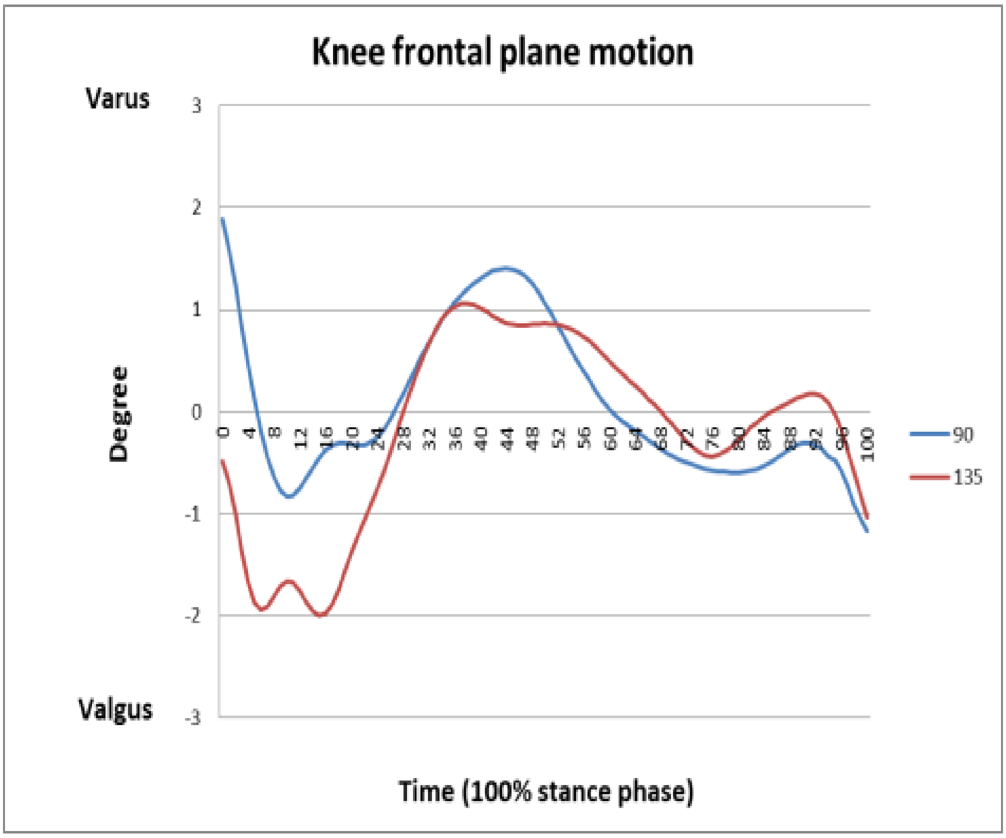
Variable	90° COD		135° COD		Raw p-value	Adjusted p-value[a]	ES
	Mean	SD	Mean	SD			
Kinematics							
Peak hip sagittal plane angle (°) at							
IC	39.67	7.39	37.03	8.18	0.005	0.001	0.50
PEKAM	41.44	7.44	39.47	8.60	0.04	0.002	0.37
PVGRF	42.28	7.56	39.65	8.53	0.008	0.001	0.47
PKVA	44.33	8.73	42.75	10.20	0.20	0.003	0.22
60 ms	46.06	8.36	43.67	10.28	0.04	0.002	0.35
PKFA	47.72	9.37	49.59	12.52	0.19	0.003	0.22
Hip sagittal ROM angle (°) between IC and							
PEKAM	1.77	1.84	2.44	2.46	0.06	0.002	0.22
PVGRF	2.61	2.35	2.61	2.32	0.86	0.017	0.02
PKVA	4.66	4.45	5.72	5.66	0.43	0.005	0.09
60 ms	6.39	4.37	6.64	4.87	0.71	0.008	0.06
PKFA	8.05	5.39	12.55	7.30	0.000*	0.001	0.81
Peak hip frontal angle (°) at							
IC	-20.45	6.61	-21.94	7.56	0.04	0.002	0.35
PEKAM	-19.42	6.65	-20.79	7.77	0.06	0.002	0.32
PVGRF	-19.21	6.67	-20.77	7.81	0.03	0.002	0.37
PKVA	-17.97	6.83	-19.60	8.15	0.05	0.002	0.34
60 ms	-18.09	6.89	-20.23	7.89	0.005	0.001	0.50
PKFA	-15.92	6.98	-17.04	7.97	0.14	0.002	0.25
Hip frontal ROM angle (°) between IC and							
PEKAM	1.04	1.05	1.15	1.44	0.69	0.007	0.05
PVGRF	1.24	1.24	1.16	1.49	0.22	0.003	0.15
PKVA	2.49	2.29	2.33	3.37	0.29	0.004	0.12
60 ms	2.36	1.96	1.71	2.21	0.02	0.001	0.27
PKFA	4.53	3.75	4.89	4.93	0.96	0.050	0.01
Peak hip transverse plane angle (°) at							
IC	4.83	9.18	7.82	9.04	0.005	0.001	0.50
PEKAM	-0.71	9.01	0.13	8.71	0.45	0.006	0.13
PVGRF	-1.42	9.02	-0.17	8.68	0.25	0.004	0.19
PKVA	-6.82	10.72	-6.73	11.02	0.95	0.025	0.01
60 ms	-4.20	9.01	-2.30	9.43	0.11	0.002	0.27
PKFA	-11.03	9.62	-9.98	9.23	0.30	0.004	0.18
Hip transverse rotation ROM angle (°) between IC and							
PEKAM	-5.54	4.21	-7.69	6.30	0.03	0.001	0.26
PVGRF	-6.25	4.49	-7.99	6.35	0.05	0.002	0.23
PKVA	-11.65	7.24	-14.55	7.78	0.05	0.002	0.34
60 ms	-9.03	5.53	-10.12	6.41	0.36	0.003	0.14
PKFA	-15.86	6.42	-17.80	7.26	0.08	0.002	0.21
Knee frontal plane angle (°)							
IC	1.89	3.92	-0.48	4.26	0.000*	0.001	1.09
Peak	-4.20	5.00	-5.87	5.78	0.005	0.001	0.50
ROM	-6.09	3.84	-5.39	4.07	0.20	0.003	0.22

Knee sagittal plane angle (°)							
IC	17.87	6.32	18.86	4.97	0.13	0.002	0.26
Peak	61.61	8.09	63.02	8.83	0.32	0.005	0.17
ROM	43.74	7.09	44.17	7.90	0.74	0.010	0.06
Moments							
PVGRF (*BW)	2.15	0.31	2.16	0.35	0.76	0.0125	0.05
PEKAM (Nm/Kg)	1.23	0.57	2.34	1.11	0.000*	0.001	1.04

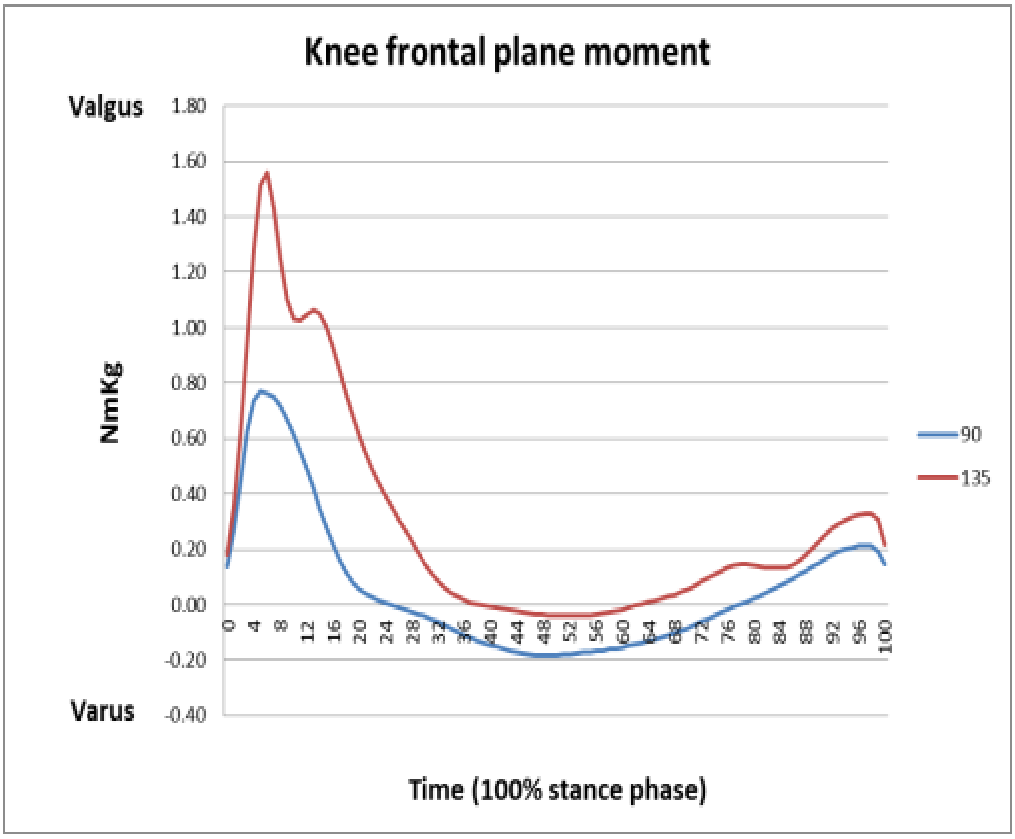
* The mean difference is significant.
Change of direction (COD); standard deviation (SD); effect size (ES); body weight (BW); angle (°); peak external knee abduction moment (PEKAM); newton meter per kilogram (Nm/Kg); peak vertical ground reaction force (PVGRF); initial contact (IC); peak knee valgus angle (PKVA); millisecond (ms); peak knee flexion angle (PKFA).
Sign conventions shows the position of the joints as; hip flexion (+), hip Extension (-), hip abduction (-), hip adduction (+), hip internal rotation (+) and hip external rotation (-).
[a] By holm method.

Figure 5-1 Averages of hip and knee angles, knee abduction moment and vertical force time curves during the stance phase for 90° and 135° COD manoeuvres (n=36), X-axis is percentage of stance phase.

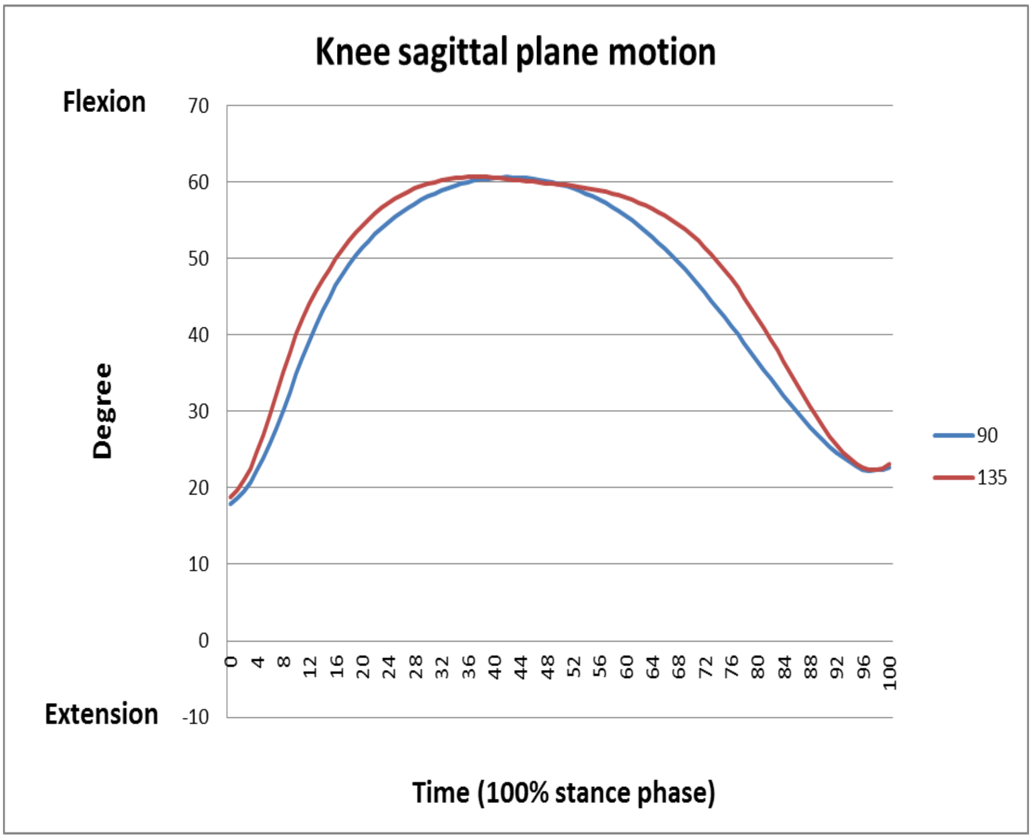




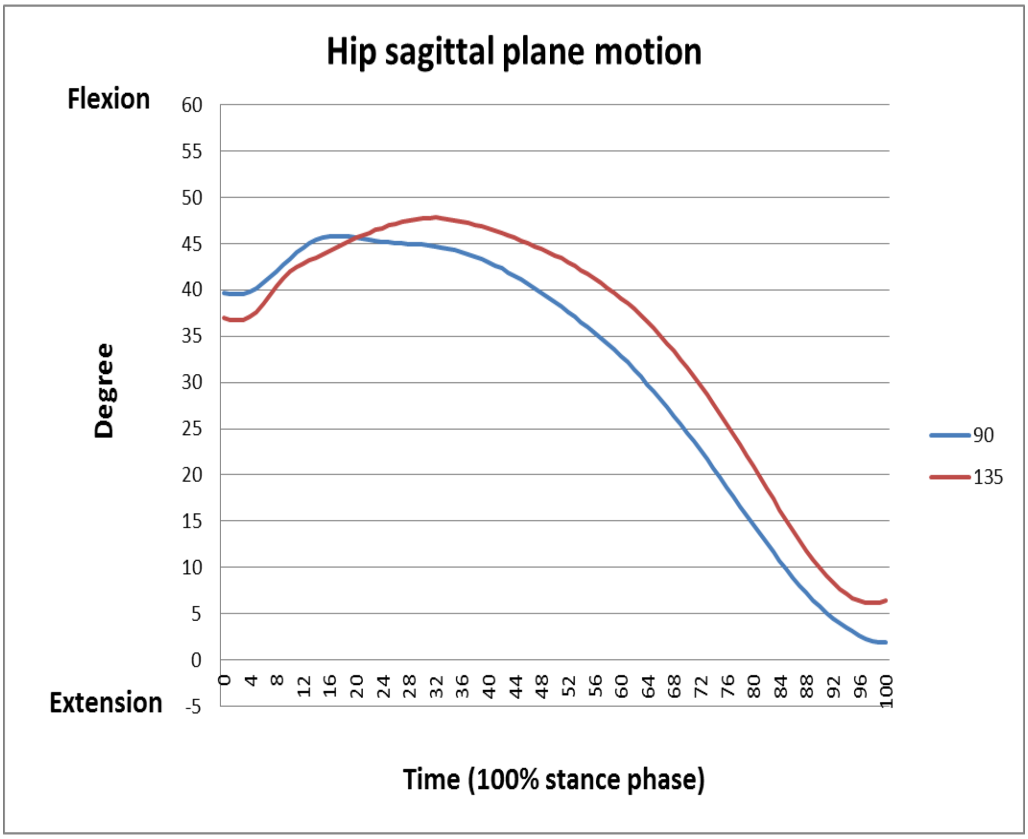
Knee frontal plane motion
Varus
Valgus
Degree
3
2
1
0
-1
-2
-3
0 4 8 12 16 20 24 28 32 36 40 44 48 52 56 60 64 68 72 76 80 84 88 92 96 100
90
135
Time (100% stance phase)



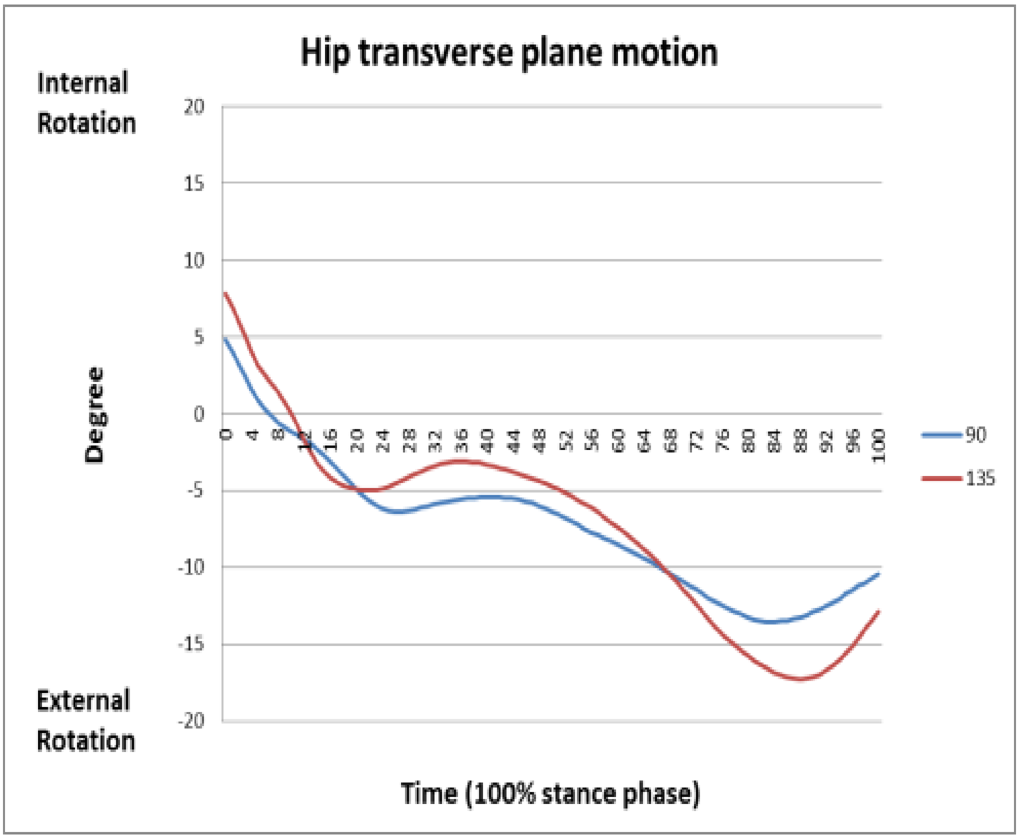
Knee frontal plane moment
Valgus
Varus
NmKg
1.80
1.60
1.40
1.20
1.00
0.80
0.60
0.40
0.20
0.00
-0.20
-0.40
0 4 8 12 16 20 24 28 32 36 40 44 48 52 56 60 64 68 72 76 80 84 88 92 96 100
90
135
Time (100% stance phase)

