# Evaluation of Movement Variability and Knee Biomechanics During Functional Sporting Tasks in Healthy Individuals and Individuals Following Anterior Cruciate Ligament Injury (ACL) and ACL Reconstruction.

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## List of Abbreviations

| Abbreviation | Meaning |
|---|---|
| **ACL** | Anterior Cruciate Ligament |
| **MV** | Movement Variability |
| **CV** | coefficient of variation |
| **SLL** | Single-Leg-Landing |
| **SLS** | Single-Leg-Squat |
| **ACL-RC** | Anterior Cruciate Ligament Reconstruction/Reconstructed |
| **ACL-DF** | Anterior Cruciate Ligament Deficiency/Deficient |
| **GRF** | Ground Reaction Force |
| **PTOA** | Post-Traumatic Osteoarthritis |
| **OA** | Osteoarthritis |
| **COM** | Centre of Mass |
| **SD** | Standard Deviation |
| **GRF** | Ground Reaction Force |
| **3D** | Three Dimension |
| **2D** | Two Dimension |

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# Abstract

## Background

Injury to the anterior cruciate ligament (ACL) is a serious problem which can occur during some sporting activities. Knee biomechanics and movement variability (MV) comprise two very important aspects relating to injury risk, where it has been suggested that the latter may play a role in ACL injury.

## Aim

The aims of this thesis were: to investigate MV and knee biomechanics during four common functional sporting tasks in a healthy cohort, as well as in individuals following ACL injury and following ACL reconstruction (ACL-RC); to investigate the effect of leg dominance and the effect of the individual tasks on MV and knee biomechanics; to investigate the relationship between MV and knee biomechanical risk factors; and to investigate differences in MV and knee biomechanics between all three groups.

## Methods

Kinematic and kinetic data were obtained using 3D motion analysis and force platforms. Following examination of the reliability of 3D biomechanical outcome measures and MV outcome measures participants (45 healthy, 20 ACL deficient (ACL-DF) and 20 unilateral ACL reconstructed (ACL-RC) males) performed four different sporting tasks: running, cutting to 90 degrees, a single leg landing (SLL) and a single leg squat (SLS) with both limbs. The MV was assessed using coefficient of variation (CV) (the second order version). Differences were assessed between the two legs in each group, as well as differences between groups. The relationship between MV and knee biomechanical risk factors was also examined.

## Results

### Healthy

In terms of healthy group, the reference values for MV and knee biomechanics were provided. In addition, significant differences in MV and knee biomechanics were found to occur between tasks. Of all four tasks, cutting to 90 degrees demonstrated the highest values in terms of knee biomechanical risk factors (knee valgus moment), which indicates that it is the riskiest task, following by SLL. However, cutting to 90 degrees task demonstrated lower MV, as compared to the other tasks. In addition, no significant differences emerged in MV and knee biomechanics between dominant and non-dominant legs during all four tasks. A significant negative relationship was found to occur between MV and peak knee valgus moment during all four functional sporting tasks.

### ACL-DF and ACL-RC

In terms of the ACL-DF group, the injured limb demonstrated a significantly lower peak knee extension moment, peak knee flexion angle and peak GRF than the uninjured one. However, it demonstrated greater MV than the uninjured one. Interestingly, the ACL-RC group reported similar findings to the ACL-DF cohort, where the reconstructed limb demonstrated a significantly lower peak knee extension moment, lower peak knee flexion angle and peak GRF than the uninjured one. However, the injured side demonstrated greater MV than the uninjured one.

### Between three groups

In the comparisons between groups both the ACL-RC and ACL-DF groups demonstrated a significantly lower peak knee extension moment as compared to the healthy control group. They also showed a lower peak knee flexion angle than the control group during running, SLL and cutting tasks. In addition, both ACL-RC and ACL-DF groups demonstrated lower MV than the healthy control group. Importantly, the uninjured limb in the ACL-RC group demonstrated the greatest between-group knee valgus moment during SLL and cutting tasks.

**Conclusion**

ACL injury can significantly affect MV and knee biomechanics. In addition, MV should be considered when developing ACL injury risk mitigation and ACL reconstruction rehabilitation RTS programmes. As both the ACL-DF and ACL-RC cohorts displayed practically identical impairments in MV and knee biomechanics, this raises additional questions regarding rehabilitation and RTS programmes. The uninjured limb in the ACL-RC individuals reported the greatest knee valgus moment, which places it at higher risk of sustaining a second injury and, therefore, must be targeted during rehabilitation. These findings will be of assistance in the development of rehabilitation and RTS programmes for ACL-injured patients who either choose surgical or conservative treatment options. In addition, they will be of assistance when developing ACL injury/reinjury risk mitigation programmes.

# Chapter 1

## 1.1 Introduction

ACL injuries have been identified as one of the most frequently occurring types of knee injury. Moses, Orchard and Orchard (2012) calculated the median incidence of ACL injury to be generally in the region of 0.03% per person annually, while this figure can escalate to 3.7% among a professional athlete cohort. A United Kingdom-based study spanning a twenty-year period from 1997 to 2017 found that the age-standardised and sex-standarised rate of ACL reconstruction due to ACL injuries had risen twelvefold, while increases of 2.4 were recorded for meniscal repair. Significant costs are usually associated with this form of injury, which can require a range of interventions including imaging, reconstructive surgery, postoperative bracing and rehabilitation, with the average annual outlay on ACL injuries amounting to in excess of two billion dollars (Gottlob et al., 1999; Malek, DeLuca, Kunkle, & Knable 1996). In addition, other more intangible costs need to be considered, most notably the psychological impact of injury, and this is particularly true in the case of elite athletes. In most cases, an athlete will have to sit out 6–9 months of a competition when they sustain an ACL injury. This can have very significant repercussions for them, depending on their level of engagement in competitive play for monetary gain.

ACL injury can be classified into different risk factors. Of these, biomechanical risks factors are most important, and several of them have been identified in the literature. Furthermore, movement variability (MV) is another important biomechanical-related issue, although variability within subjects cannot be avoided. In addition, it has been suggested that MV is a key element in that it can increase the likelihood of injury occurring (Baida, Gore, Franklyn-Miller, & Moran, 2018; James, 2004; Konradsen, 2002; Konradsen & Voigt, 2002). In particular, it has also been proposed to correlate with risk of further musculoskeletal injuries, such as ACL rupture (McLean et al., 1999), lower extremity overuse injury (James et al., 2000) and patellofemoral pain (Heiderscheit, Hamill, & van Emmerik, 2002). Based on the findings of previously cited research conducted on MV, coupled with its potential impact on musculoskeletal injuries, it is essential that any studies undertaken in this field pay particular attention to the knee joint, given that it is a site prone to ACL injury. In addition, the examination of all biomechanical-related variables is warranted, especially those identified as risk factors for injury, for example, the knee valgus angle, the knee valgus moment, the knee flexion angle, and ground reaction force (GRF). While differences can emerge between all these risk factors, nevertheless, it is important to monitor any significant task-related movement change, as it may elevate the risk of sustaining an ACL injury.

The most frequently investigated tasks in sport injuries are landings, squatting, running and sidestep cutting. Most episodes of non-contact ACL injuries occur during cutting and landing tasks. Other less complex tasks, such as running and SLS can also be included as they assist in acquiring an understanding of biomechanics and MV when performing both complex and less complex tasks. All these tasks are investigated in this thesis. Another important issue worthy of consideration is leg dominance, as it is believed that it may exert an effect on biomechanics and MV, and, consequently, it has been included in this investigation.

According to Munro, Herrington and Carolan (2012), three-dimensional (3D) motion analysis is regarded as a gold standard method to adopt in the assessment of joint kinematics and kinetics. It can facilitate the measurement of movement along three different planes, namely the sagittal, frontal and transverse. Kinetic data (moments and GRF) can also be compiled using 3D motion analysis. Hence, this data is often used to examine lower limb biomechanics (Blackburn & Padua, 2008; Ford, Myer, & Hewett, 2003; Hewett et al., 2005). Consequently, it was selected for the purposes of the current study to measure biomechanical data. While its advantages are clearly evident, nevertheless, in reality, it is prohibitively expensive and, as a result, is not widely used in general population sport screening programmes or in other clinical environments. Furthermore, additional spatial and temporal costs can be incurred when using 3D motion analysis. Less expensive, more user-friendly and transportable options are typically selected instead, most notably, 2D analysis. This software has been employed to measure knee valgus angles, including athletes in the sporting world, as well as members of the public who have sustained an injury.

With regard to 3D motion analysis, it is important to test its reliability as this method can be subject to measurement errors derived from different sources. Rater is one of the common sources of errors, due to marker placement error, which is known as within-assessor error. Throughout the literature, a large body of studies have investigated the reliability of 3D motion analysis in terms of either its within-day or between-day effects under different conditions. Within-day reliability has been shown to demonstrate better reliability than between-day reliability (Kadaba et al., 1989). However, a number of studies have found that an error in marker placement impacts significantly upon between-day reliability (Ferber et al., 2002; Ford, Myer, & Hewett, 2007; Queen et al., 2006). Furthermore, Cappozzo, Catani, Leardini, Benedetti and Croce (1996) identified skin movement artefact as another source of error. Therefore, a reliability study will be conducted in the first instance, prior to embarking on the main study.