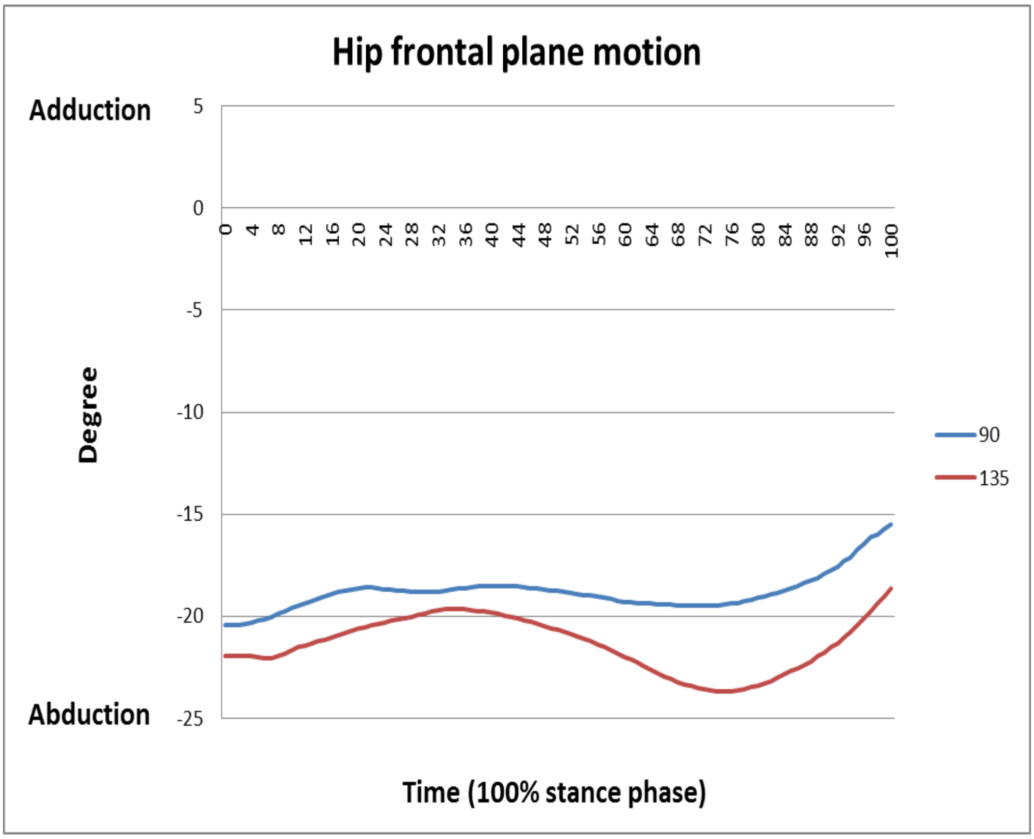


Knee sagittal plane motion
Flexion
Extension
Degree
70
60
50
40
30
20
10
0
-10
0
4
8
12
16
20
24
28
32
36
40
44
48
52
56
60
64
68
72
76
80
84
88
92
96
100
90
135
Time (100% stance phase)



Hip sagittal plane motion
Flexion
Extension
Degree
60
55
50
45
40
35
30
25
20
15
10
5
0
-5
0
4
8
12
16
20
24
28
32
36
40
44
48
52
56
60
64
68
72
76
80
84
88
92
96
100
90
135
Time (100% stance phase)


Hip transverse plane motion
Internal Rotation
External Rotation
Degree
20
15
10
5
0
-5
-10
-15
-20
0 4 8 12 16 20 24 28 32 36 40 44 48 52 56 60 64 68 72 76 80 84 88 92 96 100
90
135
Time (100% stance phase)


Hip frontal plane motion
Adduction
Abduction
Degree
5
0
-5
-10
-15
-20
-25
0 4 8 12 16 20 24 28 32 36 40 44 48 52 56 60 64 68 72 76 80 84 88 92 96 100
90
135
Time (100% stance phase)

5.6 **Discussion**

The aim of this study was to examine the biomechanical differences between 90° and 135° COD angle in recreational healthy male soccer participants. This is the first study to investigate the hip and knee biomechanical differences between COD manoeuvres to 90° and 135°. This is important given that COD manoeuvres performed to such angles are common in multidirectional sports (Bloomfield et al., 2007). In the present investigation, the results of this study indicate that biomechanical differences exist between 90° and 135° COD manoeuvres. In addition, recreational healthy male soccer shows different movement patterns while performing 90° and 135° COD manoeuvres. While few studies have explored hip and knee biomechanical differences between 45° and 90° (Havens & Sigward, 2015b; Imwalle et al., 2009) and between 45° and 110° (Sigward et al., 2015) and only knee biomechanical differences between 90° and 135° (Schreurs et al., 2017), this investigation is the first to examine the hip and knee biomechanical differences between the COD manoeuvres at 90° and 135°. After analysing hip and knee kinematics and kinetics during a COD manoeuvres, the current study results support biomechanical differences between the two COD manoeuvres and found that sharper angles place the knee more at risk. Furthermore, in terms of movement pattern, different COD angles demand different hip and knee kinematics and kinetics. When comparing the current study findings with previous studies (Havens & Sigward, 2015b; Imwalle et al., 2009; Schreurs et al., 2017; Sigward et al., 2015), similarities exist in which sharper COD angles place the knee more at risk.

External knee abduction moments were almost two times greater during the 135° COD manoeuvre compared with the 90° COD manoeuvres. These findings are concerning because greater external knee abduction moments have been associated with increased risk for ACL injury (Hewett et al., 2005; Jones et al., 2015; Kristianslund et al., 2014; Sigward et al., 2015) and also lead to increase ACL strain (Markolf et al., 1995; Markolf et al., 1990). This finding is similar to previous studies (Havens & Sigward, 2015b; Sigward et al., 2015) where greater knee valgus moments was observed in sharper COD angle. However, Schreurs et al. (2017) found that both COD manoeuvre at 90° and 135° degree had similar knee valgus moment magnitude. The peak external knee abduction moments in the current study was 2.34 Nm/Kg during COD to 135° angle. Consistent with this study's values, Marshall et al. (2015) found that peak external Knee abduction moment was 2.5 Nm/Kg during COD to 105° angle. The peak external Knee abduction moments in the current study were 1.23 Nm/Kg during COD to 90° angle. Consistent with this study's values, Jones et al. (2015) found that peak external Knee abduction moment was 1.26 Nm/Kg during COD to 90° angle. Generally, knee valgus moment was found to be greater during the 110° COD compared to a 45° angle (Sigward et al., 2015). In the current study external knee valgus moment was two times greater during the 135° COD manoeuvre compared with the 90° COD manoeuvres. This is in line with the study conducted by Sigward et al. (2015), which found that external knee valgus moment to be 2.4 times greater during the 110° COD manoeuvres compared to a 45° angle. In contrast, Schreurs et al. (2017) found that greater knee valgus moments in athletes demonstrating sharper CODs (90°, 135° and 180°) compared to 45° COD, however, no differences in knee valgus moments between 90°, 135° and 180° CODs. It should be noted that Schreurs et al. (2017) have investigated only 13 male and 16 female looked at five different types of movement: running forward, and COD at 45°, 90°, 135° and 180°. Each type of movement were with a different running speed. The average speed for males was 4.7 m/s, 3.8 m/s, 3.5 m/s and 3.4 m/s, during COD at 45°, 90°, 135° and 180° respectively. For females these averages were 4.2 m/s, 3.6 m/s, 3.3 m/s and 3.2 m/s during COD at 45°, 90°, 135° and 180° respectively. Standardising the running speed of participants when comparing kinematics and kinetics, to ensure the same speed enables a more accurate comparison between individuals, which is not affected by their speed. Increased running speeds have been shown to cause change in the kinematics and kinetics of the lower extremities (Nedergaard et al., 2014; Vanrenterghem et al., 2012), and the majority of researchers have confirmed that running speed needs to be standardised (Colby et al., 2000; Kadaba et al., 1989; Malinzak et al., 2001; Pollard et al., 2004; Queen et al., 2006). Therefore, speeds in the current study were controlled between participant and both manoeuvres. The participants used an approach running speed of 4.2 m/s ± 0.5 for COD at 90° and 135° manoeuvres. Furthermore, the differences in knee valgus moment may be due to the knee valgus moment being influenced by the extent of the GRF, as well as the moment arm in the frontal plane (Kristianslund et al., 2014). However, in the current study similar peak VGRF were found during COD manoeuvres to a 135º angle compared to a 90° COD. Other factors may influence knee valgus moment is the moment arm in the frontal plane (Kristianslund et al., 2014). Greater moment arms could occur as a result of greater hip abduction (Havens & Sigward, 2015b; Sigward et al., 2015). In the current study, the hip abduction angle increased when participants perform COD manoeuvres to a 135º angle compared to a 90° COD, 19° and 21° respectively. This may place the VGRF more lateraly thus increase moment arm and then as a result could increase knee valgus moment. Increase external knee valgus moment have been associated with increased risk for ACL injury (Hewett et al., 2005; Jones et al., 2015; Kristianslund et al., 2014; Sigward et al., 2015) and also lead to increase ACL strain (Markolf et al., 1995; Markolf et al., 1990; Shin et al., 2009; Withrow et al., 2006). All together, these data suggest that ACL injury risk may be higher when performing COD manoeuvres to a 135° angle compared to 90° angle.

In addition, greater knee valgus angles (Jones et al., 2016; Jones et al., 2015; Kristianslund et al., 2014; McLean et al., 2005; Sigward et al., 2015) occur with sharper CODs. In the present investigation, the results of this study indicate that knee valgus angle at IC during 135° COD manoeuvres were significantly higher than 90° COD manoeuvres. This finding is concerning because greater knee valgus angles has been identified as mechanisms and characteristics linked to ACL injuries (Grassi et al., 2017; Johnston et al., 2018; Koga et al., 2010; Montgomery et al., 2018; Walden et al., 2015). This finding is consistent with previous study (Sigward et al., 2015) greater knee valgus angle was observed in sharper COD angle (-1.33° and -0.61 during 110° and 45° COD manoeuvres respectively). In the current study, the knee valgus angles at IC were -0.48° and 1.89° during 135° and 90° COD manoeuvres respectively, with effect size = 1.09. However, Jones et al. (2015) found that knee valgus angles at IC was -1° during COD to 90° angle. Although Jones et al. (2015) used similar approach velocity (4.42 ms) during COD to 90° angle, investigated female soccer players may affect the knee valgus angle compare to male participant in current study which explained the differences. The peak knee valgus angle in the current study were -5.87° and -4.20° during 135° and 90° COD manoeuvres respectively. Peak knee valgus angle was close to reach level of significant with moderate effect size when compare 135° and 90° COD manoeuvres. In addition, previous studies have shown that non-contact ACL injuries occur quickly after ground initial contact. These studies have demonstrated that the estimated time of non-contact ACL injury ranged between 40 and 60 milliseconds after initial contact (Bates et al., 2020; Koga et al., 2010; Krosshaug et al., 2007). Within the current study, the peak knee valgus angle occurred at 13% (70 ms) and 15% (59 ms) of the stance phase during 135° and 90° COD manoeuvres respectively. It has also been demonstrated that greater knee valgus angles has been identified as mechanisms and characteristics linked to ACL injuries (Grassi et al., 2017; Johnston et al., 2018; Koga et al., 2010; Montgomery et al., 2018; Walden et al., 2015).

Hip and knee internal rotation angles, hip flexion angle were increased when carrying out COD manoeuvres at 90° compared with COD at 45° (Havens & Sigward, 2015a; Imwalle et al., 2009). Despite the knee biomechanical differences between 90° and 135° COD manoeuvres, no significant differences in hip sagittal, frontal and transverse plane kinematics were observed. The only difference was that COD to 135° angle shows significant greater hip flexion ROM than COD to 90° angles, but only in one phase (from IC to PKFA). Hip external rotation motion and ROM observed in both manoeuvres suggest that the hip contributes to body rotation into the new direction. Hip abduction was observed at IC and the rest phases during 90° and 135° COD manoeuvres, it was followed by adduction, to a less abducted position (Grassi et al., 2017; Johnston et al., 2018; Koga et al., 2010; Montgomery et al., 2018; Walden et al., 2015).

The current study helped to gain more understanding in differences between COD manoeuvre at 90° and 135° degree on hip and knee kinetics and kinematics. However, these data should be interpreted with caution, since testing took place in a laboratory setting. So, it's a closed skill task without any of the variability and unpredictability of carrying out the task in sport. In this study, only the preferred limb was considered, but earlier in chapter 4 results showed that there were no differences in kinetics and kinematics between the preferred and non-preferred limb during COD manoeuvre at 90° and 135° degree. Finally, this study included a recreational healthy male soccer population and results cannot be transferred to very athletic, inactive or patient populations as each population have different level of skill.

5.7 **Conclusion**

This study is the first to provide a biomechanical comparison of 90° and 135° degree COD manoeuvres in recreational healthy male soccer players. The purpose of the current study was to identify hip and knee biomechanical differences between COD manoeuvres at 90° and 135° angles. These data demonstrate that COD manoeuvres with sharper redirection demands result in greater frontal plane knee loading and therefore, sharper COD angles place the knee more at risk. Despite the knee biomechanical frontal plane differences between 90° and 135° COD manoeuvres, no significant differences in hip sagittal, frontal and transverse plane and knee sagittal plane kinematics were observed. The only difference was that COD to 135° angle shows significant greater hip flexion ROM than COD to 90° angles, but only in one phase (from IC to PKFA). After analysing hip and knee kinematics and kinetics during COD manoeuvres, the results of the current study support biomechanical differences between the two COD manoeuvres and found that sharper angles place the knee at greater risk. This could be explained by the finding that external knee abduction moments were almost two times greater during the 135° COD manoeuvre compared with the 90° COD manoeuvre. This may be because, in the current study, the hip abduction angle increased when participants performed COD manoeuvres to 135° angles compared with 90°, 19° and 21° COD respectively. This may place the VGRF more laterally and thus increase the moment arm and as a result also increase the knee abduction moment. Increasing the external knee abduction moment has been associated with an increased risk of ACL injury (Hewett et al., 2005; Jones et al., 2015; Kristianslund et al., 2014; Sigward et al., 2015) and lead to an increase in ACL strain (Markolf et al., 1995). In addition, the results of this study indicate that the knee valgus angle at IC during 135° COD manoeuvres was significantly higher than at 90° COD manoeuvres. The knee valgus angles at IC were -0.48° and 1.89° during 135° and 90° COD manoeuvres respectively, with an effect size of 1.09. At the initial contact, the limb is in a more extended and valgus profile with greater hip abduction during 135° COD manoeuvres. This may place the VGRF more laterally and thus increase the moment arm and as a result increase the knee abduction moment.

This finding is concerning because greater knee valgus angles have been identified as mechanisms and characteristics linked to ACL injury (Grassi et al., 2017; Johnston et al., 2018; Koga et al., 2010; Montgomery et al., 2018; Walden et al., 2015). In addition, previous studies have shown that non-contact ACL injuries occur quickly after initial contact with the ground. These studies have demonstrated that the estimated time of non-contact ACL injury ranged from 40 to 60 milliseconds after the initial contact (Bates et al., 2020; Koga et al., 2010; Krosshaug et al., 2007b). Within the current study, peak external knee abduction moments (during 90° and 135° COD manoeuvres), peak vertical ground reaction force (during 90° and 135° COD manoeuvres) and peak knee valgus angles (during 90° COD manoeuvres) occurred before 60 milliseconds. Thus, all the above mention differences between 90° and 135° COD manoeuvres occurred before 60 milliseconds after initial contact (see Table 5-2). But peak knee valgus angles during 135° COD manoeuvres occurred at 70 milliseconds, only 11 milliseconds after peak knee valgus angles during 90° COD manoeuvres. All together, these data suggest that an ACL injury risk may be higher when performing COD manoeuvres to a 135° angle compared to 90° angle.