It is widely acknowledged that three population types exist in relation to ACL injury, namely: healthy, ACL deficient (ACL-DF) and ACL reconstructed (ACL-RC). The inclusion of a healthy population is important in order to establish reference values for MV and knee biomechanics. In addition, investigations were undertaken to establish how task and leg dominance factors can affect MV and knee biomechanics. In relation to the ACL-DF population, two different treatments options can be provided, namely conservative and surgical. The final population to be investigated in this thesis was ACL-RC. Unfortunately, the reinjury rate in this cohort is high in either the same knee or the contralateral side (Webster & Hewett, 2019; Dekker et al., 2017; Morgan et al., 2016; Paterno et al., 2014; Wiggins et al., 2016).

Comparisons between the ACL injured populations and healthy controls are also very important as they can improve our understanding of the way in which the biomechanical characteristics of injured populations compare to healthy ones. Such comparisons can help researchers and clinicians to identify the deficits in injured population groups and to subsequently specifically target them in their treatment plans. In addition, undertaking comparisons of ACL-RC and ACL-DF cohorts is important in that they can provide broad information on the effects of both injury and surgery on knee biomechanics and MV. These kinds of investigations are beneficial, especially when the contralateral side is included in the comparisons, in gaining an understanding of the effect of ACL injury/surgery on the contralateral knee, given that the reinjury rate is high on both sides (Webster & Hewett, 2019; Dekker et al., 2017). Therefore, comparisons between the three groups will be conducted in this thesis.

Gender is another factor that should be taken into consideration during the investigation of lower biomechanics. Several recent studies have revealed significant differences between males and females during different sporting tasks. In particular, significant differences were reported during landing tasks (Hughes, 2019; Weinhandl, Irmischer, Sievert, & Fontenot, 2017;

Holden, Boreham, & Delahunt, 2016; Jenkins et al., 2017). These findings were similar to those reported during cutting tasks (Sinclair et al., 2019; Weinhandl, Irmischer, Sievert, & Fontenot, 2017; James et al., 2004; Ford, Myer, Toms, & Hewett, 2005) and SLS (Dwyer, Boudreau, Mattacola, Uhl, & Lattermann, 2010; Graci, Van Dillen, & Salsich, 2012; Weeks, Carty, & Horan, 2015; Yamazaki, Muneta, Ju, & Sekiya, 2010; Zeller, McCrory, Ben Kibler, & Uhl, 2003). Therefore, a male-only sample was included.

# Chapter 2

## 2 Scope of the thesis and its aims

### 2.1 Scope of the thesis

The main focus of this thesis was to investigate movement variability (MV) and knee biomechanics among all anterior cruciate ligament (ACL) injury-related populations namely; healthy, ACL deficient (ACL-DF) and ACL reconstructed (ACL-RC) cohorts. Furthermore, it aimed to advance the current knowledge of MV when associated with non-contact ACL injury. In the literature, MV has been identified as a key element in increasing the likelihood of injury occurring (Baida, Gore, Franklyn-Miller, & Moran, 2018; James, 2004; Konradsen, 2002; Konradsen & Voigt, 2002). In addition, it has been suggested that increased variability is associated with greater risk of further musculoskeletal injuries, such as ACL rupture (McLean et al., 1999). In order to investigate MV at the knee joint, it is important to consider knee biomechanics, as it provides objective information on knee motion from which we can assess MV. Firstly, the healthy group was included because they are considered to be at risk of sustaining this injury and to achieve different sub-aims, the first of which is to provide important reference values for MV and knee biomechanics. These values can assist in understanding MV and knee biomechanics in healthy cohorts and will allow clinicians and researchers to carry out biomechanical assessments, having gained a solid understanding of how these variables typically present in healthy individuals. It will also assist them to differentiate between normal and abnormal characteristics, in particular, in relation to the biomechanical variables that are considered to be risk factors for knee injury.

Another reason for including a healthy population in this thesis is to establish whether MV and knee biomechanics can be affected by either leg dominance or the task. In the literature, leg dominance has been widely investigated in different sporting tasks, as some researchers consider that it may be an influential factor in knee biomechanical risk factors for ACL injury. The effect of the task is also important as different tasks can demonstrate different knee biomechanics. These differences can occur between tasks in biomechanical risk factors for ACL injury, which can lead to increased risk of this injury. The most widely investigated tasks in the literature are landing, sidestep cutting, single leg squats (SLS) and running and, accordingly, all four tasks have been included in this thesis. Noncontact ACL injury commonly occurs during sidestep cutting and single leg landing (SLL) and, therefore, this is the rationale for the inclusion of these tasks in the study. Regarding running and SLS, it is uncommon to see this type of injury occurring during these two tasks and in the literature, they have been widely investigated by researchers for a number of different reasons. The slower motions involved in squats enable the visualisation of inappropriate motions with greater clarity, and they can offer immediate remedial suggestions. In addition, the squat task requires more control of both movement and load transmission, which can provide greater insights into neuromuscular coordination. The running task has also been the subject of extensive research, as this task is integral to most sporting activities. In addition to important complex tasks, the less complex tasks, such as forward running, are also crucial, as they allow clinicians and researchers to undertake a comparative analysis between where injuries occur when performing risky tasks and where injuries do not occur when carrying out non-risky tasks. Such comparisons can assist in developing a greater understanding of the mechanisms surrounding noncontact ACL injury, as well as helping to address the biomechanical risk factors associated with this injury.

The final important element to consider when including healthy individuals in this thesis is to determine whether a relationship exists between MV and biomechanical risk factors. Such an investigation is crucial in establishing the role MV plays in biomechanical risk factors for ACL injury. Furthermore, gaining such knowledge will aid clinicians and researchers in the development of ACL injury mitigation programmes. Consequently, they can target MV in order to minimise the risk of this injury occurring. In light of all sub-aims mentioned above, the importance of including a healthy population is clearly obvious and, therefore, Chapters 5 and 6 are exclusively dedicated to investigating this cohort in order to achieve all the aims outlined.

The current thesis initially investigated healthy individuals, as they are considered to be at risk of sustaining this injury. It subsequently investigated ACL-DF individuals. In the literature, to the best of the author's knowledge, several studies have investigated MV in a lower limb-injured population. However, of the few that have studied MV in an ACL-DF population, they tended to have involved less complex sporting tasks (walking and running). It is known that ACL injury can be treated conservatively or surgically. Hence, it is important to investigate both treatments so as to determine if they can be of assistance to each group. Chapter 6 exclusively investigates an ACL-DF cohort. It is widely known that ACL injury has some consequences for the injured individuals. Several studies have reported that ACL-injured individuals demonstrate impairments in knee biomechanics during different tasks, such as walking (Ismail et al., 2015; Ferber et al., 2002; Hart, Ko, Konold, & Pietrosimione, 2010; Lewek et al., 2002; Roberts et al., 1999; Rudolph et al., 2001; Chmielewski et al., 2001); running (Waite et al., 2005; Lewek et al., 2002; Rudolph et al., 2001; Chmielewski et al., 2001); cutting (Waite et al., 2005) and SLS tasks (Yamazaki et al., 2010). In terms of movement variability (MV), according to a recent systematic review undertaken by Baida et al. (2018), there is an overall trend towards greater MV in injured groups. Acquiring an understanding of these impairments in both knee biomechanics and MV is important in order to be cognisant of the effects of ACL injury on movement. Such knowledge can subsequently assist in targeting

MV and biomechanical deficits in the development of rehabilitation programmes when conservative treatment is considered. In addition, it will assist clinicians and researchers when developing ACL reinjury mitigation programmes, as it will enable them to minimise the risk of further injury occurring either on the affected side or the contralateral side.

A significant number of ACL-injured individuals undergo a surgical treatment known as ACL reconstruction. The current thesis has included this third population group following on from the ACL-DF cohort where Chapter 8 exclusively focuses on this group. In the literature, this population has been the subject of more studies than ACL-DF, as a significant number of researchers and clinicians regard this type of treatment as gold standard for ACL injury, despite the debate surrounding this topic, which will be discussed later in the thesis. This population has been included for two important reasons. Firstly, it is crucial that clinicians and researchers are familiar with biomechanics and MV for the ACL-RC knee as compared to the uninjured side. Such comparisons will contribute to the current knowledge of the differences between the two limbs. It will also assist in establishing whether biomechanical deficits exist in the reconstructed knee as compared to the uninjured knee, so as to be able to target the former during the development of ACL-RC rehabilitation programmes. Secondly, it will also provide information on MV and knee biomechanics for the uninjured limb. Unfortunately, the incidence of ACL reinjury in ACL-RC individuals is high, especially on the contralateral side. Therefore, the investigation of the uninjured limb in ACL-RC individuals will assist in the development of ACL reinjury mitigation programmes for the affected side or for the uninjured side. In the literature, to the best of the author’s knowledge, MV has been investigated in few studies and this has not extended to include all four tasks, which will be explained later. Therefore, investigating MV and knee biomechanics for an ACL-RC group is one of the main aims of this thesis and it will be the focus of attention in Chapter 8.