It can be concluded that different COD angles demand different hip and knee kinematics and kinetics. COD manoeuvres at 90° may be useful for evaluating individuals but may not be challenging enough to reveal poor neuromuscular control over hip and knee motions. Therefore, sharper angles of examination should be utilized in the evaluation of risk for individuals. In addition, the COD test should be utilized gradually for the assessment of patients following ACL reconstruction to evaluate postoperative functional recovery and the risk of re-injury. An accurate evaluation of functional recovery is necessary to allow for the safe return to sports activity.

6 Chapter 6

The relationship between the hip kinematic and muscle strengths and the non-contact anterior cruciate ligament injury risk factors during 90° and 135° change of direction manoeuvres.

6.1 Introduction

Anterior cruciate ligament injuries are one of the common injuries of the knee (Kim, Bosque, Meehan, Jamali, & Marder, 2011). In England, the rate of anterior cruciate ligament (ACL) reconstruction increased 12 times from 2 per 100,000 person-years in 1997–1998 to 24.2 per 100,000 person-years in 2016–2017 (Abram et al., 2019). Over seventy percent of ACL injuries can be described as non-contact (Benis et al., 2018; Johnston et al., 2018; Waldén et al., 2015) and therefore understanding this non-contact risk of injury is of increasing importance. 60-70 % of non-contact ACL injuries typically happen during change of direction (COD) (Johnston et al., 2018; Montgomery et al., 2018). COD manoeuvres are important for many field sports, however they are unfortunately associated with non-contact ACL injuries (Boden et al., 2000; Cochrane et al., 2007; Havens & Sigward, 2015a; Krosshaug et al., 2007; Olsen et al., 2004). The results of previous research have shown that players perform 727 CODs during a 90 minute soccer game (Bloomfield et al., 2007). Although the players frequently COD at 90° and larger angle, limited knowledge is available on the kinetic and kinematic during COD at 90° and 135° angle. Especially, understanding of the knee and hip biomechanics and their relationship throughout COD manoeuvres at 90° and 135° angles is limited.

Alteration in the frontal plane biomechanics increase knee valgus (abduction) angle and knee valgus (external abduction) moment are thought to increase risk of non-contact ACL injury (Kristianslund & Krosshaug, 2013; McLean et al., 2008; Pollard et al., 2015; Sigward & Powers, 2007). High knee valgus angle and moment during COD manoeuvres are associated with joint positions including increased hip flexion, abduction and internal rotation angles during COD manoeuvres at 45° and 90° angles (Havens & Sigward, 2015a; Imwalle et al., 2009; McLean et al., 2005; Sigward & Powers, 2007). In addition, a number of researchers have discovered a link between reduced hip muscle strength and greater knee valgus angles during double leg and single leg (Gehring et al., 2009; Hollman et al., 2009; Jacobs et al., 2007; Willson et al., 2006), as well as knee valgus moments (Lawrence et al., 2008) during landings, squat and single leg step-down. Furthermore, isometric abduction and external hip rotation strength has been reported to independently predict ACL injuries (Khayambashi et al., 2016), indicating that weakness in hip abductor and external rotator muscles is a modifiable risk factor. These findings are concerning because greater knee valgus angles have been identified as mechanisms and characteristics linked to ACL injuries (Koga et al., 2010; Montgomery et al., 2018; Walden et al., 2015) and are linked to increase knee valgus moment (Jones et al., 2015; Kristianslund et al., 2014; Sigward et al., 2015), which can increase ACL strain (Markolf et al., 1995; Markolf et al., 1990). These findings, therefore, point towards the importance to look at the relationship between hip kinematics and strength and knee frontal plane variables which have been postulated to increase ACL injury risk. Thus, understanding such relationship is essential to mitigate injury risk and for the development of successful ACL injury prevention programs. Therefore, the aim of this study is to examine the hip kinematics and muscle strength and find the relationship to the ACL injury risk factors during COD manoeuvres.

Few studies have investigated the relationship between hip kinematics and knee valgus moments and angles (Havens & Sigward, 2015a; Imwalle et al., 2009; McLean et al., 2005; Sigward & Powers, 2007), however, the majority of those studies investigated the relationship involving small angles (30°, 45°, 60°). In addition, it should be noted that the studies that have investigated knee and hip biomechanics with relationship to non-contact ACL injury risk factors have been examined COD biomechanics only at one time point. For example, previous studies have examined the hip and knee relationship at various stance phase time points, at peak vGRF (Imwalle et al., 2009; Sigward et al., 2015) and at maximum knee flexion (Havens & Sigward, 2015a). However, there is a lack of evidence on the role of hip frontal, transverse and sagittal planes on the knee frontal plane loading at the time when the knee motion would be most affected such as maximum knee valgus angles and moment and at the time when the ACL injury expect to happen (60 ms after initial contact) (Koga et al., 2010). In addition, involving the knee abduction angle range of motion (ROM) as well is essential; as such details will help to further understand the true altered of the knee valgus which is more risk for ACL. Therefore, examining knee and hip joints ROM alongside peak angles during different time periods: (1) at initial contact (IC), (2) IC to peak vertical ground reaction force (PVGRF), (3) IC to the first 60 millisecond of stance (60 ms), (4) IC to peak external knee abduction moment (PEKAM), (5) IC to peak knee valgus angle (PKV), and (6) IC to peak knee flexion (PKF) could provide more complete analysis and additional understanding of the hip relationship to ACL injury mechanism. Therefore, all time intervals were specifically chosen due to the lack of agreement as to which interval is increase ACL stress; all are frequently used in ACL injury biomechanics research.

As far as the researcher is aware, no published studies have investigated the relationship between hip kinematics and strength and ACL injury risk factors (knee valgus angle and moment) over multi-phases throughout COD manoeuvres at 90° and 135° angles. Therefore, the aim for this study was to explore the relationships between ACL injury risk factors (knee valgus angle and moment) and hip kinematics and muscles strength during COD manoeuvres.

6.2 Research questions

- Is there a relationship between hip and knee biomechanical variables during 90° and 135° COD manoeuvres?
- Is there a relationship between isometric hip abductors, extensors and external rotators strength and knee biomechanical variables during 90° and 135° COD manoeuvres?

6.3 Null hypothesis

- There are significant relationships between hip sagittal, frontal and transvers kinematics and ACL injury risk factors (knee valgus angle and moment) during COD manoeuvres at 90° and 135°.
- There are significant relationships between hip abductors, extensors, and external rotators normalised peak torque during maximum voluntary isometric contraction (MVIC) test and ACL injury risk factors (knee valgus angle and moment) during COD manoeuvres at 90° and 135°.

6.4 Methods

For the details on the motion capture at both of the laboratories please refer to the method chapter (section 3.1.1 and 3.1.2).
For the details on the three-dimensional motion capture and markers placement please refer to the method chapter (section 3.1.3).
For the details on the study procedure please refer to the method chapter (section 3.1.5).
For the details on the recruitment please refer to the method chapter (section 3.1.6).

For the details on the inclusion and exclusion criteria please refer to the method chapter (section 3.1.7).

For the details on the data processing please refer to the chapter 4 (section 4.4.1).

For the details on the main outcome measures please refer to the chapter 4 (section 4.4.2).

6.4.1 Statistical analysis

The statistical analyses were carried out using Statistical Package for Social Sciences software (version 21, SPSS Statistics 20.Ink). Normality for each variable was checked with a Shapiro-Wilk test and histograms to check wither the data were normally distributed or not (parametric or non-parametric). For parametric variables, Pearson's correlation coefficient (r) was used to explore the relationships between hip kinematics and strength and ACL injury risk factors (knee valgus angle at IC, peak and ROM and peak external knee abduction moment) during 90° and 135° COD manoeuvres and MVIC test. Relationships involving nonparametric variables were explored using Spearman's rank correlation (p). The alpha level was set as $p<0.05$. In addition, Table 6-1 illustrates the interpretation of the strength of correlation coefficients used in this study (Hopkins, Marshall, Batterham, & Hanin, 2009). The mean and standard deviation (SD) value of five trials of each test were calculated to find the relationship.

Table 6-1 Correlation coefficient scores and levels of association.

Correlation coefficient score	Level of association
(0.10–0.29)	Small
(0.30–0.49)	Moderate
(0.50–0.70)	Large
(0.70-0.90)	very large
(0.90–1)	Extremely large

6.5 Result

36 participants were used in this analysis. The demographic characteristics of the participants are summarised in table 6-2. All participants completed five successful COD manoeuvres to 90° and 135° and MVIC test using their preferred limb. Participants were physically active, free from lower extremity injury, and had no history of lower extremity surgery. Participants were male recreational soccer players who frequently performing COD manoeuvre to 90° and 135°.

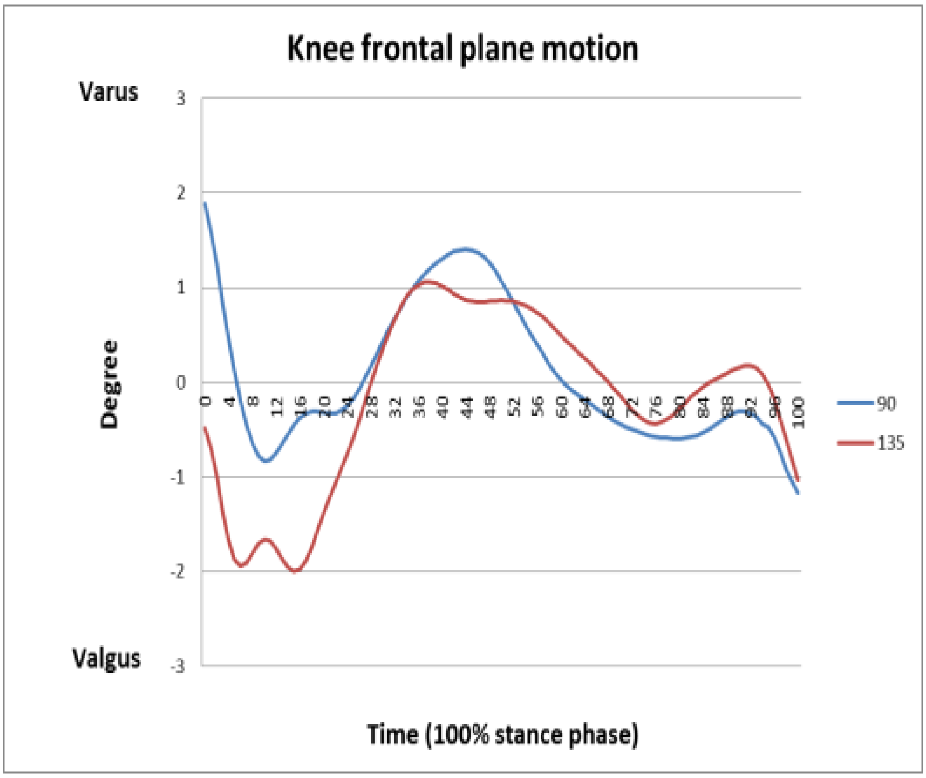
Table 6-2 Demographic information for all participants

Demographic	Mean	SD
Age (years)	24.25	6.21
Height (m)	1.72	0.06
Mass (kg)	66.41	10.83
BMI (kg/m²)	19.28	2.89

Standard deviation (SD); metre (m); kilogramme (kg); body mass index (BMI); kilogramme per square meter (kg/m²).

The normality test (Shapiro-Wilk) for kinematic variables and normalised peak torque revealed that all variables were normally distributed. The results found that peak knee valgus angle and external knee abduction moment were noted during early stance of both manoeuvres (Figure 6-1 and Table 6-3). At initial contact, the hip was flexed, abducted and internally rotated during both 90° and 135° COD manoeuvres. Then participants moved into a more hip flexed, adducted and externally rotated position during the deceleration phase both 90° and 135° COD manoeuvres (Figure 6-2). The results of the correlational analyses presented in table 6-4 for 90° COD manoeuvre and table 6-5 for 135° COD manoeuvre and table 6-6 for MVIC test.

Figure 6-1 Averages of knee valgus angle and external knee abduction moment curves during the stance phase for 90° and 135° COD manoeuvres (n=36), X-axis is percentage of stance phase.



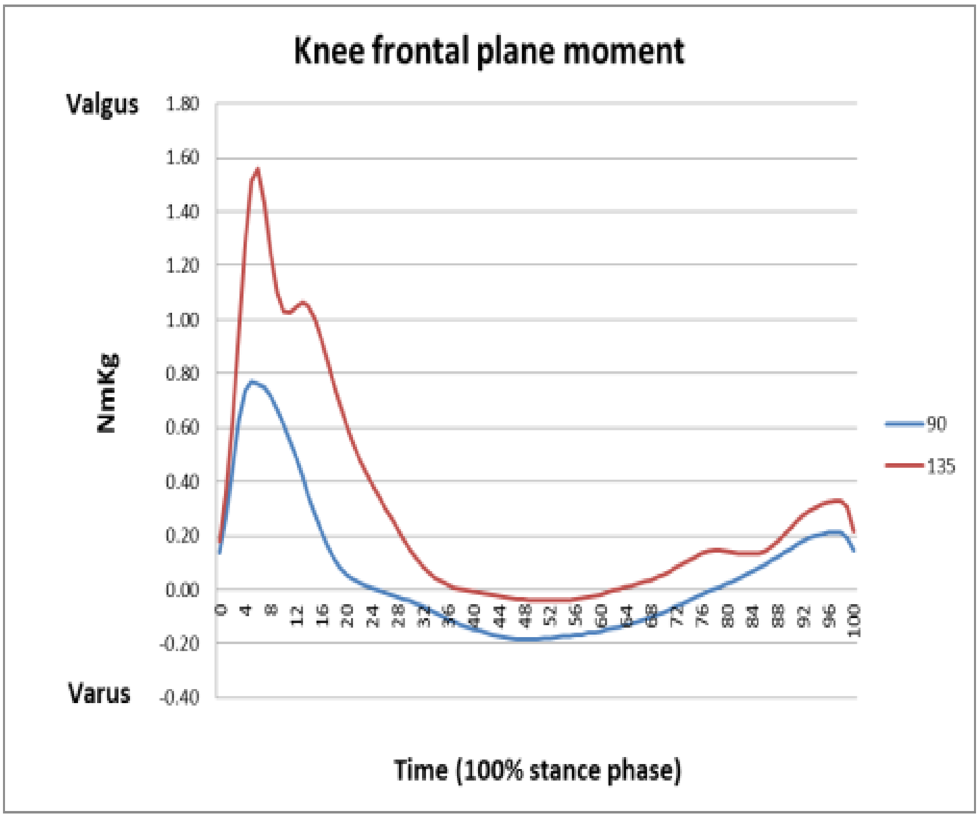
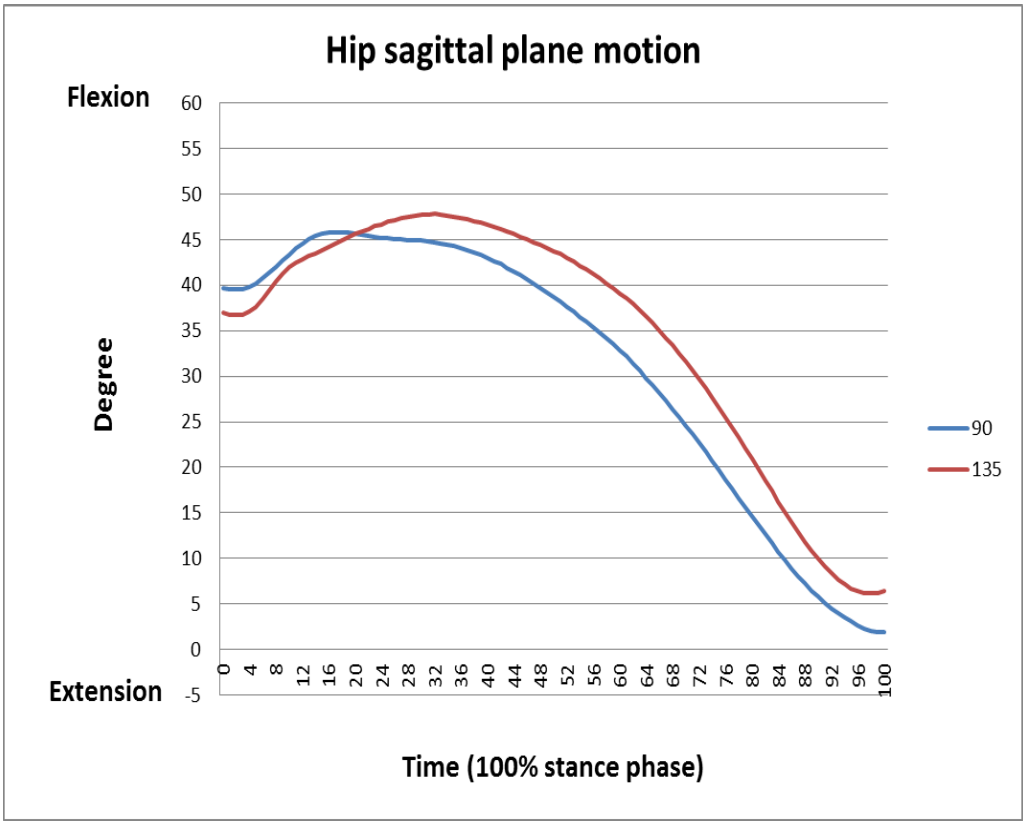


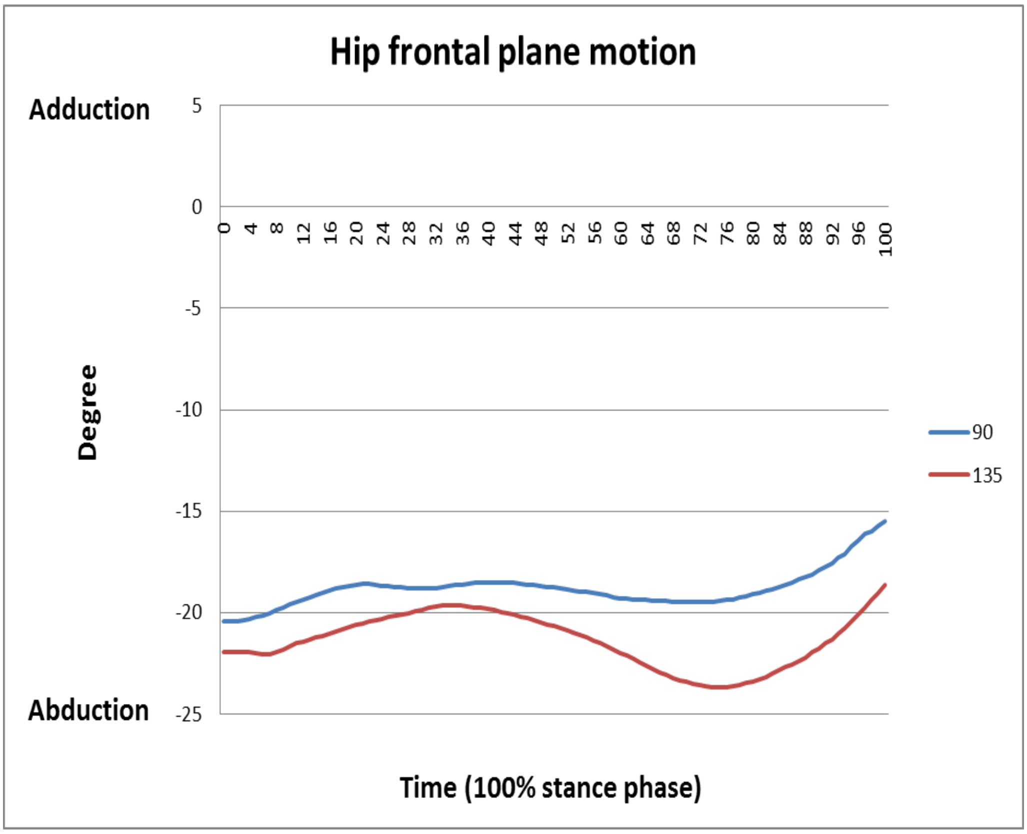
Table 6-3 Time in millisecond for all phases.

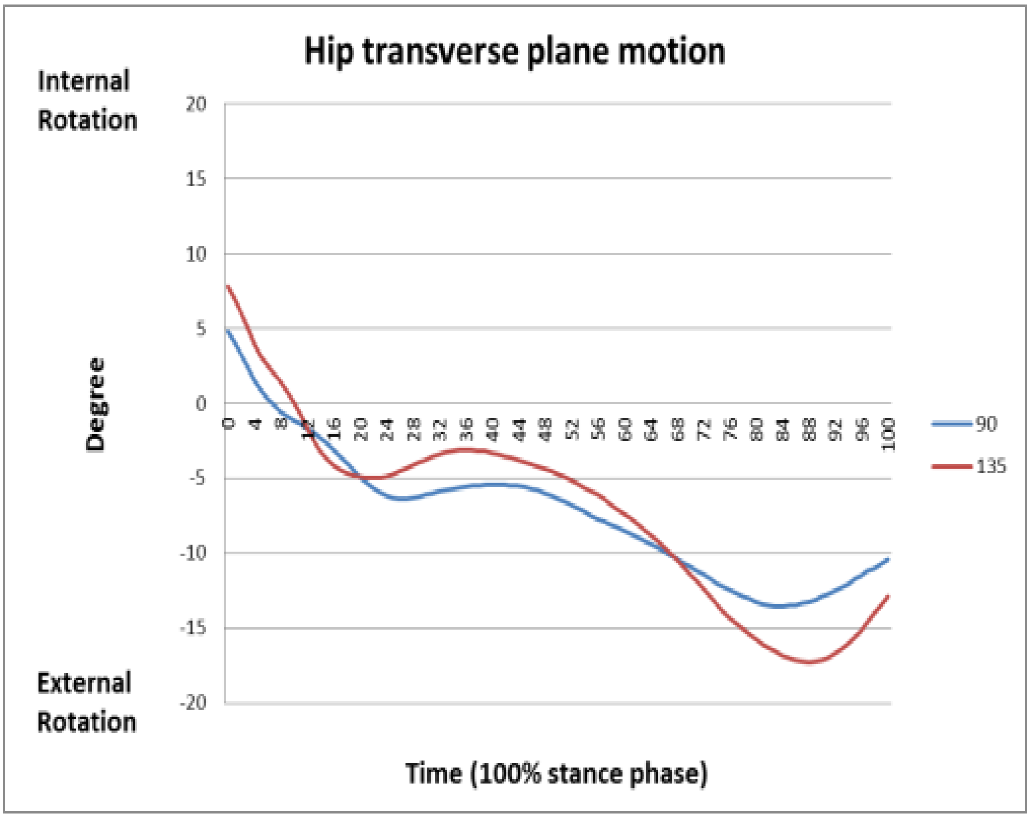
ms	90°	135°
Full Phase	395	520
IC to PEKAM	28	40
IC to PVGRF	34	44
IC to PKVA	59	70
IC to PKFA	165	189

Peak external knee abduction moment (PEKAM); peak vertical ground reaction force (PVGRF); initial contact (IC); peak knee valgus angle (PKVA); millisecond (ms); peak knee flexion angle (PKFA).

Figure 6-2 Averages of hip sagittal, frontal and transverse angles curves during the stance phase for 90° and 135° COD manoeuvres (n=36), X-axis is percentage of stance phase.







Hip Kinematics and muscle strength relationship with Knee kinetic and kinematics:

Hip kinematics (Table 6-4 and 6-5):

Knee valgus angle at IC:

A significant moderate correlation between hip frontal plane angle and knee valgus angle at IC only during 135° COD manoeuvre. This relationship suggests that as hip abduction angle increases at IC, knee valgus angle at IC increases.

Peak knee valgus angle:

A significant moderate correlation between hip transverse plane angle and peak knee valgus angle during 90° and 135° COD manoeuvres. This relationship suggests that as hip external rotation angle increases, peak knee valgus angle increases. However, a significant moderate correlation between hip frontal plane ROM and peak knee valgus angle was found only during 90° COD manoeuvre. This relationship suggests that as hip adduction ROM increases, peak knee valgus angle increases.

Knee valgus angle ROM:

A significant moderate to large correlation between hip transverse plane angle and knee valgus ROM angle during 90° and 135° COD manoeuvres. This relationship suggests that as hip external rotation angle increases, knee valgus ROM angle increases. However, a significant moderate to large correlation between hip frontal plane angle and knee valgus

ROM angle was found only during the 90° COD manoeuvre. This relationship suggests that as hip abduction angle increases, knee valgus ROM angle increases. In addition, a significant moderate to large correlation between hip transverse plane ROM and knee valgus ROM angle only during 90° COD manoeuvre. This relationship suggests that as hip external rotation ROM angle increase, knee valgus ROM angle increase. Furthermore, a significant moderate correlation between hip sagittal plane angle and ROM and knee valgus ROM angle only during 135° COD manoeuvre. This relationship suggests that as hip flexion angle and ROM increase, knee valgus ROM angle increase.

Peak external knee abduction moment:

A significant moderate to large correlation between hip sagittal plane ROM angle and peak external knee abduction moment during 90° and 135° COD manoeuvres. This relationship suggests that as hip flexion ROM increases, external knee abduction moment increases. In addition, a significant moderate correlation between hip frontal plane angle and peak external knee abduction moment during 90° and 135° COD manoeuvres. This relationship suggests that as hip abduction angle increases, external knee abduction moment increases. Furthermore, a significant moderate correlation between hip frontal plane ROM angle and peak external knee abduction moment during 90° and 135° COD manoeuvres. This relationship suggests that as hip adduction ROM increase, external knee abduction moment increase. However, a significant moderate correlation between hip transverse plane ROM angle and peak external knee abduction moment only during 90° COD manoeuvres. This relationship suggests that as hip external rotation ROM increase, external knee abduction moment increase.

Hip muscle strengths (Table 6-6):

In terms of the relationship of the hip strength with knee frontal plane kinematics and kinetics, no significant correlations were found between normalised peak torque hip extensors, abductors and external rotators muscles and peak knee valgus angle and peak external knee abduction moment.

Table 6-4 Pearson's correlation (R) and p value (P) between hip motion and kinematic/kinetic data of the knee joint during 90° COD manoeuvres.

Variable	Peak knee valgus angle		Knee valgus angle dis.		Peak EKAM		Knee valgus angle at IC	
	R	P	R	P	R	P	R	P
Peak hip sagittal plane angle (°) at								
IC	0.00	0.98	-0.27	0.12	-0.23	0.18	0.27	0.12
PEKAM	0.05	0.79	-0.24	0.15	-0.12	0.48	0.30	0.08
PVGRF	0.09	0.59	-0.19	0.25	-0.15	0.38	0.31	0.07
PKVA	0.15	0.39	-0.26	0.12	-0.13	0.45	**0.45**	0.01
60 ms	0.12	0.48	-0.19	0.26	-0.05	0.77	**0.34**	**0.04**
PKFA	0.13	0.46	-0.14	0.41	-0.05	0.75	0.30	0.07
Hip sagittal ROM angle (°) between IC and								
PEKAM	0.17	0.31	0.08	0.63	**0.42**	**0.01**	0.14	0.42
PVGRF	0.29	0.09	0.21	0.21	0.23	0.19	0.15	0.37
PKVA	0.28	0.09	-0.08	0.66	0.12	0.48	**0.44**	0.01
60 ms	0.22	0.19	0.08	0.63	0.29	0.09	0.20	0.24
PKFA	0.22	0.20	0.12	0.49	0.22	0.20	0.16	0.35
Peak hip frontal angle (°) at								
IC	0.14	0.42	**0.46**	**0.00**	**-0.37**	**0.03**	-0.28	0.10
PEKAM	0.14	0.42	**0.49**	**0.00**	-0.30	0.07	-0.31	0.07
PVGRF	0.15	0.37	**0.50**	**0.00**	**-0.33**	**0.05**	-0.29	0.09
PKVA	0.22	0.20	**0.47**	**0.00**	-0.32	0.06	-0.20	0.30
60 ms	0.23	0.17	**0.50**	**0.00**	-0.30	0.08	-0.18	0.25
PKFA	0.28	0.10	**0.54**	**0.00**	-0.21	0.21	-0.18	0.31
Hip frontal ROM angle (°) between IC and								
PEKAM	0.02	0.89	0.22	0.20	**0.39**	**0.02**	-0.18	0.28
PVGRF	0.10	0.55	0.21	0.22	0.18	0.30	-0.07	0.67
PKVA	0.26	0.13	0.05	0.75	0.11	0.51	0.28	0.10
60 ms	**0.36**	**0.03**	0.21	0.23	0.19	0.26	0.25	0.14
PKFA	0.28	0.10	0.20	0.25	**0.35**	**0.03**	0.16	0.34
Peak hip transverse plane angle (°) at								
IC	0.31	0.07	0.20	0.24	0.26	0.12	0.20	0.24
PEKAM	**0.43**	**0.01**	0.31	0.07	0.12	0.47	0.24	0.15
PVGRF	**0.44**	**0.01**	0.33	0.05	0.16	0.36	0.24	0.16
PKVA	**0.41**	**0.01**	**0.53**	**0.00**	0.12	0.48	-0.01	0.98
60 ms	**0.37**	**0.03**	**0.38**	**0.02**	0.14	0.42	0.09	0.59
PKFA	**0.45**	**0.01**	**0.47**	**0.00**	0.17	0.32	0.12	0.49
Hip transverse rotation ROM angle (°) between IC and								
PEKAM	0.24	0.16	0.22	0.20	**-0.36**	**0.03**	0.09	0.61
PVGRF	0.25	0.14	0.25	0.14	-0.21	0.21	0.07	0.68
PKVA	0.21	0.22	**0.54**	**0.00**	-0.15	0.39	-0.26	0.12
60 ms	0.08	0.64	0.29	0.09	-0.21	0.22	-0.18	0.30
PKFA	0.23	0.17	**0.41**	**0.01**	-0.12	0.50	-0.11	0.54

Significant correlations are noted in bold.
Peak external knee abduction moment (PEKAM); peak vertical ground reaction force (PVGRF); initial contact (IC); peak knee valgus angle (PKVA); millisecond (ms); peak knee flexion angle (PKFA).

Table 6-5 Pearson correlation's (R) and p value (P) between hip motion and kinematic/kinetic data of the knee joint during 135° COD manoeuvres.

Variable	Peak knee valgus angle		Knee valgus angle dis.		Peak EKAM		Knee valgus angle at IC	
	R	P	R	P	R	P	R	P
Peak hip sagittal plane angle (°) at								
IC	0.00	1.00	-0.24	0.16	-0.11	0.54	0.23	0.19
PEKAM	0.07	0.68	-0.25	0.14	0.06	0.73	**0.34**	**0.04**
PVGRF	0.09	0.62	-0.23	0.17	0.02	0.91	**0.34**	**0.04**
PKVA	-0.05	0.76	**-0.43**	**0.01**	0.08	0.64	**0.34**	**0.04**
60 ms	0.11	0.52	-0.31	0.06	0.05	0.75	**0.45**	**0.01**
PKFA	0.12	0.48	-0.28	0.09	0.11	0.51	**0.44**	**0.01**
Hip sagittal ROM angle (°) between IC and								
PEKAM	0.25	0.14	-0.09	0.58	**0.56**	**0.00**	**0.43**	**0.01**
PVGRF	0.32	0.05	-0.03	0.88	**0.45**	**0.01**	**0.46**	**0.00**
PKVA	-0.09	0.59	**-0.43**	**0.01**	0.30	0.08	0.28	0.09
60 ms	0.23	0.17	-0.26	0.12	0.29	0.08	**0.57**	**0.00**
PKFA	0.21	0.22	-0.22	0.20	0.31	0.06	**0.49**	**0.00**
Peak hip frontal angle (°) at								
IC	-0.07	0.70	0.31	0.06	**-0.40**	**0.01**	**-0.39**	**0.02**
PEKAM	-0.02	0.92	0.32	0.06	**-0.36**	**0.03**	-0.32	0.05
PVGRF	-0.02	0.91	0.31	0.07	**-0.36**	**0.03**	-0.32	0.05
PKVA	-0.04	0.83	0.26	0.12	-0.22	0.19	-0.30	0.08
60 ms	-0.03	0.86	0.28	0.10	-0.29	0.09	-0.31	0.07
PKFA	0.11	0.51	**0.38**	**0.02**	-0.19	0.27	-0.20	0.23
Hip frontal ROM angle (°) between IC and								
PEKAM	0.26	0.13	0.07	0.67	0.16	0.37	0.28	0.09
PVGRF	0.23	0.18	0.05	0.79	0.15	0.37	0.27	0.11
PKVA	0.06	0.73	-0.07	0.70	**0.37**	**0.03**	0.15	0.40
60 ms	0.12	0.50	-0.08	0.65	**0.35**	**0.04**	0.23	0.17
PKFA	0.29	0.09	0.13	0.46	0.31	0.06	0.27	0.12
Peak hip transverse plane angle (°) at								
IC	0.20	0.24	0.31	0.07	-0.04	0.82	-0.02	0.90
PEKAM	0.20	0.25	0.22	0.19	-0.09	0.62	0.06	0.74
PVGRF	0.21	0.21	0.22	0.19	-0.06	0.72	0.07	0.67
PKVA	**0.35**	**0.04**	**0.47**	**0.00**	-0.16	0.35	0.03	0.88
60 ms	0.25	0.14	**0.40**	**0.02**	-0.19	0.26	-0.04	0.81
PKFA	**0.35**	**0.04**	**0.46**	**0.00**	-0.11	0.53	0.03	0.85
Hip transverse rotation ROM angle (°) between IC and								
PEKAM	-0.01	0.94	-0.13	0.44	-0.06	0.71	0.11	0.52
PVGRF	0.01	0.97	-0.13	0.45	-0.03	0.87	0.13	0.44
PKVA	0.26	0.12	0.31	0.07	-0.18	0.28	0.06	0.72
60 ms	0.08	0.62	0.15	0.38	-0.23	0.18	-0.03	0.87
PKFA	0.19	0.26	0.20	0.24	-0.09	0.60	0.07	0.69

Significant correlations are noted in bold.
Peak external knee abduction moment (PEKAM); peak vertical ground reaction force (PVGRF); initial contact (IC); peak knee valgus angle (PKVA); millisecond (ms); peak knee flexion angle (PKFA).

Table 6-6 Pearson correlation's (R) and p value (P) between normalised peak torque of hip muscles and kinematic/kinetic data of the knee joint during 90° and 135 COD° manoeuvres.

Variable	Peak knee valgus angle		Knee valgus angle dis.		Peak EKAM		Knee valgus angle at IC	
Nm/Kg	R	P	R	P	R	P	R	P
COD 90								
Hip Abductors	-0.14	0.42	-0.04	0.82	0.03	0.84	-0.14	0.42
Hip Extensors	0.01	0.94	-0.08	0.64	0.05	0.76	0.10	0.58
Hip External Rotators	-0.08	0.64	-0.03	0.85	-0.03	0.85	-0.07	0.68
COD 135								
Hip Abductors	0.02	0.92	0.22	0.20	-0.20	0.24	-0.19	0.28
Hip Extensors	0.01	0.94	0.03	0.84	-0.20	0.25	-0.01	0.94
Hip External Rotators	0.00	0.98	0.00	1.00	-0.27	0.11	0.00	0.98

Figure 6-3 Scatter diagrams illustrating the linear relationship between peak knee abduction angle and peak hip external rotation angle during 90° COD manoeuvre.