Furthermore, in order to broaden the scope of our understanding of MV and knee biomechanics in all ACL injury-related populations, analysis of the differences between all groups should be conducted which will be presented in Chapter 9, which exclusively investigates the differences between all three groups. In the literature, to the best of the author's knowledge, several studies have investigated differences in knee biomechanics between ACL-DF and healthy controls. The researchers in these studies have compared the ACL-DF limb to the healthy controls during less complex tasks (walking and running), which will be discussed in detail later in the thesis. However, none to date have compared the uninjured limb to the healthy controls. Undertaking such an investigation is very important in understanding why reinjury in the contralateral side is high. In addition, including complex tasks in the comparisons, such as sidestep cutting and SLL is important for this group, as it will provide biomechanical information following injury during high-load sporting activities, which they are expected to perform when they return to sport. Comparing MV and knee biomechanics for the ACL-DF to those of the healthy controls is essential to determine the impairments in MV and biomechanics in ACL-DF individuals so as to be able to specifically target them during the development of rehabilitation programmes, when conservative treatment is the preferred choice.

In line with this, comparing ACL-RC individuals to healthy controls is also important in understanding MV and knee biomechanics following this surgical procedure. Such investigations can provide information on how normal knee biomechanics and MV manifest in ACL-RC individuals. It can also determine whether impairments in knee biomechanics and MV occur in ACL-RC as compared to healthy controls. Such information is important in ascertaining how effective the rehabilitation programme is, while also assisting in improving existing ACL-RC rehabilitation programmes. In the literature, comparing the biomechanics of ACL-RC individuals to healthy controls has attracted significant interest from researchers during different sporting tasks. However, to the best of the author's knowledge, no study to date has compared the knee biomechanics of the uninjured knee to the healthy controls during the four sporting tasks. In terms of MV, to the best of the author's knowledge, while few studies have investigated MV in an ACL-RC cohort, none have examined all four sporting tasks. Furthermore, none of them have compared MV for the uninjured limb to those of healthy controls. The rationale for comparing knee biomechanics and MV for the uninjured limb to those of healthy controls is to establish whether knee biomechanics and MV impairments occur following surgery, as the incidence of reinjury in the contralateral side is high. Thus, such an investigation will be of assistance during the development of ACL reinjury mitigation programmes for the uninjured limb, as they will help to reduce the risk of ACL reinjury in the contralateral side.

Furthermore, the pairwise comparisons outlined in Chapter 9 also include those between ACL-RC and ACL-DF cohorts. Undertaking a comparative analysis between these two population groups can enhance our understanding of the differences between ACL-RC and ACL-DF populations. Such investigations are very important in understanding the effects of surgical treatment on those who have undergone surgery, as compared to those who have not. As ACL reconstruction is believed to improve knee stability and function, the ACL-RC should demonstrate knee biomechanics and MV values close to normal and should differ from ACL-DF. Unfortunately, to the best of the author's knowledge, no published study to date has compared biomechanics or MV between ACL-RC and ACL-DF during the four sporting tasks. In addition, the thesis compares the uninjured limb between the ACL-DF and the ACL-RC groups, as this is an interesting topic to examine in terms of risk of reinjury on the contralateral side. This investigation can aid our understanding of whether surgery can influence knee biomechanics and MV on the contralateral side, as the risk of reinjury on this side is high.

Accordingly, this information will help to enhance ACL reinjury mitigation programmes for the uninjured limb, which will result in minimising the risk of ACL reinjury on the uninjured side.

## 2.2 Aims of the thesis

There are four main aims of this thesis which have been outlined and the importance of each has been highlighted. In addition, all the sub-aims emanating from them have also been explained above. These main aims have been set out as follows, based on the population type, as well as exploring the different combinations of all three groups:

1) Examine the reliability of 3D motion analysis as a measure of knee biomechanics and CV as a measure of MV during four common sporting tasks.
2) Investigate MV and knee biomechanics in healthy population to establish reference values for MV and knee biomechanics and to establish the relationship between MV and knee biomechanical risk factors of ACL injury during these four tasks.
3) Investigate MV and knee biomechanics in ACL-DF and ACL-RC populations during the four tasks.
4) Compare MV and knee biomechanics between the three groups during the four sporting tasks.

## 2.3 Primary research questions

Based on the main aims of the thesis and all sub-aims emanating from them, seven questions have been set out as main questions to address these aims as follows:

1. What are the typical MV values in healthy individuals during four common sporting tasks: running, cutting to 90 degrees, SLL and SLS, as measured using 3D motion

analysis across knee kinematics and kinetics?

2. Does leg dominance affect MV and knee biomechanics?
3. Do the individual tasks affect MV and knee biomechanics?
4. Is increased or decreased variability associated with biomechanical risk factors of ACL injury?
5. Do any differences occur in MV and knee biomechanics between deficient knees and contralateral knees in individuals who have ACL-DF?
6. Do any differences occur in MV and knee biomechanics between reconstructed knees and contralateral knees in individuals who have undergone ACL-RC?
7. Do any differences occur in MV and knee biomechanics at the knee joint between individuals with ACL-RC, ACL-DF patients and healthy controls during these four common tasks using 3D motion analysis?

## 2.4 Alternative hypotheses

$H_1$: All lower extremity kinematics and kinetics as well as CV (as an outcome measure of variability) are reliable between days in all four tasks.

$H_2$: There are significant differences either in MV or knee biomechanics between risky and non-risky tasks in healthy individuals.

$H_3$: There is a relationship between MV and ACL risk factors.

$H_4$: There are significant differences either in MV or knee biomechanics between deficient knees and contralateral knees in ACL-DF individuals during the four tasks.

$H_5$: There are significant differences either in MV or knee biomechanics between reconstructed knees and contralateral knees in ACL-RC individuals during the four tasks.

$H_6$: There are significant differences either in MV or knee biomechanics observed between ACL-RC, ACL-DF and healthy individuals during the four functional tasks.

## 2.5 Thesis Structure

The following shows the structure of the thesis and how this was structured in order to answer the aims explained previously.

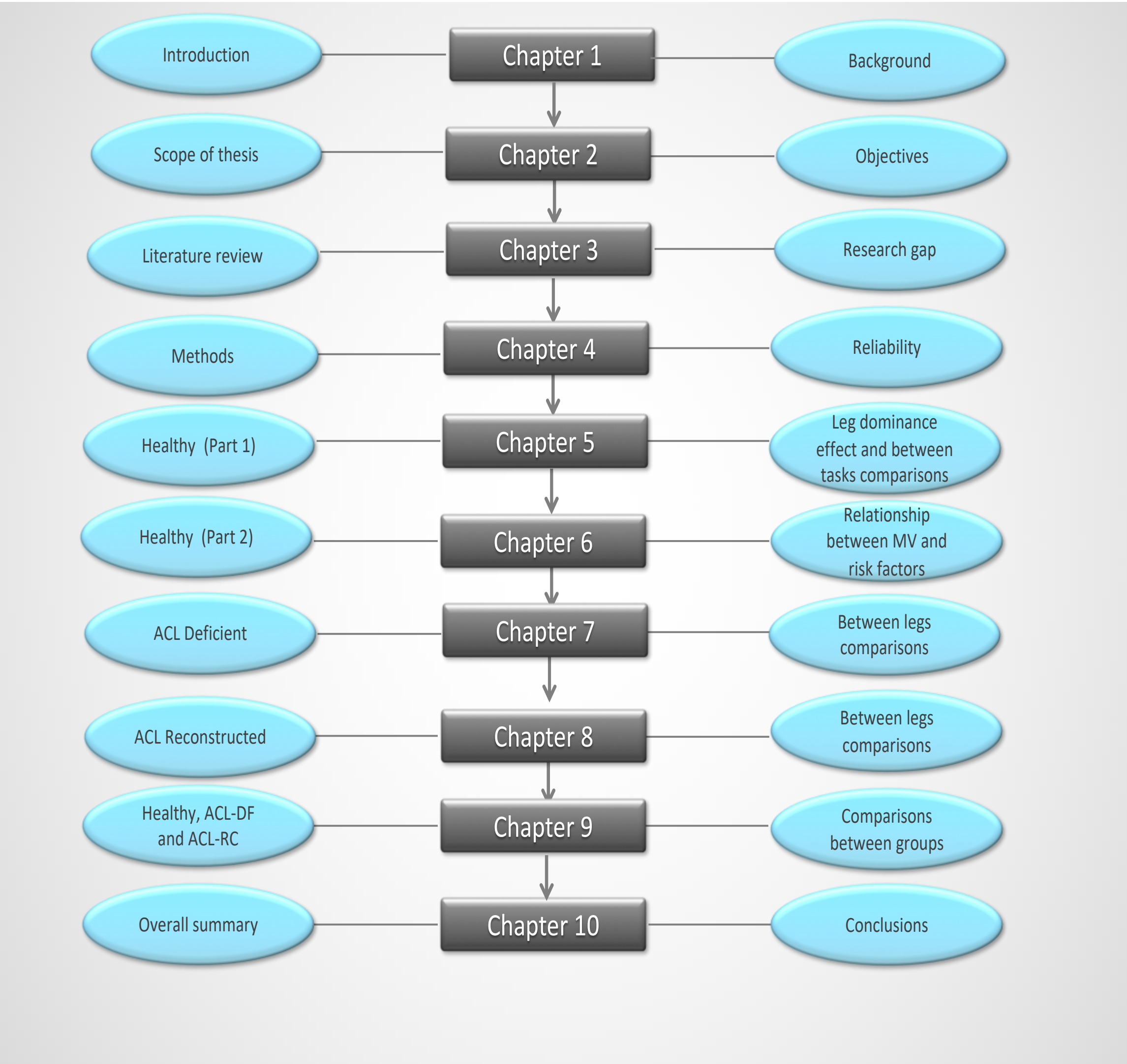



*Figure 2-1 The structure of the thesis*

# Chapter 3

## 3 Literature Review

This chapter aims to review relevant literature on the knee joint, ACL injury and its associated risk factors, particularly biomechanical risk factors, in order to: identify the condition; establish its importance based on the extent of the problem and the burden it generates; and explore the mechanisms of ACL injury, as well as its management. In addition, a review will subsequently be undertaken of movement variability (MV) and its importance in terms of knee biomechanics and injury. This chapter also contains a review of 3D motion analysis and sport screening tasks. Finally, research questions will be developed based on an identified research gap, while the aims of the study which seek to bridge this gap will also be set out.