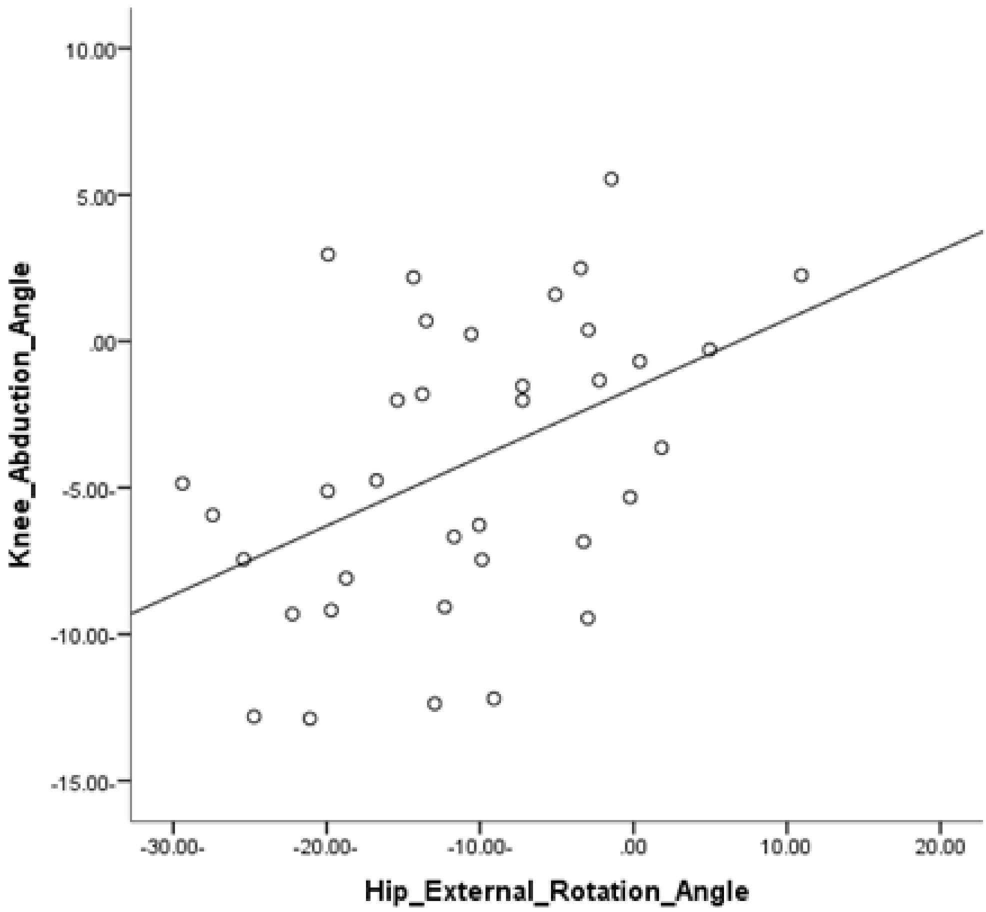
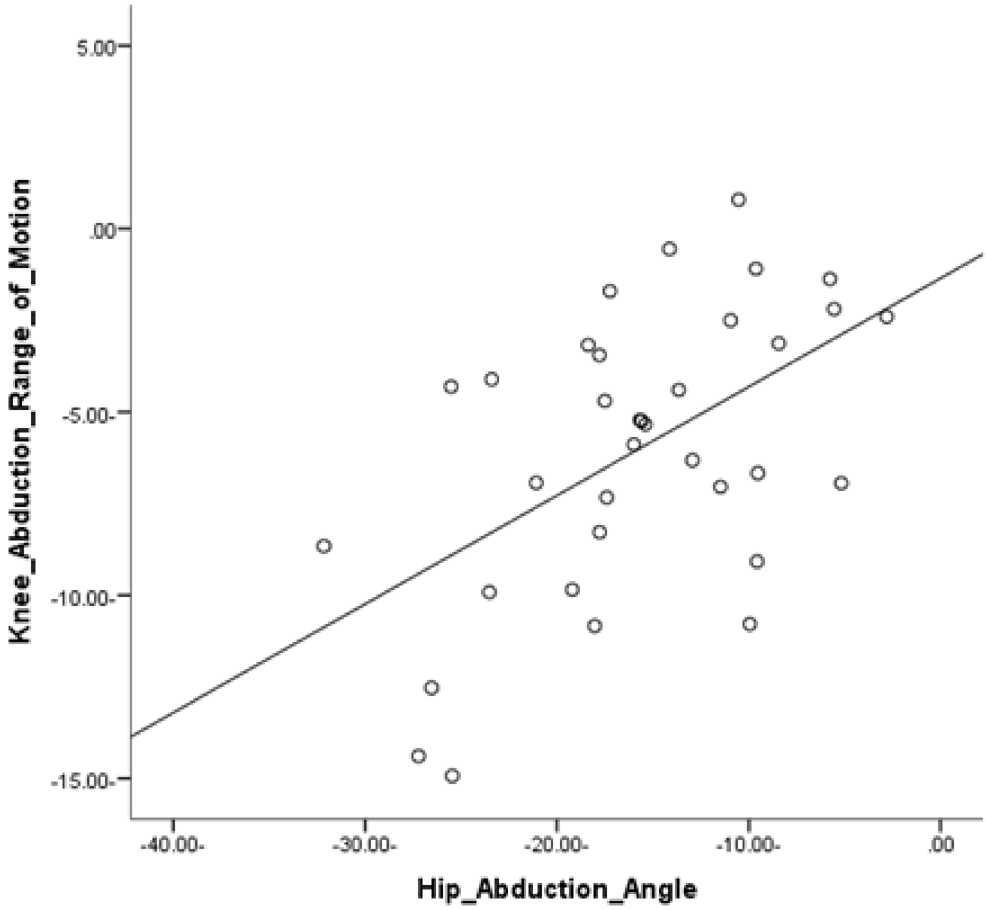


Figure 6-4 Scatter diagrams illustrating the linear relationship between knee abduction angle range of motion and peak hip abduction angle during 90° COD manoeuvre.



6.6 **Discussion**

The aim of this study was to examine the relationship between hip kinematics and muscle strength and ACL injury risk factors (knee valgus angle and moment) during 90° and 135° COD manoeuvres in recreational healthy male soccer participants. This is the first study to investigate the role that hip kinematics and muscle strength have with knee biomechanical variables during COD manoeuvre to 90° and 135°. This is important given that COD manoeuvres performed to such angles are common in multidirectional sports (Bloomfield et al., 2007). In the present investigation, the results of this study indicate that a moderate to large significant correlation was found between the hip sagittal, frontal and transverse angles and knee frontal plane kinematics and kinetics during 90° and 135° COD manoeuvres. The results highlight an important role the hip motion may play in kinematic and kinetic risk factors of ACL injury during COD manoeuvre. The significant moderate to large correlation of hip sagittal, frontal and transverse angles to knee abduction angle and external knee abduction moment is an important finding in this study. This findings are of some concern due to greater knee valgus angles being identified as mechanisms and characteristics linked to ACL injuries (Koga et al., 2010; Montgomery et al., 2018; Walden et al., 2015) and are linked to increase knee valgus moment (Jones et al., 2015; Kristianslund et al., 2014; Sigward et al., 2015), which can increase ACL strain (Markolf et al., 1995; Markolf et al., 1990).

During COD manoeuvres, external knee abduction moment is associated with hip abduction angle at IC during 45° COD manoeuvre (Sigward & Powers, 2007). However, this relationship does not hold true for COD performed to larger angle (90 and 110) (Havens & Sigward, 2015a; Sigward et al., 2015). However, in the current study findings, greater hip abduction angle at all phases result in increased knee abduction angle range of motion during 90° COD manoeuvres, which may increase risk of non-contact ACL injury.

This study found that a greater hip abduction angle resulted in increased external knee abduction moments during 90° and 135° COD manoeuvres. Carrying out a COD manoeuvre requires an abducted hip position, because greater hip abduction is needed to ensure a larger lateral foot plant distance and to generate medial-lateral forces. However, greater moment arms could occur as a result of greater hip abduction angles (Havens & Sigward, 2015b; Sigward et al., 2015). This may also place the VGRF more laterally, increase the moment arm, and then as a result increase the knee valgus moment. Increased external knee valgus moments have been associated with an increased risk for ACL injury (Hewett et al., 2005; Jones et al., 2015; Kristianslund et al., 2014; Sigward et al., 2015) and lead to increased ACL strain (Markolf et al., 1995). All together, these data suggest that ACL injury risk may be higher when performing COD manoeuvres with a more abducted hip position. Furthermore, this study found that when the hip was externally rotated, the knee abduction angle and range of motion increased during both 90° and 135° COD manoeuvres. The rotation of the pelvis on a femur creates an external hip rotation angle as the knee load is valgus. This is due to an increased hip adduction range of motion to compensate for the lack of ROM. The external hip rotation motion and ROM observed in both manoeuvres suggest that the hip contributes to body rotation in the new direction.

This relation is in contrast to previous studies that found a larger angle of hip internal rotation at initial contact which was related to greater knee abduction angle during COD manoeuvres to 45° (Sigward & Powers, 2007). However, the smaller angle of hip internal rotation at initial contact was related to greater external knee abduction moment during COD manoeuvres to 90° and 110° (Havens & Sigward, 2015a; Sigward et al., 2015). This suggests that more hip internal rotation during COD to larger angle (≥ 90) is beneficial for knee loading but to smaller angle (45) it is potentially risky for knee loading. Therefore, it would appear that the relationship between hip internal rotation and knee loading differs during COD performed at different angles. The result of this study also showed that greater hip flexion range of motion resulted in an increased external knee abduction moment during 135° COD manoeuvres. However, a relationship between sagittal plane hip kinematics and knee frontal plane loading was not assessed previously during COD manoeuvres in the study of Havens and Sigward (2015a) for example. This is the first study to examine the relationship between hip sagittal plane and knee abduction angle and external knee abduction moment during 90° and 135° COD manoeuvres.

The inconsistencies in the findings of this study when compared with the previous studies may be related to several possible factors. It should be noted that the previous studies have analysed hip kinematics at different time points. For example, previous studies have examined the hip and knee relationship only at IC (Sigward & Powers, 2007) or during the phase from IC to the first trough of the vertical GRF (Sigward et al., 2015) or during the phase from IC to the first 20% of the COD cycle (Sigward & Powers, 2007), or during the phase from IC to maximum knee flexion (Havens & Sigward, 2015a). However, the current study analysed hip kinematics over multiphases (IC, PEKAM, PVGRF, PKVA and 60 ms) which could provide a more complete analysis and additional understanding of hip function and their relationship to ACL injury mechanism. This study is the first study to provide a complete analysis of the hip kinematics curve during COD manoeuvres. The current study found that during all the time intervals there was a relationship between hip kinematics and ACL risk factors (knee abduction angle and moment). Specifically, PKVA, PEKAM and 60 ms time intervals appear to be the most involved in this relationship and during those time points, frontal knee loading and movement increased and thus risk of ACL injury may increase. In contrast, a less significant relationship between hip kinematics (frontal plane only) and knee abduction angle was found at initial contact phase. In general, this suggests that it is important to investigate hip kinematics over multiphases not only at one time point as this may affects the interpretation and the relationships present. Analysing the hip kinematics during different phases, would appear to explain the difference between current results and previous studies results (Havens & Sigward, 2015a; Sigward et al., 2015). Furthermore, there are differences in the limb assessed, the dominant foot (foot they would kick a ball with) (Havens & Sigward, 2015a; Sigward et al., 2015), right foot (McLean et al., 2005; Sigward & Powers, 2007), which could be different than preferred foot as not all the players COD by using the dominant limb, they may COD by using the preferred limb as the push-off limb. In the current study, the preferred limb was defined as it is the frequently used limb of the participants to use during COD manoeuvres rather than the limb which the participant kicked the ball with. In addition, when compare the current study results with previous studies, it should be considered that previous studies have been carried out with different populations. In this study, only male recreational soccer players were evaluated. In contrast, previous studies evaluate professional soccer player and included female with male as well participants (Havens & Sigward, 2015a; Sigward et al., 2015). The impact of the population sampled on the result could partially explain the differences in the results between current study and previous studies. Finally, it should be noted that standardising the running speed of participants when comparing kinematics and kinetics is important, the same speed enables a more accurate comparison between individuals. In addition, increased approach velocities have been shown to cause change in the kinematics and kinetics of the lower extremities, especially greater knee valgus moment (Kimura & Sakurai, 2013; Nedergaard et al., 2014; Vanrenterghem et al., 2012). different running speeds were used during COD manoeuvres in the previous studies. Therefore, speeds in the current study were controlled between participant and both manoeuvres. However, Havens and Sigward (2015a) also asked participants to run as fast as possible, which could be a reason explaining the findings for the differences with previous studies.

In addition, no significant correlation was found between hip strength and knee frontal plane kinematics and kinetics. A number of researchers have discovered a link between reduced hip muscle strength and greater knee valgus angles (Gehring et al., 2009; Hollman et al., 2009; Jacobs et al., 2007; Willson et al., 2006), as well as knee valgus moments (Lawrence et al., 2008). However, all the aforementioned studies have examined the relationship between hip muscle strength and Knee abduction angle and moment only during landing (Gehring et al., 2009; Jacobs et al., 2007; Lawrence et al., 2008; Suzuki, Omori, Uematsu, Nishino, & Endo, 2015), squat (Willson et al., 2006) and single leg step-down (Hollman et al., 2009). Although 60-70% of non-contact ACL injuries occurred while a player performed a COD manoeuvre (Johnston et al., 2018; Montgomery et al., 2018), there has been no published research correlating the hip strength and frontal plane knee biomechanics during COD manoeuvres. Therefore, this is the first study to investigate the hip muscle strength and knee biomechanics correlation during COD manoeuvre to 90° and 135°. In the current study, the results show that no significant correlations were found between normalised peak isometric torque of the hip extensors, abductors and external rotators muscles and peak knee abduction angle and peak external knee abduction moment during both 90° and 135° COD manoeuvres. However, during the other manoeuvres, data suggests that negative correlations were observed between hip muscle strength and frontal plane knee biomechanics related to increase risk of ACL injury (Hollman et al., 2009; Hollman, Hohl, Kraft, Strauss, & Traver, 2013; C. Jacobs & Mattacola, 2005; Lawrence et al., 2008; Stickler, Finley, & Gulgin, 2015; Suzuki et al., 2015). In contrast, other studies failed to find similar relationships. For example, (Nilstad, Krosshaug, Mok, Bahr, & Andersen, 2015) reported that no relationship between hip strength and frontal plane knee valgus angles during a drop-landing manoeuvre in 279 Norwegian elite female soccer players. In addition, Homan, Norcross, Goerger, Prentice, and Blackburn (2013) looked at the impact of hip abductor and external rotator peak strength on knee abduction; hip adduction, and hip internal rotation angles while landing, and participants with lower peak strength had similar frontal and transverse plane hip and knee kinematics, whereas those with greater peak strength, demonstrated that peak strength on its own cannot be used to predict hip and knee kinematics while performing dynamic manoeuvres. Moreover, the effect of implementing a hip strengthening protocol for knee kinematics was analysed by Stearns and Powers (2014), and they found that four weeks post training, there was an increase in the peak strength of the hip abductors and extensors, yet there was only a 1.2° reduction in peak knee abduction angle for a landing manoeuvres, although it was not considered statistically significant ($p = 0.07$). While it is intuitive to consider hip muscle weakness a risk factor for ACL injury, experimentally the relationship between hip strength and injury mechanics continues to be inconsistent. These inconsistencies may be because of the variety of manoeuvres used for each study. In addition, this may be due to the manner in which strength is tested. All the above-mentioned studies have quantified strength by measuring the peak torque produced during a MVIC, peak torque not normally being reached until 250-400 milliseconds (ms) (Aagaard, 2003; Aagaard & Andersen, 1998) after torque onset. This is different to ACL injuries, which typically occur within the first 40-60 ms during single-leg landing and COD manoeuvres (Kernozek & Ragan, 2008; Koga et al., 2010; Krosshaug et al., 2007), also maximum hip adduction is normally reached during landing within 130-150 ms following the initial contact (Lephart et al., 2002). Therefore, the maximal torque parameter is may not reveal true relationship. More comprehensive measures of muscle function are needed, such as examine rate of force development in shortest time or eccentric muscle strength, and this emphasises the necessity to exert maximal torque in the shortest minimal time, as well as when injuries such as to the ACL arise.