### 3.1 Anatomy

The knee is described as a modified hinge joint capable of handling a wide variety of weight stresses, whilst still maintaining a high level of stability, thus permitting flexion and rotation. The knee comprises two joints, namely the femorotibial and patellofemoral (Goldblatt & Richmond, 2003). The stability that the knee joint can demonstrate, as highlighted by Simon (2000), is primarily due to the bony architecture of the femur, tibia and patella, as well as the static and dynamic restraints of the ligaments, capsule and musculature that encase the joint. The underlying bone structure can, to some degree, govern the overall movements permitted by the joint. Comprising two condyloid articulations, the femorotibial joint is the largest joint in the human body. Both the lateral and medial femoral condyles move with the tibial plateaus, to which they are paired. Medial and lateral menisci support the joint, which improves its conformity, whilst at the same time aiding in the rotation of the knee joint (Figure 3-1).

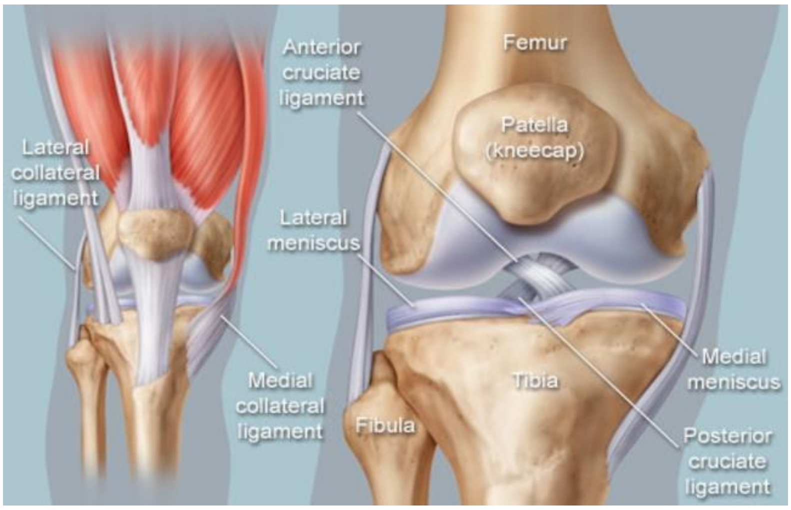
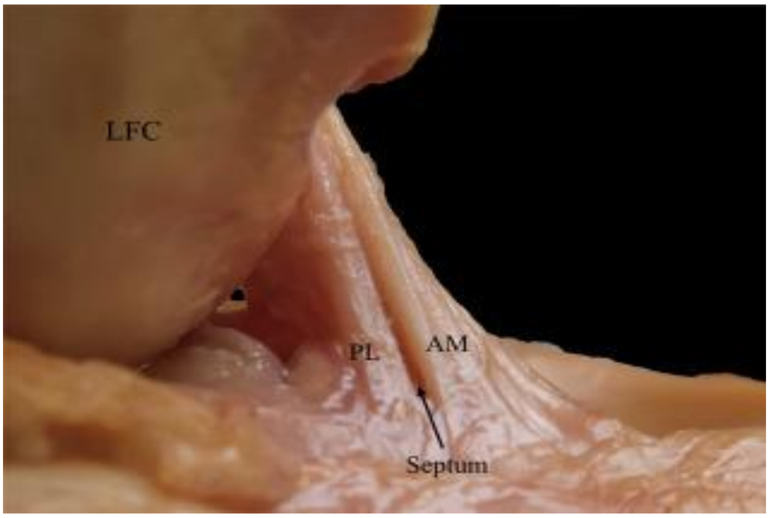
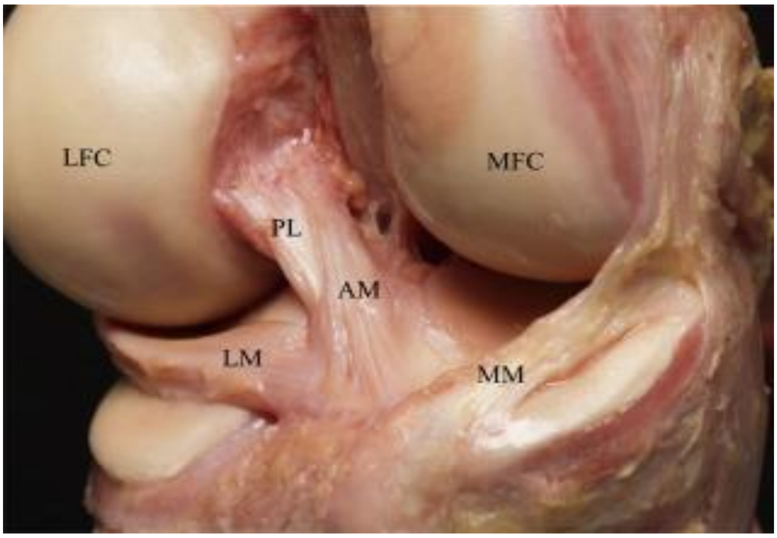



*Figure 3-1 Anatomy of the knee joint (Hoffman, 2019)*

Albers, Chambers, Sheean, and Fu (2019) have described the anatomy of the ACL as multi-fascicular in structure and complex in shape. It links the medial wall of the lateral femoral condyle to the midportion of the tibial plateau. The latter authors also found that the two major fibre bundles of the ACL, known as the anteromedial (AM) and the posterolateral (PL) bundles, are visible by the 16$^{th}$ week of foetal development during pregnancy. As illustrated in Figure 2-2 (A) below, the ACL's double bundle anatomy is depicted from an anterior angle, and the knee is in a flexion to a minimum of 90 degrees. While maintaining this posture (see Figure 3-2 (B) below), the PL bundle adopts a posterior and lateral position on the site of the tibial insertion, and an anterior angle on the femoral insertion site. Conversely, the AM bundle is situated in a more anterior and medial position on the tibial side, while it is located more on the posterior of the femoral side (Giuliani, Kilcoyne, & Rue, 2009; Irarrázaval, Albers, Chao, & Fu, 2017). Widespread variation occurs at population level in terms of ACL tibial insertion site length, with Albers et al. (2019) recording a mean span of 17 ± 2 mm. In addition, a mean AM bundle length of 9.1 ± 1.2 mm and width of 9.2 ± 1.1 mm was noted. These mean figures were found to be slightly lower for the PL bundle, measuring 7.4 ± 1.0 mm in length and 7.0 ± 1 mm in width.




(A)




(B)

Figure 3-2 The ACL anatomy

*(A) The anteromedial (AM) and posterolateral (PL) bundles of the ACL separated by the septum in a knee flexion of 90 degrees. (B) Macroscopic anatomy of ACL; anteromedial (AM) bundle; posterolateral (PL) bundle; lateral femoral condyle (LFC); lateral meniscus (LM); medial femoral condyle (MFC); and medial meniscus (MM)* (Albers et al., 2019).

### 3.2 ACL functional anatomy

The anterior cruciate ligament (ACL) performs a vital role in maintaining normal knee joint functioning. In addition, it ensures that the knee remains stable, especially when performing high-intensity exercises, which demand both cutting and pivoting movements. Burnham and

Wright (2017) identified that a young active cohort, most notably young female athletes, are at greater risk of sustaining ACL tears. Furthermore, the ACL is the main stabilising factor in anterior translation and the subsequent stabilising factor in the tibia's internal rotation (Andriacchi & Dyrby, 2005; Chaudhari, Briant, Bevill, Koo, & Andriacchi, 2008; Chen, Wang, Warren, & Maher, 2017; Georgoulis et al., 2010; Micheo, Hernández, & Seda, 2010). The ACL also prevents excessive movement of the knee, including extension, abduction and adduction (Liu-Ambrose, 2003). The ACL has two bone attachments (Femoral and tibial) as explained above. It has been shown that the ACL femoral attachment constitutes a central area separated by dense fibres, which have been directly inserted into the femur, as well as both the anterior and posterior fanlike extension areas (Mochizuki & Akita, 2016). The same study also reported that the central area had the capacity to resist 82–90% of the load when an anterior drawer force was applied, whereas the anterior fanlike area could resist 2–3%, while the posterior fanlike area could withstand 11–15%. In all four central areas, the majority of load occurred proximate to the roof of the intercondylar notch. In addition, the ACL's fanlike extension fibres attached to the surface of the bone, irrespective of the knee flexion angle, the fibres location and orientation do not change, in relation to the femoral surface. Conversely, alterations were noted during knee movements in the orientation of the ACL midsubstance fibres related to the femur.

The majority of earlier studies have reported that the tibial insertion site of the ACL is oval in shape. More recently, this theory has altered, with a number of authors. More recently, this theory has altered, with a number of authors (Lord, El-Daou, Zdanowicz, Śmigielski, & Amis, 2019; Siebold et al., 2015; Śmigielski et al., 2015) proposing that it is C-shaped, while the ACL resembles a ribbon. Lord et al. (2019) observed that the C-shaped dense collagen fibre area provided between 68–92% of the resistance to anterior tibial translation during the knee flexion, where a more substantial contribution is made at 90° as opposed to 0°. The latter authors also highlighted that the peripheral anteromedial fibres constitute a key region in the ACL tibial attachment site, as they play a crucial role in the restraint of tibial anterior translation and internal rotation when isolated and coupled displacement occur.

### 3.3 Injuries in sport

Approximately 80% of sports-related injuries are musculoskeletal, thus making them the most common injuries to occur as a result of engagement in sporting activities (John et al., 2016). In both professional and recreational sports, the latter authors have explained that the joints, most notably the knee, account for a large portion of the injuries sustained by sports players. It has been estimated that knee injuries arising from various sporting activities account for 15–50% of all injuries (Thacker, Stroup, Branche, & Gilchrist, 2003). These kinds of injuries can be the subject of lengthy rehabilitation periods following surgery and some, according to a publication by Swenson et al. (2013), can result in the individual being chronically disabled in either their field of sport or even their work environment. In their investigations of the healthcare costs associated with the treatment of knee injuries, studies by Gage, McIlvain, Collins, Fields and Comstock (2012) and Eggar (1990) both found that they were the most expensive and serious types of injuries to treat, as compared to other presentations.

In the case of knee injuries, one of the most frequent injuries seen are ACL injuries. It has been estimated that the median incidence of ACL injury to be generally in the region of 0.03% per person annually, while this figure could escalate to 3.7% among a professional athlete cohort. In addition, According to a 2001 study undertaken by Murrell, Maddali, Horovitz, Oakley, & Warren , this injury is linked to early osteoarthritis (OA) development. Weiss & Whatman (2015) added that ACL injury can limit patient capacity to participate in everyday activities, including sport. A previous study conducted in 2000 by Starkey reported that ACL can result in a sports player never fully recovering again to achieve their original performance levels. A cohort study of 503 patients carried out by Ardern, Webster, Taylor, & Feller (2011) found that following anterior cruciate ligament reconstruction (ACL-RC), in reality, only 33% of patients fully recovered to regain their original performance levels after a 12-month period. Therefore, knee-related sport injuries, as well as ACL injuries, in particular, have become the main focus of interest by a significant number of researchers, who wish to identify ways in which to minimise this problem.

### 3.4 Incidence and economic burden of ACL injuries

The number of ACL injuries that typically occur in the United States of America (USA) has been estimated to be between 60,000 and 200,000 per annum (Gornitzky et al., 2016; Herzog et al., 2018). Females have been shown to be of greater risk of ACL injury, where a four- to six-fold increased risk has been reported (Arendt, Agel, & Dick, 1999; Márquez, Alegre, Jaén, Martin-Casado, & Aguado, 2017). The vast majority of ACL injuries (70 percent) occur during non-contact episodes (Quatman & Hewett, 2009). The monetary implications associated with this type of injury are substantial, due to the cost of imaging, reconstructive surgery, postoperative bracing and rehabilitation, creating average yearly spending of over 2 billion US dollars for all ACL-related injuries (Gottlob, Baker, Pellissier, & Colvin, 1999; Malek, DeLuca, Kunkle, & Knable, 1996). Furthermore, the psychological effect of this injury is significant on each individual person, and this should also be taken into account, especially in the case of high-level athletes. Most often, an athlete suffering from an ACL injury will be unable to participate in a competition for a 6–9-month period. This can have very significant repercussions for them, especially those athletes dependent on competitive play for monetary gain.

Furthermore, as outlined by Lohmander et al. (2004), there can often be long-term impacts on the knee following an ACL injury, comprising a change in knee kinematics, harm to the meniscus and cartilage, and, in extreme cases, OA. In addition, athletes who have undergone ACL-RC have been found to be at significantly greater risk of developing secondary injuries and knee OA in the long term ( Lohmander, Englund, Dahl, & Roos, 2007). A recent systematic review and meta-analysis conducted by (Grassi et al., 2015) reported that, despite almost eight out of ten patients resuming sporting activities following ACL revision surgery and demonstrating good stability, only half of all patients returned at the same pre-injury skill level.

### 3.5 Repair of ACL function

Many people with an ACL rupture or tear undergo a surgical procedure, which is known as ACL reconstruction (ACL-RC), in order to stabilise the injured knee (Joseph et al., 2013; Mather III et al., 2013). The incidence of ACL-RC differs from country to country. Table 3-1 presents these figures for eight different countries worldwide. Notably, an increase in the incidence rates has occurred over the years, thus signalling a rise in the number of new cases presenting with this type of injury. For example, in England, the ACL-RC rate increased twelve-fold between 1997 and 2017 (Abram, Price, Judge, & Beard, 2019). In fact, the period of most rapid growth occurred between 2005/2006 and 2008/2009. While, the underlying reasons for the increase are unclear, they could be attributed to a rise in the level of competition in this country during these seasons. Despite the increase in ACL-RC rates among the population in England, the incidence rate (24.2/100000) is low, as compared to other countries, such as the USA (74.6/100000). This increase was also observed in US-based patients aged 65 years or less, which represents a rise from 61/100000 in 2002 to 74.6 in 2014 (Abram et al., 2019). In fact, the reasons why this increase has occurred still remain unclear, and could probably be attributable to unexplored factors. Analysis of these numbers and rates clearly highlight the extent of this problem. Therefore, in order to understand the causes of this injury, it is very important to identify potential contributory risk factors.

*Table:3-1 Incidence rate of ACL-RC in eight different countries worldwide*

| Country | ACL reconstruction incidence per 100,000 person-years (PYs) | Time period | Reference |
|---|---|---|---|
| **Australia** | 52 | 2000–2015 | (Zbrojkiewicz et al., 2018) |
| **Denmark** | 38 | 2005–2007 | (Lind et al., 2009) |
| **England** | 24.2 | 1999–2017 | (Abram et al., 2019) |
| **New Zealand** | 37 | 2000–2005 | (Gianotti et al., 2019) |
| **Norway** | 34 | 2004–2007 | Granan et al., 2009) |
| **Scotland** | 8 | 1996–2000 | (Clayton et al., 2008) |
| **Sweden** | 32 | 2004–2007 | (Granan et al., 2009) |
| **USA** | 74.6 | 2002–2014 | (Herzog et al., 2018) |

## 3.6 Functional consequences of a loss of ACL function

Accordingly, this means that following any ACL tear, a reduction in external rotation will occur (Andriacchi & Dyrby, 2005), as well as an increase in internal rotation of the tibia within the area closest to the femur (Chaudhari et al., 2008; DeFrate et al., 2006; Gao & Zheng, 2010; Georgoulis et al., 2010; Keene, Bickerstaff, Rae, & Paterson, 1993). This can also be readily observed in any magnetic resonance imaging (MRI) scans (Barrance, Williams, Snyder-Mackler, & Buchanan, 2006). According to Georgoulis et al. (2010), this is thought to be attributable to the absence of an important 'screw home' motion, which should occur at the end of the swing phase. The latter authors found that it results in the femur and tibia being misaligned on reconnection. Medical research has shown that any severe malalignment with joint contact can result in the onset and later development of knee OA (Andriacchi, Briant, Bevill, & Koo, 2006; Andriacchi & Mündermann, 2006).

This has been further supported by additional evidence provided by Lohmander, Östenberg, Englund and Roos (2004), Micheo et al. (2010), and Von Porat, Roos and Roos (2004), who suggest that of those patients who present with such an injury, with or without ACL-RC, 50% of them will subsequently go on to develop symptomatic OA within 12–14 years of the occurrence of that injury. Goldring, Otero, Tsuchimochi, Ijiri and Li (2008) contend that, undoubtedly, there is a biological dimension to any visible presentation or exacerbation of OA. However, studies by Kanamori et al. (2002) and Solomonow et al. (1987) have highlighted that mechanical issues are also likely to exert an influential effect. In addition, other researchers have demonstrated that cartilage degeneration may be accelerated in ACL-RC or ACL-DF individuals with a high body mass index (BMI), as compared to those with a normal BMI (Culvenor et al., 2015). Men have also been shown to be of higher risk of developing early OA following ACL-RC (Culvenor et al., 2015).

However, there is still a need to undertake research in order to establish whether the early onset of OA occurs due to the increased mechanical stresses being placed upon the knee as a result of the initial injury. Alternatively, it raises the question as to whether, in fact, it is because of a different loading balance being triggered as a result of the repetitive movement of the limb subsequent to injury. It may perhaps be a combination of both factors. In addition to the important mechanical function the ACL performs in terms of maintaining the stability of knee, it also contains mechanoreceptors (2.5%) which play a crucial role in the neuromuscular control of the knee joint (Zimny, Schutte, & Dabezies, 1986).