Collectively, our data have important implications for ACL injury prevention programs and for assessment of patients following ACL reconstruction to evaluate postoperative functional recovery and the risk for reinjures. As there is a lack of data existing in literature regards knee and hip motion characteristics during COD at 90° and 135° manoeuvres, the findings of this study can provide valuable evidence which can help understand three-dimensional knee motion and hip relations during COD at 90° and 135° manoeuvres. COD manoeuvres, especially at sharper angle, puts players at a higher risk for injury or reinjury; evaluating and predicting such manoeuvre may allow clinicians to reduce injury risk. Our results suggest that lack of sagittal, frontal and transverse plane hip control during the COD at 90° and 135° manoeuvres resulted in positions of increased knee valgus angle and external knee abduction moment; which have all been shown to be significant predictors of ACL injury. A better understanding of ACL injury mechanisms might serve to improve current prevention and rehabilitation strategies, thus reducing the risk of ACL and secondary injuries. Our results suggest that recreation players who perform a COD at 90° and 135° manoeuvres with increased hip abduction angle, hip adduction ROM, hip external rotation angle and ROM and hip flexion ROM are at risk for increased ACL loading. Understanding such risky position may allow clinicians to screen players who do not keep the neuromuscular control to properly perform a COD at 90° and 135° manoeuvres and consequently reducing the risk of ACL injury and reinjury.

Based on the results of this study, it is proposed that abduction angle of the hip at IC should be included as a factor for evaluation in COD test results. For subjects with increased hip abduction angle at IC, exercises, include plyometric, dynamic stabilization and neuromuscular training, that correct the hip abduction position at initial foot contact may effectively reduce the risk for ACL injury or reinjury after ACL reconstruction. However, based on the results of this study, strategies to prevent knee valgus angle and external knee abduction moment in recreational healthy male soccer players should focus in part on the control of hip adduction ROM with training protocols that include plyometric, dynamic stabilization and neuromuscular training. A relationship between sagittal plane hip mechanics and knee frontal plane loading was found in this study, as hip flexion ROM increases during early COD phase (first 44 ms), external knee abduction moment increases. Thus, prevention programs should focus on avoid increase hip flexion when performing COD manoeuvres early after IC. Furthermore, performing COD at 90° and 135° manoeuvres in a more hip external rotation posture may result in increased knee valgus angle, knee valgus angle ROM and external knee abduction moment, which were predictive of greater knee frontal plane loading and thus increase risk of ACL injury. This suggests that injury prevention programs should discourage hip external rotation movements and may reduce injury risk factors during COD at 90° and 135° manoeuvres.

In summary, narrow cut, less hip external rotation and less hip flexion ROM during COD at 90° and 135° manoeuvres should be targeted in future ACL injury prevention programme to reduce knee valgus angle and external knee abduction moment and likely reduce ACL injury risk. The knowledge of the relation between hip kinematics variables and knee valgus angle and external knee abduction moments can help inform the development of training protocols and training goals to promote safer COD technique.

While the current study helped to gain more understanding of the role of the hip on ACL injury risk factors during COD manoeuvres at 90° and 135° degree, caution must be taken before applying these data to injury prevention training. These data were tested in a laboratory setting so cannot create a true natural scenario in which ACL injury most often occurs. In this study, only the preferred limb was considered, but earlier in chapter 4 results showed that there were no differences in kinetics and kinematics and muscle strength between the preferred and non-preferred limb during COD manoeuvre at 90° and 135° degree. The participants represented a recreational healthy male soccer population without lower-limb problems. It is not known if similar results would be found in other populations, for example; athletic, inactive or patient populations. Peak hip muscle torque was considered in this study. It is likely that other meauerment method (e.g., rate torque development) could provide additional and different results.

6.7 **Conclusion**

This is the first study to analyse the relationship between hip kinematics and muscle strength and knee valgus angle and external knee abduction moment, which have been postulated to increase ACL injury risk during COD manoeuvres to 90° and 135° over multiple phases. These data demonstrate that hip kinematics play an important role in the ACL risk factors. Increased hip flexion range of motion between initial contact and peak external knee abduction moment and between initial contact and peak vertical ground reaction force during 135° COD manoeuvre resulted in an increased external knee abduction moment and thus may increase risk of non-contact ACL injury. Moreover, greater peak hip abduction angle at initial contact, peak external knee abduction moment, peak vertical ground reaction force, peak knee valgus angle, 60 ms and peak knee flexion angle result in increased knee abduction angle range of motion during 90° COD manoeuvres, which may increase risk of non-contact ACL injury. Furthermore, greater peak hip external rotation angle and range of motion at peak external knee abduction moment, peak vertical ground reaction force, peak knee valgus angle and peak knee flexion angle result in increased knee abduction angle and knee abduction range of motion during 90° and 135° COD manoeuvres, which may increase risk of non-contact ACL injury. However, the findings suggest that normalised peak torque of hip muscles during MVIC are not correlated to the frontal knee loading and movement during COD manoeuvres. This suggest that other factors (e.g. rate of torque development) might be more important in controlling knee movement during COD manoeuvres.

7 Chapter 7: Overall Summary, conclusion and recommendations

7.1 Summary:

Change of direction (COD) manoeuvres presents a contradiction to players. On the one hand, quick changes of direction are necessary for successful participation in multidirectional sports (Karcher & Buchheit, 2014; Orendurff et al., 2010; Prodromos et al., 2007). On the other hand, COD manoeuvres are associated with anterior cruciate ligament (ACL) injury risk (Benis et al., 2018; Grassi et al., 2017; Johnston et al., 2018; Montgomery et al., 2018). Players using both limbs allow for variety when performing COD manoeuvres. However, to date, no published study has evaluated limb asymmetry during COD manoeuvre at 90° and 135°. Previous studies have postulated that a link between ACL injury biomechanical risk factors (knee valgus angle and moment) and hip joint kinematics and muscle strength (Gehring et al., 2009; Havens & Sigward, 2015a; Hollman et al., 2009; Imwalle et al., 2009; Jacobs et al., 2007; Lawrence et al., 2008; McLean et al., 2005; Sigward & Powers, 2007; Willson et al., 2006). However, at the time of writing the thesis, no published studies investigated the relationship between hip kinematics and strength and ACL injury risk factors (knee valgus angle and moment) over multiple phases throughout COD manoeuvres at 90° and 135° angles.

An ounce of prevention is worth a pound of cure statement by Benjamin Franklin. In order to mitigate the poor long-term consequences of ACL tears and reconstruction, such as knee instability, pain, and early onset osteoarthritis of the joint (Lohmander et al., 2007), this injury must be prevented from happening in the first place. However, movement patterns thought to decrease the risk for injury are still not fully understood during COD to a large angle (≥ 90°). In order to develop training programs that reduce the risk for injury, an understanding of mechanics during COD manoeuvres and their relationship to injuries knee loading is needed. Therefore, the purposes of this thesis was to (1) determine whether asymmetry in knee and hip biomechanics kinematics and kinetics and hip muscle strength between preferred and non-preferred limbs during COD manoeuvres at 90° and 135° angles exists, (2) determine whether differences in knee and hip biomechanics kinematics and kinetics between COD manoeuvres at 90° and 135° angles exists and (3) explore the relationships between ACL injury risk factors (knee valgus angle and moment) and hip kinematics and muscles strength during 90° and 135° COD manoeuvres. In order to achieve these aims, the current thesis had specific elements with specific aims:

1) To examine the between-days reliability of 3D during change of direction tasks.

2) To examine the between-days reliability of the Biodex system to measure hip muscle strengths.
3) Establish whether asymmetry in knee biomechanics kinematics and kinetics and hip kinematics as well as hip muscle strengths between preferred and non-preferred limbs during COD manoeuvres at 90° and 135° angles exists.
4) Establish whether differences in knee biomechanics kinematics and kinetics and hip kinematics between 90° and 135° change of direction manoeuvres exists.
5) To explore the relationship between the hip kinematic and muscle strengths and the non-contact ACL injury risk factors during 90° and 135° COD manoeuvres.

7.2 **Conclusion:**

With respect to aim one and two, intra-rater between-day reliability and measurement error of 3D variables during the COD manoeuvre was investigated. The ICC values for all variables for 3D variables were good to excellent (ICCs 0.85-0.98) with low measurement error revealed (SEM 0.44-1.68°, 0.03-0.10 Nm/kg and 0.01-0.05 *BW). These results suggest that 3D variables measured for the COD manoeuvre are highly reliable and reproducible. Therefore, 3D variables were noted be reliable and therefore suitable for use within-day sessions. Furthermore, the measurement error values for 3D variables have been set out. In addition, regarding the between-day reliability of isometric muscle, the ICC values for normalised peak torque of the hip abductors, extensors, and external rotators muscles was found to be excellent, ranging between 0.92 and 0.95. The SEM values for normalised peak torque of hip abductors ranged from 0.01 to 0.03 Nm/kg. The first two studies revealed an increase in confidence concerning the ability to collect reliable data by following the measurement instructions set out in Chapter 3, which makes assessing differences and relationships more likely to lead to valid results in the main study.

To reach the third aim, it was necessary to explore how the participants performed during two manoeuvres. It was also important to find out whether symmetry between limbs existed in order to discover whether it is feasible for one limb to define the other’s performance. In the event of differences existing, it may be possible to reach a better clinical and biomechanical understanding and control of dynamic knee valgus. Thus, the purpose of Chapter 4 was to compare hip and knee biomechanics and hip muscle strength between preferred and non-preferred limbs during manoeuvres with different direction demands (90˚ and 135˚ COD manoeuvres). Consistent with our hypotheses, there were no significant differences in hip and knee biomechanical variables between limbs during 90°

and 135° COD manoeuvres. In addition, there were no significant differences in the hip muscles normalised peak torque between limbs during MVIC test. This chapter provided that the biomechanical and m analysis of the COD manoeuvre and MVIC in recreational healthy male soccer players showed no differences exist between preferred and non-preferred limb, thus only preferred limb was examined in chapter 5 and 6.

Many sports require players to change direction using different COD angles, but there is limited information on the kinematics and kinetics during COD at sharper 90° and 135° angles, as well as the differences between 90° and 135° COD manoeuvres. Therefore, the purpose of Chapter 5 was to compare the hip and knee biomechanical between COD manoeuvres performed to 90° and 135. Contrary to our hypotheses, there were significant differences in knee biomechanical variables between 90° and 135° change of direction manoeuvres. During the 135° COD, peak external knee abduction moment and knee valgus angle at IC were significantly higher than 90° COD manoeuvres. These data suggest that ACL injury risk may be higher when performing COD manoeuvres to sharper angles. Furthermore, when analysing lower limb biomechanics, different directions should be examined as they will have different profiles.

Despite its importance in sports, little is known about the hip mechanics and muscle strength related to increase risk of ACL injury during 90° and 135° COD manoeuvres, which could be critical information to injury prevention programs. It is essential to identify the factors associated with the risk of injury in order to introduce effective screening and interventions to prevent ACL injuries. Therefore, the purpose of Chapter 6 was to explore the relationships between ACL injury risk factors (knee valgus angle and moment) and hip kinematics and muscles strength during COD manoeuvres. Contrary to our hypotheses, there were significant relationships between hip sagittal, frontal and transverse plane kinematics and ACL injury risk factors (knee valgus angle and moment) during COD manoeuvres at 90° and 135°. However, consistent with our hypotheses, there were no significant relationships between hip abductors, extensors, and external rotators normalised peak torque during maximum voluntary isometric contraction (MVIC) test and ACL injury risk factors (knee valgus angle and moment) during COD manoeuvres at 90° and 135°.

The results from this thesis provide a more thorough understanding of the mechanics during 90° and 135° COD manoeuvres. Chapter 5 revealed that the biomechanical demands of COD to smaller and larger angles differ, and that COD to larger angle increase risk of ACL injury. COD manoeuvres at 90° may be useful for evaluating individuals but may not be challenging enough to reveal poor neuromuscular control over hip and knee motion. Therefore, sharper angles of examination should be utilized in the evaluation of individuals. However, the hip kinematics that were increased knee frontal plane loading and motion as found in chapter 6, were similar between the manoeuvres in chapter 5. Together, the results of this thesis show that increasing COD angle affects only knee loading and motion but not hip motion. Furthermore, the hip kinematics play an important role to increase risk of ACL injury. Moreover, these results may help provide an appropriate manipulation and intervention on a COD manoeuvre to reduce the risk of ACL injury. The findings from the current research will provide additional knowledge on the occurrence of ACL injury, and may assist in planning and designing better neuromuscular, plyometric, and strength training protocols in an attempt to prevent injuries.

In light of the findings of this thesis, current COD research should be interpreted with several methodological factors in mind. First, this thesis has clearly shown that the angle at which a COD is performed affects knee loading and motion. It is therefore important to consider the angle at which COD are made when interpreting the results of other studies. Second, the movement patterns described in this thesis should be interpreted with the experimental design in mind. The tasks chosen for this thesis were pre-planned, which allowed subjects to make anticipatory adjustments. If these tasks were performed under unanticipated conditions (Boden et al., 2000; Cochrane et al., 2007), different movement patterns would likely have been observed. Third, the recreational healthy soccer players in this thesis would be considered trained and had never incurred an ACL injury or had any lower extremity surgery, performed these tasks at standardised speed. Perhaps as a result of the movement speed or COD angle, some of the joint mechanics qualitatively seem to differ from previously published studies, when comparing the same COD angular magnitude. Fourth, these are the first studies to have analysed the knee and hip biomechanical over multiphases. As a result of this, some mechanics seem to differ from previously published studies, when comparing the same COD manoeuvres. Thus, caution must be taken when comparing COD literature, as this thesis has pointed to several methodological factors that can influence mechanics.

There were several limitations within the present thesis. One limitation of the study may be that the COD manoeuvre was tested in a laboratory setting so cannot create a true natural scenario in which an ACL injury most commonly occurs. In this study, the participants wore standard training shoes on a Mondo running surface to standardise the effect of shoe wear between subjects; this fails to represent typical shoe-surface tractions in real games, such as studded boots on grass and trainers on AstroTurf. The interaction between such shoes and surfaces may not reflect the actual interaction for some sports, such as football, or other sports that are played on grass. However, the majority of studies that have investigated ACL injuries are performed on flat surfaces with no boots with cleats, which is a limitation. Moreover, in a rehab environment, this is routinely done on gym floors. There is a further need for research in this area although only a few facilities in the world have access to 'real-life' biomechanical testing where boots can be worn and data can be collected (e.g., Manchester Institute of Health and Performance 3G Performance Capture).

Another limitation may be that the participants used an average time to complete speed during the task of 4.2 m/s ± 0.5 for COD at 90° and 135° manoeuvres. However, using the timing Gate System placed at the start and end of the 8 m path to measure the average time to complete the task speed may not be accurate as the subject could approach slowly and then exit fast or approach fast and then exit slowly. Therefore, measure the approach speed instead of average time to complete the task speed would be useful to standardise the running speed of participants when comparing kinematics and kinetics, as ensuring the same speed enables a more accurate comparison between individuals, which is not affected by their speed. Further details on the ways to calculate approach speed accurately in chapter 3, section 3.1.5.1.

In addition, the movement patterns described in this thesis should be interpreted with the experimental design in mind. The tasks chosen for this thesis were planned, which allowed subjects to make anticipatory adjustments. Besier et al., (2001a) discovered that the planned and un-planned tasks are actually two different tasks. Furthermore, the majority of non-contact ACL injuries are occurring in un-anticipated COD manoeuvres. However, this thesis only measured responses from an anticipated COD manoeuvres. Therefore, if the tasks for this thesis were performed during un-planned conditions, different mechanisms that may actually cause the injury would likely have been observed with different results.

Another limitation is that motion and force plate data were filtered with cut-off frequencies of 12 Hz for motion data and 25 Hz for force data. Different cut-off frequencies may result in large moment peaks. Kristianslund et al. (2012) indicated that significantly larger moments were observed during a sidestep cut when different cut-off frequencies were applied to motion and GRF data (10 and 50 Hz, respectively) compared to when the same cut-off frequencies were applied (10 and 10 Hz, respectively). It should be noted that the significant differences were observed only when the motion and force data were filtered at large different cut-off frequencies (10-50 Hz). This may not stand true if the motion and force data were filtered at different small cut-off frequencies (12-25 Hz).

Another limitation of this thesis is related to the fact that the hip was flexed at 90 degree during the hip external rotation strength test when the hip flexion angle ranges from 30° to 50° during COD manoeuvres (Havens and Sigward 2015a, Sigward and Powers, 2007). Therefore, the hip flexion angle during hip external rotator test is way outside the +/- 20 degree range thus this test may not representative of the actual strength of hip external rotation muscle at their functioning length during COD manoeuvres. However, isometric hip external rotator strength is most commonly measured at 90° of hip flexion in seated position and is easier for individuals to perform. Understanding whether changing this hip flexion angle during isometric tests for the external rotators is a potential future direction.

Moreover, due to peak strength being measured throughout the strength assessment, even though some participants may only use submaximal strength; to address this, practice trials were conducted, and rest periods were offered to assist the participants in producing maximum force. Furthermore, peak hip muscle torque was considered in this study. As ACL injury estimated time ranged between 17 and 60 milliseconds after initial contact (Bates et al., 2020; Koga et al., 2010; Krosshaug et al., 2007), it is likely that other measurement method (e.g., rate torque development) could provide additional and different results. In addition, the research has included recreational health soccer players, which means it is not possible to generalise the results to other sports' athletes, injured subjects or healthy active populations, as they may differ. Moreover, in Chapter 5 and 6, only the preferred limb was considered, but earlier in chapter 4 results showed that there were no differences in kinetics and kinematics and muscle strength between the preferred and non-preferred limb during COD manoeuvre at 90° and 135° degree. Finally, since no females were recruited, these findings cannot be generalized across genders.