## 3.7 Return to sport following ACL injury

The majority of athletes who have injured their ACL undergo ACL-RC in order to return to their preinjury levels of sport (Webster & Hewett, 2019). However, only approximately 50% achieve this within one year, while approximately 60% will require two years (Webster & Hewett, 2019). Another cohort study of 503 patients carried out by Ardern, Webster, Taylor and Feller (2011) found that following ACL-RC, in reality, only 33% of patients fully recovered to regain their original performance levels after a 12-month period. However, other research has reported that following ACL-RC, 35% of athletes did not return to their preinjury sport levels within two years (Van Melick et al., 2016). In addition, ACL can result in a sports player never fully recovering again to achieve their original performance levels (Starkey, 2000). Therefore, knee-related sport injuries, as well as ACL injuries, in particular, have become the main focus of interest by a significant number of researchers, who wish to identify ways in which to minimise this problem.

## 3.8 Risk factors of ACL injury

Multifactorial risks associated with ACL injury have been identified in the literature. In order to acquire a better understanding of these risk factors, they can be categorised into two main types, namely extrinsic and intrinsic. Intrinsic factors can be defined as factors which are inherent to the individual and can be subdivided into two types: modifiable or non-modifiable. Conversely, extrinsic risk factors are those that remain outside of the control of the individual.

### 3.8.1 Extrinsic risk factors

A systematic review published in the last year identified 10 extrinsic risk factors (Pfeifer, Beattie, Sacko, & Hand, 2018). Half of these were influenced by weather conditions, two by the surface of the play area, while the remaining three risk factors were ski type, sport level and degree of engagement in sport. In terms of the impact of the weather, it was found that high evaporation rates over a four-week period prior to a match was associated with greater risk of injury. Other risk-generating conditions include low rainfall amounts in the twelve months prior to a match, in combination with no precipitation occurring throughout a match. Finally, ice and snow accumulations or skiing as snow is falling poses increased risk for female recreational skiers.

#### 3.8.1.1 Environmental conditions

Weather conditions surrounding matches over several seasons of match play was investigated by few researchers (Orchard, Seward, McGivern, & Hood, 1999; J. W. Orchard, Chivers, Aldous, Bennell, & Seward, 2005). The Longitudinal study observing match weather conditions by Orchard et al., (1999) revealed that high evaporation ratings recorded over a 28-day period prior to a match date resulted in players being at greater risk of sustaining ACL injury (relative risk (RR) = 2.8). In addition, findings from these two studies concurred that such risk was elevated if low precipitation levels occurred in the year preceding the games (RR = 1.5 to 1.93). According to Orchard et al. (2005), players were also at greater risk of injury if no rain whatsoever fell throughout the matches (RR = 1.55). Conversely, the combination of low evaporation ratings and high rainfall amounts substantially decreased the risk of ACL injury. The latter authors accounted for this pattern by positing that a reduction in shoe-surface traction occurs as a result of the rain softening the soil, thus requiring less force to be transmitted across the knee when carrying out particular movements, most notably, pivoting.

In addition, Orchard et al. (2005) claimed that covering pitches when water evaporation levels are high, along with watering them on an ongoing basis may reduce the risk of ACL injury. A further weather conditions study by Ruedl et al. (2011) noted that female recreational skiers were of greater risk of sustaining an ACL injury during icy snow conditions (odds ratio (OR) = 24.33) or while skiing during snowfalls (OR = 16.63). Heightened risk may be due to the impact of a number of factors on knee movement, for example, the use of artificial snow and snow grooming processes or icy patches forming in ski resorts with glaciers.

#### 3.8.1.2 Playing surface

The surface of a play area can also contribute to ACL injury. For example, the installation of rubber matting on an army obstacle course increases the risk of sustaining ACL injury, as evidenced by a study conducted by Pope (2002) in the Australian army ($\chi^2 = 4.76$). The use of Bermuda grass has also been shown to heighten the risk of injury among Australian football players (RR = 1.87) (Orchard et al. (2005). In contrast, the latter reported that rye grass was associated with fewer injuries. Therefore, they found grass species to be a key determinant in accounting for ACL injury, particularly among Australian Football League (AFL) players in northern states. Orchard et al. (2005) identified the most likely contributory cause of this pattern as being due to ongoing reduced shoe-surface traction as a result of the grass having a shallower thatch layer. Despite these findings, one single element, namely the type of grass species, would not appear to be sufficient in itself to entirely explain the high rate of ACL injury during the early stages of the football season. Nevertheless, the season stage remains a risk factor worthy of note, having also taken grass type into consideration.

#### 3.8.1.3 Other extrinsic risk factors

Pfeifer et al. (2018) identified a number of other extrinsic risk factors, most notably high levels of engagement in sporting activities on a weekly basis, as well as a greater propensity for injury among college as opposed to high school student cohorts participating in sport and among recreational female skiers who use traditional rather than carving skis (Ruedl et al., 2011). In fact, some of these factors are non-modifiable, such as a playing surface and the weather, as they are outside the control of the individual. In contrast other factors can be modified by the individual, such as participation levels. Furthermore, the type of ski a recreational skier selects is dependent upon individual choice.

### 3.8.2 Intrinsic risk factors

Intrinsic risk factors can be classified into five broad types, namely anatomic, neuromuscular, physiologic, biomechanical, and genetic.

#### 3.8.2.1 Anatomical risk factors

Anatomical risk factors can account for ACL injury, in particular in those presenting with intercondylar notch stenosis or with a narrow intercondylar notch, as measured using the notch width index (Pfeifer et al., 2018). In addition, posterior or lateral tibial slope increases are more common among individuals with ACL-DF knees than those who have not acquired any injury. Those with an open lower extremity physes have been classified as belonging to a paediatric cohort. Greater likelihood of risk is associated with individuals with an open physes, in combination with an increased medial tibial slope, as demonstrated by Vyas, Van Eck, Vyas, Fu and Otsuka (2011). Athletes who are more prone to ACL injury may display the following characteristics: a deep medial tibial plateau and lateral tibial plateau; greater distance between their tibial tuberosity and trochlear groove; alpha angle increases, as determined using sagittal view films of the intercondylar notch roof by the long axis of the femur; and general joint laxity. Characteristics directly associated with the ligament, most notably, a reduction in width, volume or size, combined with an increase in its length, are all deemed to create more vulnerability to injury.

### 3.8.2.2 Neuromuscular risk factors

According to Zazulak, Hewett, Reeves, Goldberg, and Cholewicki (2007), those with hip abductors and external rotators, which are insufficiently strong to support their body weight, are more prone to injury. Khayambashi, Ghoddosi, Straub, and Powers (2016) also identified that decreased hamstring strength can create greater susceptibility to injury. Furthermore, ACL injury is more likely to manifest when an imbalance occurs between the hamstrings and the quadriceps muscle groups (Alentorn-Geli et al., 2015). Zebis, Andersen, Bencke, Kjær, and Aagaard (2009) highlighted that elite female handball and soccer athletes may experience elevated risk in the event that the electromyography muscular pre-activity levels of their lateral hamstrings are reduced in comparison to their lateral quadriceps. An earlier study by Kramer, Denegar, Buckley, and Hertel (2007) also demonstrated that a reduction in iliotibial band flexibility levels is associated with ACL injury and changes in lower extremity movement may account for this elevated risk.

### 3.8.2.3 Physiologic risk factors

Hägglund and Waldén (2016) highlighted that an above average weight or BMI is associated with increased risk of injury. A later study by Pfeifer et al. (2018) reported that during the pre-ovulatory menstrual cycle, women are more likely to acquire an injury, and this was also found to be true of those who are post menarche. Furthermore, Beynnon et al. (2014) identified gender-related risk factor differences, in that females are more likely to develop an initial, non-contact ACL injury as opposed to males. In addition, age differences are evident, where Hägglund and Waldén (2016) noted that young people below the age of 14 years are more likely to experience an ACL injury.

#### 3.8.2.4 Leg dominance

The terms leg dominance, lateral preference or laterality all describe the tendency to be more skilled in using a limb on one side of the body when carrying out a motor activity, as opposed to the other, thus leading to this side becoming dominant (Maloney, 2019; Carpes, Mota, & Faria, 2010). For example, identifying a preferred leg to kick a football signals both limb and skill dominance. Accordingly, both researchers and clinicians are keen to establish whether one specific limb is of increased risk of sustaining an injury. Pioneering research undertaken by Hewett et al. (2005) reported significantly greater dominant as opposed to nondominant asymmetries in landing peak knee abduction moment in nine young female athletes who injured their dominant limb, as compared to those who remained uninjured. Interestingly, of the nine athletes who suffered from dominant limb injury, six would have classified it as their preferred kicking leg. Furthermore, a prospective study conducted by Paterno et al. (2010) noted that athletes who sustained a second ACL injury were more likely to display greater between-limb asymmetries in knee extensor moments during landing.

In addition, multiple studies have identified leg dominance as a potential risk factor for ACL injury (Ruedl et al., 2012; Brophy, Silvers, Gonzales, & Mandelbaum, 2010; Matava, Freehill, Grutzner, & Shannon, 2002; Negrete, Schick, & Cooper, 2007). More recently, in a sample comprising 302 participants with noncontact ACL tears, Negrete et al. (2012) sought to explore the relationship between gender, leg dominance and side of injury occurrence. Findings revealed that while no significant gender effect occurred in the relationship between leg dominance and side of injury, women were more likely to tear their left ACL as opposed to their right ($P$ = .06) (Negrete et al., 2012). According to the data provided in the study, a majority of the female participants (91%) reported that their dominant leg was the right leg, and 77% of the women had an ACL injury in their nondominant leg. In a mixed gender study of soccer players, Brophy et al. (2010) highlighted that approximately 68% of the females with a noncontact ACL rupture sustained an injury to their non-dominant leg, with a lower proportion of males (26%) experiencing a similar injury. They discovered that there was a higher statistical probability that the males would injure their preferred kicking leg, whereas the females tended to sustain an injury to their preferred support leg. In seeking to establish whether leg dominance was a risk factor for noncontact ACL injuries in a cohort of 65 male and 128 female recreational skiers, Ruedl et al. (2012) examined the relationship between gender, leg dominance and side of knee injury occurrence. This study revealed that the likelihood of females sustaining an ACL rupture was twofold greater in their nondominant leg.

#### 3.8.2.5 Genetic risk factors

Several studies have been carried out to explore genetic-related susceptibility to ACL injury. For example, variation in human collagen genes, such as COL1A1 or COL12A1, was found to increase ACL risk among both genders (O'Connell et al., 2015). Furthermore, a correlation may exist between matrix metalloproteinase genes, which perform a key function in tissue remodelling, and injury (Posthumus et al., 2012).

#### 3.8.2.6 Biomechanical risk factors

In addition to the aforementioned risk factors presented above, there are also known biomechanical risk factors, and a few of these could be modifiable. Numerous biomechanical risk factors associated with the lower leg are regarded as possible threats to injury. There is a substantial body of literature on biomechanical studies, where the risk factors relating to these injuries have been investigated. The movements that have been examined biomechanically occur in three planes, namely frontal, sagittal and transverse.

#### 3.8.2.6.1 Frontal plane

Movements in the frontal plane, namely knee abduction (knee valgus), hip adduction and foot eversion have been linked to ACL injury. Beginning with knee valgus movement and moment, it has been found that an increase in this area, in particular, is considered to be the main risk factor for ACL injury (Numata et al., 2018; Myer et al., 2015; Hewett et al., 2005; Hewett, Torg, & Boden, 2009; Cochrane et al., 2007; Ebstrup & Bojsen-Møller, 2000; Kobayashi et al., 2010; Olsen et al., 2004). Studies have indicated that the knee's valgus load is elevated by minor alterations in valgus movement, for example, the ACL's pressure could be six times higher with just five degrees of valgus (Bendjaballah, Shirazi-Adl, & Zukor, 1997). Furthermore, a mere two degrees of change could result in a 100% rise in knee valgus moment, with valgus load rises causing a greater chance of injury (McLean, Lipfert, & Van Den Bogert, 2004), while an increase in valgus motion will probably lead to less control of the joint at the frontal plane, thus heightening the injury risk (Ford et al., 2003). To be precise, an almost fully extended knee valgus, combined with risks such as anterior tibia shear force, tibial rotation and foot pronation, may increase the risk of non-contact ACL damage (Chappell, Yu, Kirkendall, & Garrett, 2002; Ireland, 1999). Additionally, ACL injuries can be predicted by the presence of knee valgus motion and moment (Numata et al., 2018; Myer et al., 2015; Hewett et al., 2005).

Several points should be taken into consideration in relation to the aforementioned studies which identify knee valgus motion/moment as risk factors for ACL injury. Firstly, in the case of such investigations, a prospective study is the best approach to adopt, and three studies were found to have met this criterion: Hewett et al. (2005); Myer et al. (2015); and Numata et al. (2018). However, all the remaining studies used video analysis (Hewett et al., 2009; Cochrane et al., 2007; Ebstrup & Bojsen-Møller, 2000; Kobayashi et al., 2010; Olsen et al., 2004).

Another important point is that with two exceptions, all of the remaining studies comprised females only, and males were excluded, despite the fact that this injury is common among both genders. Kobayashi et al. (2010) and Hewett et al. (2009) are the only two studies that included males and females in their investigations.

Furthermore, the two studies which investigated knee valgus moment (Hewett et al., 2005; Myer et al., 2015) failed to normalise for valgus moment, as it is sensitive to other factors, such as body weight and height. While different normalisation methods are used in biomechanics studies, it is believed to play a very important role in reducing the residual effects on GRF and moments (Wannop, Worobets, & Stefanyshyn, 2012). In contrast, Numata et al. (2018) did not investigate moments as they used an alternative method (2D motion analysis as opposed to 3D motion analysis). Another very important consideration is the sample size selected, as this type of investigation requires large sample sizes in order to avoid type 1 errors occurring. Unfortunately, none of the aformentioned studies performed power calculations to determine their sample sizes. Furthermore, all prospective studies undertook their investigations during one sporting task only, which was landing, while this injury also commonly occurs during cutting tasks.

In terms of the hip joint, it has been suggested that hip adduction, which is the second movement in the frontal plane, is a risk factor for ACL injury, and, in fact, it has been shown to correlate with the knee valgus angle (Hollman et al., 2014; Willson & Davis, 2008). In particular, increased hip adduction was found to significantly correlate with increased knee valgus (Hollman et al., 2014; Willson & Davis, 2008). However, these studies were not prospective in nature and they only included one gender (females). In addition, both studies did not perform power calculations to determine the sample size, while, notably, the research undertaken by Willson and Davis (2008) comprised a small sample size (n=20) which may result in a type 1 error.

In addition, hip adduction moment has been reported to also strongly correlate with a knee valgus moment (Hewett et al., 2005). However, this relationship was found to occur in people who had undergone ACL-RC. Despite differences of opinion in relation to the exact function hip movement serves in the frontal plane (abduction or adduction) in ACL injury, when performing different tasks, nevertheless, it is believed that hip movement in the frontal plane plays a very important role in the stability of the knee joint and could potentially be a risk factor. There is a need to conduct high-quality prospective studies in order to determine the exact function the hip frontal plane performs in this type of injury, as well as to clearly identify its associated risk factors. This will assist clinicians and researchers to target this area when developing ACL injury/reinjury risk mitigation programmes.

The final movement in the frontal plane is foot eversion, which is linked with tibial internal rotation during movement (Nawoczenski, Saltzman, & Cook, 1998). There is a lack of clear evidence available regarding the role foot movement plays in terms of the risk of developing an ACL injury. However, a recent study by Teng, Kong and Leong (2017) tested the effects of three-foot rotation positions on knee valgus on eleven recreational basketball players during a single-leg drop landing task. They found that foot rotation exerted a significant effect on the knee valgus angle ($p < 0.001$, $\eta^2 = 0.66$). They also concluded that athletes should land with less foot pronation.

Undoubtedly, the need exists to conduct high-quality prospective studies to clearly determine whether foot frontal plane biomechanics will identify the risk factors associated with ACL injury.

#### 3.8.2.6.2 Sagittal plane

In terms of ACL injury, the most commonly investigated movements in the sagittal plane are knee movement, followed by hip movement. It is believed that ACL maximum strain occurs at the angle closest to full extension of the knee joint (Leppänen, Pasanen, Kujala et al., 2017; Markolf et al., 1995; Berns, Hull, & Patterson, 1992; Boden, Dean, Feagin, & Garrett, 2000). Alternatively, decreased knee flexion has been shown to be a risk factor associated with ACL injury during landing, as reported in a recent prospective study undertaken by Leppänen et al. (2017) and in video analysis research conducted by Boden et al. (2000). In contrast, another study by Krosshaug et al. (2007) demonstrated that increased knee flexion has been detected in many cases of non-contact ACL injury. However, as the latter relied upon video analysis-based research, this type of approach cannot confirm risk factors as could have been achieved using a prospective study. Biomechanically, increased flexion in the lower extremity joint will lead to higher absorption of energy in muscles and less energy transmission to passive elements of the knee (Alentorn-Geli et al., 2009). Therefore, decreased knee flexion is more widely recognised as a risk factor for ACL injury, as it will lead to lower absorption of energy in the muscles and more energy transmission to passive elements of the knee (Alentorn-Geli et al., 2009), which may cause injury.

In addition, increased external knee flexion moment has been identified as a risk factor for ACL injury during landing in female athletes (Leppänen et al., 2017). The authors of this study suggested that individuals who had ACL injuries had increased quadriceps forces. Such findings support the notion that the quadriceps are capable of producing significant ACL loading, particularly at angles closest to the full knee extension (Beynnon et al., 1995; DeMorat, Weinhold, Blackburn, Chudik, & Garrett, 2004; Yu & Garrett, 2007). This prospective study solely investigated the sagittal plane (both movements and moments). However, it comprised only one gender (females) and a single task (vertical drop jump).