7.3 Recommendations

7.3.1 Recommendation for Practitioner

The results of this research show that intervention programmes need to target hip motion control as this could play a key role in supporting knee biomechanics, including knee valgus angle and moment. Such programmes could have ACL prevention and rehabilitation strategies, which are important for patients with poorer lower limb biomechanics, as analysed during COD manoeuvres. It may be advisable, therefore, to implement hip motion control programmes alongside other programmes, for example, providing visual verbal feedback on COD strategies for different directions. These new protocols could help to reduce the likelihood of an ACL, as opposed to traditional strengthening programmes alone. The relationship found between hip sagittal, frontal and transvers planes and knee valgus angle and moment suggests that interventions to improve hip motion control might be important.

The main considerations of this thesis reflect Stage 2 of the sequence of injury prevention (Van Mechelen et al., 1992) and the TRIPP model (Finch, 2006) in the context of soccer. From study one, people should be biomechanical symmetrical when doing 90° and 135° COD manoeuvres. If asymmetry is noted, this could signify a greater risk of injury (stage 2) so an intervention should be made and be symmetrical (stage 3). From study 2, sharper angles place the knee at greater risk (stage 2) so when someone is rehabilitated they should start with 90° COD manoeuvres because these are less risky from 135° COD. Moreover, before the individual returns to sport they should be tested with 135° COD manoeuvres (stage 3). In addition, the COD test should be gradually utilized to assess patients following ACL reconstruction to evaluate postoperative functional recovery and the risk for re-injury.

From study 3, the data have important implications for ACL injury prevention/rehabilitation programs and for the assessment of patients following ACL reconstruction to evaluate postoperative functional recovery and the risk of re-injury. As there is a lack of data in the literature regarding knee and hip motion characteristics during CODs at 90° and 135°, the findings of this thesis can provide valuable evidence which can help to understand three-dimensional knee motion and hip relations during COD at 90° and 135°. COD manoeuvres, especially at sharper angles, place players at a greater risk of injury or re-injury; evaluating and predicting such manoeuvres may allow clinicians to reduce the injury risk. Our results suggest that a lack of sagittal, frontal and transverse plane hip control during the COD at 90° and 135° result in positions involving increased knee valgus angles and external knee abduction moments, which have all been shown to be significant predictors of ACL injury. A better understanding of ACL injury mechanisms might serve to improve current prevention and rehabilitation strategies, thus reducing the risk of ACL and secondary injuries.

Our results suggest that soccer recreation players who perform a COD at 90° and 135° with an increased hip abduction angle, hip adduction ROM, hip external rotation angle and ROM and hip flexion ROM are at risk for increased ACL loading (stage 2). Understanding such risky positions may allow clinicians to screen players who do not keep the neuromuscular control to properly perform a COD at 90° and 135° and consequently reduce the risk of ACL injury and re-injury. Thus, narrow cut, less external hip rotation and less hip flexion ROM during COD at 90° and 135° should be targeted in future ACL injury prevention/rehabilitation programmes. This would help to reduce the knee valgus angle and external knee abduction moment and be likely to reduce the risk of ACL injury (stage 3). Knowledge of the relationship between hip kinematics variables and the knee valgus angle and external knee abduction moments can help inform the development of training protocols and training goals to promote safer COD techniques.

However, hip muscles peak torque failed to reduce the knee valgus angle and moment. This may suggest that more explosive force produced by muscle could help in controlling the motion and thus reduce ACL injury risk factors. More research is needed to confirm the relationship between hip muscles rate of torque development and ACL injury risk factors. Moreover, the results from Chapter 4 indicate that analysing any limb during COD should represent the other limb, thus using one limb as a control.

7.3.2 Recommendation for Further Studies

The results of this research show that several questions have been posed that require further investigation. This research has increased the knowledge on the use of 3D motion analysis in the assessment of injury risk behaviours among healthy recreational players. Furthermore, the research has shown that 3D is reliable between-day, and a reliable, standardised protocol has been produced for COD manoeuvres, with associated measurement error scores for the evaluation of the participant’s performance. Primarily, the reliability study revealed that the CAST model should be used to measure kinematic and kinetic variables during COD manoeuvres in future investigations. Following the results of the reliability of 3D shown in chapter three, it is recommended that the COD manoeuvres should be used in future studies. The error measurement statistics set out in Chapter three may assist investigators in precisely determining if alterations in biomechanical variables are because of intervention or measurement errors. However, it should be noted that the results apply one rater, and it is as yet unknown whether similar reliability could be achieved by several evaluators, therefore, between-raters reliability is something that should be explored in future research as clinical situations typically involve patients being examined by different therapists.

Identification of participants who exhibit excessive knee valgus angle and moment could help reduce injury occurrence by implementing interventions to reduce knee valgus angle and moment. Improvement of poor hip biomechanics pattern may lead to reduce in knee valgus angle and moment. However, a basic approach to feedback could lead to speedy reductions in knee valgus angle and moment, possibly resulting in an immediate reduction in the risk of injury. Future work on feedback training is necessary to find out whether feedback protocols to correct the hip biomechanics would results in reduce knee valgus angle and moment. Additional research is needed in order to find out the underlying causes of poor mechanics when performing COD manoeuvres, as this would support devising more efficient injury prevention protocols. However, having established that there is no correlation between hip muscle peak torque and knee valgus angle and moment in this thesis, future work should investigate what other possible factors may affect ACL injury risk factors, such as hip muscles rate of torque development. Future research is necessary to discover whether there is a relationship between explosive hip muscle strength and frontal plane knee loading and motion during COD manoeuvres.

Further investigation into the ability of the COD manoeuvres as predictors of ACL injury risk is also needed. Future studies using the COD manoeuvres could utilise the protocols set out in Chapter Three. In addition, the extent to which these tests can detect functional deficits should be explored, based on the measurement error values provided. Furthermore, studies should involve a range of athletes or other injured populations to address generalisability, and this would assist in finding out different populations perform COD manoeuvres with regard to lower limb biomechanics and strength. Moreover, this would help in detecting athletes who are a risk due to excessive joint angles or moments, as these place such athletes at a higher risk of ACL injuries. Moreover, a large prospective study should be conducted that applies all of these measurements and tracks to better identify individuals at a high risk of knee injury during change of direction manoeuvres. Further research is required to assess the interactions between the trunk-pelvis and hip-knee during COD manoeuvres to provide more understanding of the ACL injury risk factors. The movement coordination between joints/segments are important during COD manoeuvres. The role of the variability/coordination between segments and joints in ACL injury risk during 90° and 135° COD manoeuvres should be identified.

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9 Appendices:

9.1 Ethical approval letter

University of Salford
MANCHESTER

Research, Innovation and Academic Engagement Ethical Approval Panel

Research Centres Support Team
G0.3 Joule House
University of Salford
M5 4WT

T +44(0)161 295 2280

www.salford.ac.uk/

13 September 2016

Dear Ayman,

RE: ETHICS APPLICATION HSCR 16-88 – Between-day reliability and agreement and relationship of kinematic, kinetic and isometric measurements during a sidestep 90 degree cutting task (CUT).

Based on the information you provided, I am pleased to inform you that your application HSCR16-88 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 Health-ResearchEthics@salford.ac.uk

Yours sincerely,

Sue McAndrew
Chair of the Research Ethics Panel

Amendment Notification Form

Please complete this form and submit it to the Health Research Ethics Panel that reviewed the original proposal: Health-ResearchEthics@Salford.ac.uk

Title of Project: Between-day reliability and agreement and relationship of kinematic, kinetic and isometric measurements during a sidestep 90 and 135 degree cutting tasks (CUT)

Name of Lead Applicant: **Ayman Abdullah Alhammad** *School:* **Health of Sciences**

Date when original approval was obtained: 13-09-2016 *Reference No:* HSCR16/88

Please outline the proposed changes to the project. ***NB.*** *If the changes require any amendments to the PIS, Consent Form(s) or recruitment material, then please submit these with this form highlighting where the changes have been made:*

The only amendments in the ethics forms from the originally approved document are an additional angle (135 degrees added) and also an increase in the number of subjects to account for an additional 28 subjects needed.

These have been reflected in the below documents along with the title.
Participant Information Sheet
Informed Consent Form
Data Collection Sheet
Health Research Ethics Application

Please say whether the proposed changes present any new ethical issues or changes to ethical issues that were identified in the original ethics review, and provide details of how these will be addressed:

There are no new ethical issues or changes to ethical issues that were identified in the original ethics review.

Chair's Signature:

Approved: 09-02-2018 ✓

Version 1.0 – 19 June 2017

9.2 Abstract from Scandinavian Sports Medicine Congress

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8 The effect of change of direction angle on knee mechanics – implications for ACL injury FREE

Ayman Alhammad, Lee Herrington, Paul Jones, Ritchard K Jones

Author affiliations +

Abstract

Introduction Change of direction (COD) manoeuvres are important for many field sports, however they are unfortunately associated with non-contact ACL injuries. Although players frequently COD at >90° angles, limited knowledge is available on knee joint kinematics and kinetics during COD at 90° and 135° and whether limb preference impacts knee mechanics during COD at these angles, which formed the aims of this study.

Methods Healthy male recreational soccer players were recruited to take part in the study. 3D kinematics and kinetics were collected during COD manoeuvres at 90° and 135°. Running speed was controlled at 4±0.4 m/s and 3.5±0.3 m/s, respectively. To determine differences on variables associated with ACL risk; knee abduction angle and moment, across cutting angles and preferred legs, a paired sample t-test was conducted using a Holm method correction, α=(0.05/ (8 comparisons – rank +1).

Results 36 individuals took part in the study (24.25±6.21 years, 1.72±0.06 m and 66.41±10.83 kg). COD at 135° showed greater knee abduction angles and moments than at 90° but with similar peak VGRF. There were no differences between preferred and non-preferred legs, apart from the increased knee flexion angle during COD manoeuvres at 90° in the non-preferred leg.

Conclusion In male recreational soccer players, sharper cutting angles place the knee at more risk for ACL injuries with little asymmetry between preferred and non-preferred limbs. Sharper angles of examination should be utilized in the evaluation of individuals.

9.3 Informed Consent Form

University of Salford MANCHESTER

Informed Consent Form

Study title:
Between-day reliability and agreement and relationship of kinematic, kinetic and isometric measurements during a sidestep 90 and 135 degree cutting tasks (CUT).

1. I have read the Participant Information Sheet (Participant Information Sheet - Version 4.0 - 01 February 2018) and have had opportunity to ask and have my questions answered satisfactorily. ☐

2. XXXXXXXXX, is a postgraduate research student at the University of Salford, has requested my participation in a research study. My involvement in the study and its purpose has been fully explained to me. ☐

3. I understand the requirements of the study and my involvement and the possible benefit of my participation in this research. ☐

4. I have been informed that I will be compensated a £20 voucher for my participation following the end of the sessions. ☐

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

6. I have been informed that any further questions that I have at any time concerning the research or my participation will be answered by XXXXXXX and I can contact him at his e-mail address (XXXXXXXXX@edu.salford.ac.uk). ☐

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

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

Name of participant:	**Signed:**	**Date:**
Name of researcher:	**Signed:**	**Date:**

One copy for participant and one copy for the researcher.

Informed Consent Form - Version 4.0 - 01 February 2018

9.4 Participant Information Sheet

University of Salford MANCHESTER

Participant Information Sheet

The participant is invited to take part in a research study which could provide important information for improving the assessment and prevention of knee injuries in sports. Please read the following information carefully.

What is the project all about?

The purpose of the study is to investigate the reliability of a test protocol which will measure the repeatability of data collection in a variety of sport tasks where most ACL non-contact injuries happen. This study will enable the development of a reliable test protocol and provide essential data for use in a future trial involving the same tasks.

Why have I been chosen?

The participant has been chosen to participate because the participant is:

1. The participant is physically active (attend at least 30 minutes of physical activity 3 times a week on a regular basis over the last 6 months).
2. The participant is a Staff/ Student at the University of Salford.
3. The participant doesn't have injuries in the lower limbs and able to bend and extend the lower limbs independently without aids.

What will I have to do?

- The experiment will include two sessions, each lasting one hour thirty minutes.
- At least a minimum period of 24 hours will be set, between giving the participant the information sheet and signing the consent form.
- To standardise participant clothing, each participant will be provided a pair of shorts and shoes.
- Each participant's age will be noted. Similarly, their weight and height will be measured to calculate their body mass index (BMI).
- The participant will be required to undergo a 5 minutes warm up. After which they will undertake the various change direction to 90° and 135° angles tasks to familiarise themselves with the experiment.
- The experiment will include two different stages. First stage: the participant's hip muscles power will be tested using hip extension, external rotation and abduction tests. Then, the participant will be required to have a 5 minutes rest.

- Second stage: a total of 40 reflective markers will be placed on the participant lower limb using hypo-allergic adhesive tape (see Figure 1).
- The participant will then be required to undertake 90° and 135° angles change direction to both sides (right and left). The testing will not involve any exertion that the participant is not accustomed with through the participant current physical levels.
- The participant will be asked to repeat the test seven days later, same day and time, at Human Performance Lab, Mary Seacole Building.

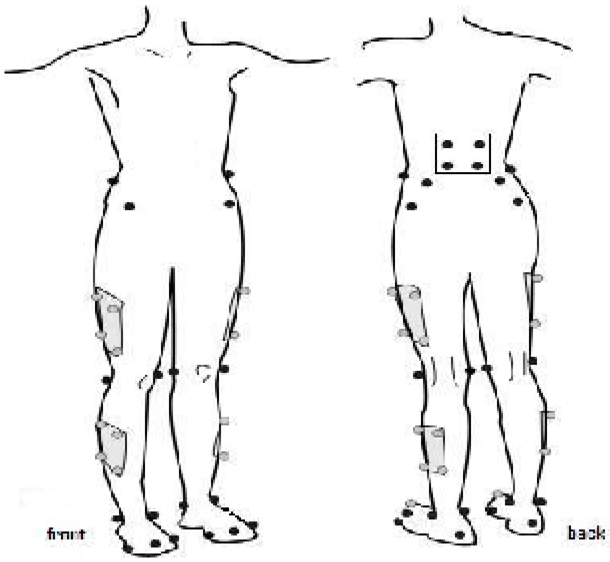

Figure 1 Markers position

Is there any risk involved?

There is no obvious risk to both the participant and researcher. However, if the participant experienced any pain during the experiment, they can contact the GP.

What if something goes wrong?

The university has insurance to cover against harm to you which may occur whilst you are taking part in this study. If you wish to complain, or if you have any concerns about any aspect of the way you have been treated during the study, you can contact the Chair of the Health Research Ethical Approval Panel Professor Susan McAndrew (Room MS1.91, Mary Seacole Building, Frederick Road Campus, University of Salford, Salford, M6 6PU. Tel: 0161 295 2278.

E: s.mcandrew@salford.ac.uk) who will be able to discuss this with you. If you do decide to take legal action, you may have to pay for this.

Who will see my details and results?

All the participant information will be strictly kept confidential. The participant identity will not be disclosed at any time. All data will be coded with a study number, so the participant information will remain anonymous and confidential, and will be stored on a computer, that is password-protected, accessed only by a researcher. The final results of this study will be available for the participant, and this study may be published.

Do I have to take part?

- The experiment will include two separate sessions, each lasting one hour thirty minutes. The participant will be asked to repeat the test seven days later, same day and time, at Human Performance Lab, Mary Seacole Building.
- The participant is free to decide whether or not to take part in this study. Moreover, the participant will be also free to withdraw from this study at any time without giving any reason.
- Please feel free to ask any further questions in the future about the project at any time.

For any further questions about the nature or demands of the project, please do not hesitate to contact me:

Mr. XXXXXXXXXXX
Email: XXXXXXXXX@edu.salford.ac.uk
Mobile number: 07XXXXXXXXXX
Address: XXXXXXXXXXXXXX

My supervisor contact information:

Professor Richard Jones PhD
Director of the Salford Gait Laboratory
Research Lead: Knee Biomechanics and Injury
Address: PO42 Brian Blatchford Building, University of Salford, M6 6PU
Telephone: (+44) 161 295 2295; Fax: (+ 44) 161 295 2432
Email: r.k.jones@salford.ac.uk

Thank you very much for taking the time to read this document and many thanks for your participation.

Participant Information Sheet - Version 4.0 - 01 February 2018

9.5 Results tables:

Study 1: Comparisons (mean ± SD) of hip and knee angles (degrees) and moments (Nm/kg) of preferred and non-preferred limb during 90° COD manoeuvre.

Variable	Preferred limb		Non-preferred limb		Raw p-value	Adjusted p-value[a]	ES
	Mean	SD	Mean	SD			
Kinematics							
Peak hip sagittal plane angle (°) at							
IC	39.67	7.39	42.77	8.13	0.01	0.001	0.47
PEKAM	41.44	7.44	44.10	7.97	0.02	0.002	0.41
PVGRF	42.28	7.56	45.06	8.22	0.02	0.002	0.43
PKVA	44.33	8.73	46.65	9.41	0.07	0.002	0.31
60 ms	46.06	8.36	49.29	9.81	0.02	0.002	0.42
Hip sagittal ROM angle (°) between IC and							
PEKAM	1.77	1.84	1.33	1.43	0.10	0.002	0.19
PVGRF	2.61	2.35	2.29	2.03	0.13	0.003	0.18
PKVA	4.66	4.45	3.88	4.56	0.40	0.007	0.10
60 ms	6.39	4.37	6.52	3.74	0.82	0.050	0.04
Peak hip frontal angle (°) at							
IC	-20.45	6.61	-18.31	7.26	0.09	0.002	0.30
PEKAM	-19.42	6.65	-17.33	7.01	0.08	0.002	0.28
PVGRF	-19.21	6.67	-17.15	7.04	0.09	0.003	0.27
PKVA	-17.97	6.83	-16.50	6.89	0.06	0.004	0.19
60 ms	-18.09	6.89	-16.21	6.95	0.09	0.003	0.24
Hip frontal ROM angle (°) between IC and							
PEKAM	1.04	1.05	0.99	1.31	0.38	0.006	0.10
PVGRF	1.24	1.24	1.16	1.54	0.44	0.010	0.09
PKVA	2.49	2.29	1.81	2.12	0.01	0.002	0.29
60 ms	2.36	1.96	2.10	2.58	0.11	0.003	0.19
Peak hip transverse plane angle (°) at							
IC	4.83	9.18	5.94	8.60	0.48	0.017	0.12
PEKAM	-0.71	9.01	1.17	9.09	0.51	0.004	0.20
PVGRF	-1.42	9.02	0.10	9.31	0.34	0.005	0.16
PKVA	-6.82	10.72	-2.97	9.58	0.04	0.002	0.36
60 ms	-4.20	9.01	-1.91	8.83	0.15	0.003	0.24
Hip transverse rotation ROM angle (°) between IC and							
PEKAM	-5.54	4.21	-4.77	3.19	0.75	0.006	0.04
PVGRF	-6.25	4.49	-5.83	3.32	0.65	0.025	0.08
PKVA	-11.65	7.24	-8.91	6.75	0.03	0.002	0.30
60 ms	-9.03	5.53	-7.85	4.62	0.30	0.004	0.18
Knee frontal plane angle (°)							
IC	1.89	3.92	2.44	4.04	0.42	0.008	0.14
Peak	-4.20	5.00	-3.16	4.87	0.21	0.003	0.21
ROM	-6.09	3.84	-5.60	2.71	0.47	0.013	0.12
Knee sagittal plane angle (°)							
IC	17.87	6.32	20.06	5.82	0.01	0.002	0.47
Peak	61.61	8.09	65.94	7.09	0.0012	0.0011	0.59

ROM	43.74	7.09	45.88	6.84	0.06	0.002	0.32
Moments							
PVGRF (*BW)	2.15	0.31	2.07	0.37	0.11	0.002	0.28
PEKAM (Nm/Kg)	1.23	0.57	1.33	0.70	0.32	0.005	0.17

* The mean difference is significant.
Standard deviation (SD); effect size (ES); body weight (BW); angle (°); peak external knee abduction moment (PEKAM); newton meter per kilogram (Nm/Kg); peak vertical ground reaction force (PVGRF); initial contact (IC); peak knee valgus angle (PKVA); millisecond (ms); peak knee flexion angle (PKFA).
Sign conventions shows the position of the joints as; hip flexion (+), hip Extension (-), hip abduction (-), hip adduction (+), hip internal rotation (+) and hip external rotation (-).
[a] By holm method.

Study 1: Comparisons (mean ± SD) of hip and knee angles (degrees) and moments (Nm/kg) of preferred and non-preferred limb during 135° COD manoeuvre.

Variable	Preferred limb		Non-preferred limb		Raw p-value	Adjusted p-value[a]	ES
	Mean	SD	Mean	SD			
Kinematics							
Peak hip sagittal plane angle (°) at							
IC	37.03	8.18	40.30	7.51	0.0012	0.0016	0.59
PEKAM	39.47	8.60	42.86	7.83	0.0043	0.0016	0.51
PVGRF	39.65	8.53	43.29	7.80	0.0015	0.0015	0.57
PKVA	42.75	10.20	45.30	9.57	0.10	0.0022	0.28
60 ms	43.67	10.28	47.37	9.33	0.01	0.0016	0.46
Hip sagittal ROM angle (°) between IC and							
PEKAM	2.44	2.46	2.55	2.39	1.00	0.0500	0.00
PVGRF	2.61	2.32	2.99	2.39	0.79	0.0045	0.03
PKVA	5.72	5.66	5.00	5.59	0.43	0.0024	0.09
60 ms	6.64	4.87	7.07	4.45	0.67	0.0033	0.07
Peak hip frontal angle (°) at							
IC	-21.94	7.56	-21.88	6.81	0.97	0.0125	0.01
PEKAM	-20.79	7.77	-20.78	6.58	1.00	0.0250	0.00
PVGRF	-20.77	7.81	-20.72	6.64	0.97	0.0100	0.01
PKVA	-19.60	8.15	-20.18	6.42	0.67	0.0036	0.07
60 ms	-20.23	7.89	-20.29	6.66	0.96	0.0083	0.01
Hip frontal ROM angle (°) between IC and							
PEKAM	1.15	1.44	1.10	1.18	0.93	0.0053	0.01
PVGRF	1.16	1.49	1.17	1.24	0.75	0.0042	0.04
PKVA	2.33	3.37	1.71	1.79	0.50	0.0025	0.08
60 ms	1.71	2.21	1.60	1.73	0.88	0.0050	0.02
Peak hip transverse plane angle (°) at							
IC	7.82	9.04	6.94	10.02	0.58	0.0026	0.09
PEKAM	0.13	8.71	0.06	9.93	0.96	0.0071	0.01
PVGRF	-0.17	8.68	-0.37	10.29	0.89	0.0056	0.02
PKVA	-6.73	11.02	-3.28	9.24	0.03	0.0018	0.38
60 ms	-2.30	9.43	-1.59	9.59	0.65	0.0031	0.08

Hip transverse rotation ROM angle (°) between IC and							
PEKAM	-7.69	6.30	-6.88	4.67	0.65	0.0029	0.05
PVGRF	-7.99	6.35	-7.31	4.77	0.62	0.0028	0.06
PKVA	-14.55	7.78	-10.22	6.82	0.0013	0.0015	0.58
60 ms	-10.12	6.41	-8.53	4.87	0.19	0.0023	0.22
Knee frontal plane angle (°)							
IC	-0.48	4.26	0.84	3.90	0.08	0.0019	0.32
Peak	-5.87	5.78	-4.53	5.02	0.08	0.0020	0.30
ROM	-5.39	4.07	-5.37	3.39	0.98	0.0167	0.01
Knee sagittal plane angle (°)							
IC	18.86	4.97	19.16	5.41	0.73	0.0030	0.06
Peak	63.02	8.83	67.26	7.74	0.01	0.0017	0.46
ROM	44.17	7.90	48.10	7.25	0.01	0.0017	0.45
Moments							
PVGRF (*BW)	2.16	0.35	2.07	0.36	0.09	0.0021	0.29
PEKAM (Nm/Kg)	2.34	1.11	2.04	1.11	0.03	0.0019	0.37

* The mean difference is significant.
Standard deviation (SD); effect size (ES); body weight (BW); angle (°); peak external knee abduction moment (PEKAM); newton meter per kilogram (Nm/Kg); peak vertical ground reaction force (PVGRF); initial contact (IC); peak knee valgus angle (PKVA); millisecond (ms); peak knee flexion angle (PKFA).
Sign conventions shows the position of the joints as; hip flexion (+), hip Extension (-), hip abduction (-), hip adduction (+), hip internal rotation (+) and hip external rotation (-).
[a] By holm method.

Study 2: Comparisons (mean ± SD) of hip and knee angles (degrees) and moments (Nm/kg) of 90° and 135° COD manoeuvres.

Variable	90° COD		135° COD		Raw p-value	Adjusted p-value[a]	ES
	Mean	SD	Mean	SD			
Kinematics							
Peak hip sagittal plane angle (°) at							
IC	39.67	7.39	37.03	8.18	0.005	0.002	0.50
PEKAM	41.44	7.44	39.47	8.60	0.04	0.002	0.37
PVGRF	42.28	7.56	39.65	8.53	0.008	0.002	0.47
PKVA	44.33	8.73	42.75	10.20	0.20	0.004	0.22
60 ms	46.06	8.36	43.67	10.28	0.04	0.002	0.35
Hip sagittal ROM angle (°) between IC and							
PEKAM	1.77	1.84	2.44	2.46	0.06	0.003	0.22
PVGRF	2.61	2.35	2.61	2.32	0.86	0.025	0.02
PKVA	4.66	4.45	5.72	5.66	0.43	0.007	0.09
60 ms	6.39	4.37	6.64	4.87	0.71	0.013	0.06
Peak hip frontal angle (°) at							
IC	-20.45	6.61	-21.94	7.56	0.04	0.003	0.35
PEKAM	-19.42	6.65	-20.79	7.77	0.06	0.003	0.32
PVGRF	-19.21	6.67	-20.77	7.81	0.03	0.002	0.37
PKVA	-17.97	6.83	-19.60	8.15	0.05	0.002	0.34
60 ms	-18.09	6.89	-20.23	7.89	0.005	0.002	0.50
Hip frontal ROM angle (°) between IC and							
PEKAM	1.04	1.05	1.15	1.44	0.69	0.010	0.05
PVGRF	1.24	1.24	1.16	1.49	0.22	0.005	0.15
PKVA	2.49	2.29	2.33	3.37	0.29	0.004	0.12
60 ms	2.36	1.96	1.71	2.21	0.02	0.002	0.27
Peak hip transverse plane angle (°) at							
IC	4.83	9.18	7.82	9.04	0.005	0.002	0.50
PEKAM	-0.71	9.01	0.13	8.71	0.45	0.008	0.13
PVGRF	-1.42	9.02	-0.17	8.68	0.25	0.006	0.19
PKVA	-6.82	10.72	-6.73	11.02	0.95	0.050	0.01
60 ms	-4.20	9.01	-2.30	9.43	0.11	0.004	0.27
Hip transverse rotation ROM angle (°) between IC and							
PEKAM	-5.54	4.21	-7.69	6.30	0.03	0.002	0.26
PVGRF	-6.25	4.49	-7.99	6.35	0.05	0.003	0.23
PKVA	-11.65	7.24	-14.55	7.78	0.05	0.003	0.34
60 ms	-9.03	5.53	-10.12	6.41	0.36	0.005	0.14
Knee frontal plane angle (°)							
IC	1.89	3.92	-0.48	4.26	0.000*	0.002	1.09
Peak	-4.20	5.00	-5.87	5.78	0.005	0.002	0.50
ROM	-6.09	3.84	-5.39	4.07	0.20	0.004	0.22
Moments							
PVGRF (*BW)	2.15	0.31	2.16	0.35	0.76	0.017	0.05
PEKAM (Nm/Kg)	1.23	0.57	2.34	1.11	0.000*	0.002	1.04

Study 3: Pearson's correlation (R) and p value (P) between hip motion and kinematic/kinetic data of the knee joint during 90° COD manoeuvres.

Variable	Peak knee valgus angle		Knee valgus angle dis.		Peak EKAM		Knee valgus angle at IC	
	R	P	R	P	R	P	R	P
Peak hip sagittal plane angle (°) at								
IC	0.00	0.98	-0.27	0.12	-0.23	0.18	0.27	0.12
PEKAM	0.05	0.79	-0.24	0.15	-0.12	0.48	0.30	0.08
PVGRF	0.09	0.59	-0.19	0.25	-0.15	0.38	0.31	0.07
PKVA	0.15	0.39	-0.26	0.12	-0.13	0.45	**0.45**	0.01
60 ms	0.12	0.48	-0.19	0.26	-0.05	0.77	**0.34**	**0.04**
Hip sagittal ROM angle (°) between IC and								
PEKAM	0.17	0.31	0.08	0.63	**0.42**	**0.01**	0.14	0.42
PVGRF	0.29	0.09	0.21	0.21	0.23	0.19	0.15	0.37
PKVA	0.28	0.09	-0.08	0.66	0.12	0.48	**0.44**	0.01
60 ms	0.22	0.19	0.08	0.63	0.29	0.09	0.20	0.24
Peak hip frontal angle (°) at								
IC	0.14	0.42	**0.46**	**0.00**	**-0.37**	**0.03**	-0.28	0.10
PEKAM	0.14	0.42	**0.49**	**0.00**	-0.30	0.07	-0.31	0.07
PVGRF	0.15	0.37	**0.50**	**0.00**	**-0.33**	**0.05**	-0.29	0.09
PKVA	0.22	0.20	**0.47**	**0.00**	-0.32	0.06	-0.20	0.30
60 ms	0.23	0.17	**0.50**	**0.00**	-0.30	0.08	-0.18	0.25
Hip frontal ROM angle (°) between IC and								
PEKAM	0.02	0.89	0.22	0.20	**0.39**	**0.02**	-0.18	0.28
PVGRF	0.10	0.55	0.21	0.22	0.18	0.30	-0.07	0.67
PKVA	0.26	0.13	0.05	0.75	0.11	0.51	0.28	0.10
60 ms	**0.36**	**0.03**	0.21	0.23	0.19	0.26	0.25	0.14
Peak hip transverse plane angle (°) at								
IC	0.31	0.07	0.20	0.24	0.26	0.12	0.20	0.24
PEKAM	**0.43**	**0.01**	0.31	0.07	0.12	0.47	0.24	0.15
PVGRF	**0.44**	**0.01**	0.33	0.05	0.16	0.36	0.24	0.16
PKVA	**0.41**	**0.01**	**0.53**	**0.00**	0.12	0.48	-0.01	0.98
60 ms	**0.37**	**0.03**	**0.38**	**0.02**	0.14	0.42	0.09	0.59
Hip transverse rotation ROM angle (°) between IC and								
PEKAM	0.24	0.16	0.22	0.20	**-0.36**	**0.03**	0.09	0.61
PVGRF	0.25	0.14	0.25	0.14	-0.21	0.21	0.07	0.68
PKVA	0.21	0.22	**0.54**	**0.00**	-0.15	0.39	-0.26	0.12
60 ms	0.08	0.64	0.29	0.09	-0.21	0.22	-0.18	0.30

Significant correlations are noted in bold.

Peak external knee abduction moment (PEKAM); peak vertical ground reaction force (PVGRF); initial contact (IC); peak knee valgus angle (PKVA); millisecond (ms); peak knee flexion angle (PKFA).

Study 3: Pearson correlation's (R) and p value (P) between hip motion and kinematic/kinetic data of the knee joint during 135° COD manoeuvres.

Variable	Peak knee valgus angle		Knee valgus angle dis.		Peak EKAM		Knee valgus angle at IC	
	R	P	R	P	R	P	R	P
Peak hip sagittal plane angle (°) at								
IC	0.00	1.00	-0.24	0.16	-0.11	0.54	0.23	0.19
PEKAM	0.07	0.68	-0.25	0.14	0.06	0.73	**0.34**	**0.04**
PVGRF	0.09	0.62	-0.23	0.17	0.02	0.91	**0.34**	**0.04**
PKVA	-0.05	0.76	**-0.43**	**0.01**	0.08	0.64	**0.34**	**0.04**
60 ms	0.11	0.52	-0.31	0.06	0.05	0.75	**0.45**	**0.01**
Hip sagittal ROM angle (°) between IC and								
PEKAM	0.25	0.14	-0.09	0.58	**0.56**	**0.00**	**0.43**	**0.01**
PVGRF	0.32	0.05	-0.03	0.88	**0.45**	**0.01**	**0.46**	**0.00**
PKVA	-0.09	0.59	**-0.43**	**0.01**	0.30	0.08	0.28	0.09
60 ms	0.23	0.17	-0.26	0.12	0.29	0.08	**0.57**	**0.00**
Peak hip frontal angle (°) at								
IC	-0.07	0.70	0.31	0.06	**-0.40**	**0.01**	**-0.39**	**0.02**
PEKAM	-0.02	0.92	0.32	0.06	**-0.36**	**0.03**	-0.32	0.05
PVGRF	-0.02	0.91	0.31	0.07	**-0.36**	**0.03**	-0.32	0.05
PKVA	-0.04	0.83	0.26	0.12	-0.22	0.19	-0.30	0.08
60 ms	-0.03	0.86	0.28	0.10	-0.29	0.09	-0.31	0.07
Hip frontal ROM angle (°) between IC and								
PEKAM	0.26	0.13	0.07	0.67	0.16	0.37	0.28	0.09
PVGRF	0.23	0.18	0.05	0.79	0.15	0.37	0.27	0.11
PKVA	0.06	0.73	-0.07	0.70	**0.37**	**0.03**	0.15	0.40
60 ms	0.12	0.50	-0.08	0.65	**0.35**	**0.04**	0.23	0.17
Peak hip transverse plane angle (°) at								
IC	0.20	0.24	0.31	0.07	-0.04	0.82	-0.02	0.90
PEKAM	0.20	0.25	0.22	0.19	-0.09	0.62	0.06	0.74
PVGRF	0.21	0.21	0.22	0.19	-0.06	0.72	0.07	0.67
PKVA	**0.35**	**0.04**	**0.47**	**0.00**	-0.16	0.35	0.03	0.88
60 ms	0.25	0.14	**0.40**	**0.02**	-0.19	0.26	-0.04	0.81
Hip transverse rotation ROM angle (°) between IC and								
PEKAM	-0.01	0.94	-0.13	0.44	-0.06	0.71	0.11	0.52
PVGRF	0.01	0.97	-0.13	0.45	-0.03	0.87	0.13	0.44
PKVA	0.26	0.12	0.31	0.07	-0.18	0.28	0.06	0.72
60 ms	0.08	0.62	0.15	0.38	-0.23	0.18	-0.03	0.87

Significant correlations are noted in bold.
Peak external knee abduction moment (PEKAM); peak vertical ground reaction force (PVGRF); initial contact (IC); peak knee valgus angle (PKVA); millisecond (ms); peak knee flexion angle (PKFA).