With regard to other sagittal movements, the hip sagittal movement has also been shown to be a factor that can increase the risk of ACL injury (Krosshaug et al., 2007; Leppänen et al., 2017). However, uncertainty exists as to whether increased or decreased hip flexion is associated with the risk of developing ACL injury. In a recent prospective study conducted by Leppänen et al. (2017), decreased hip flexion was demonstrated to be a risk factor for ACL injury during landing in female athletes. However, Krosshaug et al., 2007 reported that increased hip flexion was a risk factor for the same injury. As previously mentioned, the study by Krosshaug et al. (2007) used video analysis, while the research undertaken by Leppänen et al. (2017) identified the risk factors prospectively. Hence, in this particular case, the latter study is more accurate. The conflicting results emanating from the two studies could be attributable to the significant differences in the study designs they selected. From a biomechanical perspective, as mentioned above, increased flexion will lead to higher absorption of energy and less energy transmission; therefore, a decrease in hip flexion would increase the risk of ACL injury.

#### 3.8.2.6.3 Transverse plane

In the transverse plane, two movements have been linked to the risk of ACL injury, namely tibial rotation and hip rotation.

**Tibial Rotation**

Firstly, tibial rotation plays a crucial role in ACL injury (Weiss & Whatman, 2015). However, there is uncertainty as to whether internal rotation or external rotation of the tibia creates greater risk of injury (Weiss & Whatman, 2015). Tibia internal rotation has been observed during the mechanism of ACL injuries in videotape analysis in research undertaken by Ebstrup & Bojsen-Møller (2000), Koga et al. (2010), and Olsen, Myklebust, Engebretsen, and Bahr (2004). All these studies have suggested that internal rotation of the tibia puts the ACL at greater risk of injury. This was clearly explained by Koga et al. (2010), who adopted a model-based image-matching method. They reported that the tibia rotated internally during the first 40 milliseconds after initial contact, and external rotation was subsequently observed, possibly after the ACL had torn.

Conversely, external rotation has also been linked to ACL injury in a number of different studies (Olsen et al., 2004; Ebstrup & Bojsen-Møller, 2000; Ireland, 1999). All researchers used videotape analysis. Surprisingly, a number of the aforementioned studies found that some cases demonstrated tibial internal rotation during the injury. However, in contrast, other cases demonstrated tibial external rotation during the same injury, in the same study (Ebstrup & Bojsen-Møller, 2000; Olsen et al., 2004). This could mean that both rotations could be associated with the risk of ACL injury. Researchers, who are of the belief that tibial external rotation is more likely to be associated with the risk of ACL injury, have advanced a number of reasons to explain this finding. ACL impingement is one such reason, which can come about as a result of external rotation of the tibia (Fung, Hendrix, Koh, & Zhang, 2007). Despite the arguments made regarding rotation, high-quality prospective studies would need to be conducted in order to determine the exact role tibial rotation plays in this type of injury, as well as to clearly identify the risk factors.

**Hip rotation**

It has been suggested that increased hip internal rotation leads to an increased Q angle (Powers, 2003), as observed during the mechanism of the ACL injury (Ireland, 1999). In addition, it has been shown that increased hip internal rotation will also lead to increased tibial external rotation, which could result in ACL impingement and may heighten the risk of ACL injury (Fung et al., 2007). Other studies have found that internal hip rotation is correlated with the knee valgus angle (McLean et al., 2004; Sigward, Ota, & Powers, 2008).

In fact, the need exists to conduct high-quality prospective studies to clearly determine the risk factors for ACL injury in the transvers plane across knee and hip.

#### 3.8.2.7 Ground reaction force

Ground reaction force (GRF) is a very important element with regard to knee biomechanics, as it exerts a significant influence on all knee moments. Padua and DiStefano (2009) have stated that GRF may affect the anterior tibial shear by influencing knee flexion-extension moments in an indirect manner. Furthermore, Yu, Lin and Garrett (2006) demonstrated that increased GRF may lead to greater quadriceps muscle force, as well as placing an increased load on the ACL. Increased vertical GRF has been identified as a risk factor for ACL injury during a landing task, as demonstrated in a recent prospective study by Leppänen et al. (2017). Prior to that research being undertaken, only one prospective study had been carried out by Hewett et al. (2005), who reported that increased vertical GRF is associated with risk of ACL injury during landing. However, both studies (Hewett et al., 2005; Leppänen et al., 2017) failed to normalise for moments and GRF, as these variables are sensitive to other factors, such as body weight and height. It is believed that normalisation is very important in reducing the residual effects on GRF and moments (Wannop et al., 2012). In addition, Hewett et al. (2005) only recorded nine ACL injuries out of a total of 205 participants, which is may considered to be a small proportion.

### 3.8.3 Other risk factors

Some risk factors could not be classified within the previously cited types, and were found to be female-specific. Research undertaken by Nilstad et al. (2014) revealed that those who had already acquired an ACL injury are of greater risk of a recurring injury in the same knee. Similarly, those whose immediate family members have a history of ACL injury are more likely to subsequently sustain an injury themselves (Flynn et al., 2005; Hägglund & Waldén, 2016). Gender-specific differences also manifested, whereby higher prevalence rates of ACL injury occurred in the non-dominant leg of young female soccer players (Hägglund & Waldén, 2016). In addition, Ruedl et al. (2012) reported that female recreational skiers are of heightened risk of injuring their non-dominant rather than their dominant leg.

## 3.9 Mechanism of ACL injury

Around three quarters of ACL tears occur with limited contact at the point in injury, or with none at all (Boden et al., 2000). Comprehension of ACL ruptures' injury mechanism has been enhanced by studies which utilised various modalities such as epidemiology, cadaveric research, video analyses and computer simulations. Nevertheless, as Wetters, Weber, Wuerz, Schub, and Mandelbaum (2016) explained, even with all the data and improved understanding, we still do not have comprehensive knowledge of the mechanism. Whilst the precise nature of a noncontact ACL injury can differ according to the sport being played, they generally occur through a sudden alteration in velocity, or, when weight-bearing, the generation of a multi-directional force upon the knee. In particular, rapid deceleration is associated with ACL injuries, such as the planting of the leg to stop and quickly move direction – as seen in football – or landing, pivoting and twisting motions (Boden et al., 2000).

As Shimokochi and Shultz (2008) explain, a major characteristic of an ACL noncontact injury is the presence of multiplanar loading, or numerous vectors of force onto the knee. In particular, harmful force can be placed on the ACL when coronal and sagittal loading merges with imbalanced hamstring and/or quadriceps muscle contractions (Ahmad et al., 2006). Empirical research of the exact point of an ACL injury indicates that a twisting motion of the knee combined with a valgus are usually present, and, as outlined by Markolf et al., (1995) and Markolf, O'neill, Jackson, and McAllister (2004), it is also observed that the force on the ACL can be increased by the presence of such twisting motions on the coronal pressure in an overly-extended knee. Numerous research has indicated that force on the coronal, sagittal and transverse planes can exacerbate stress on the ACL, thus causing damage (Wetters et al., 2016).

It is difficult to fully describe the intricate underlying movements of the mechanism behind an ACL injury. The processes are complex and may relate to the multiplanar knee movement often associated with ACL injuries. According to Quatman et al. (2010), his systematic review of the articles surrounding this injury indicate that 82% of direct ACL injury mechanism studies showed a multiplanar mechanism. Through extensive research in this area, the injury can be replicated with the following movements; flexion of the knee combined with axial load being produced whilst simultaneously initiating tibial rotation and valgus force (Figure 3-3). They relate their findings to the pivot shift clinical examination technique.

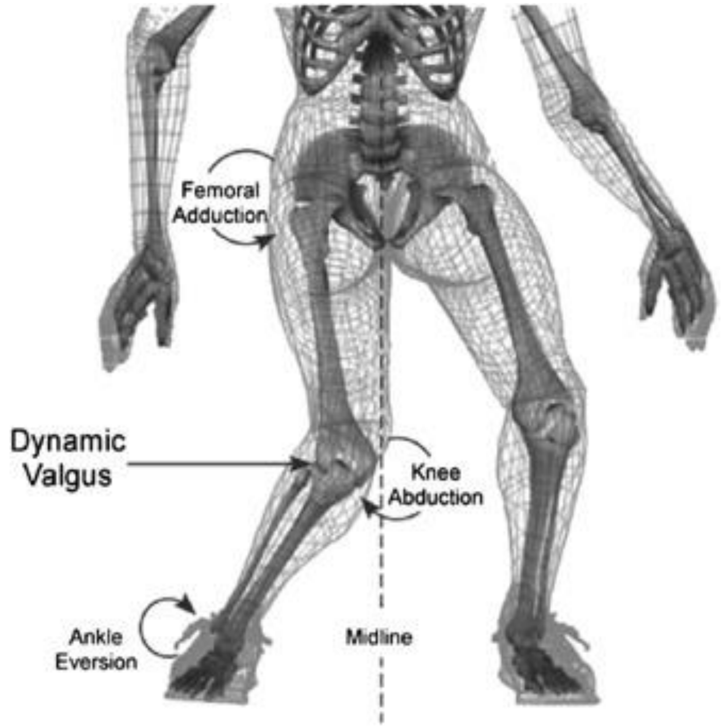



Figure 3-3 Non-contact ACL injury mechanism (Hewett, Myer, & Ford, 2006)

In an attempt to replicate precisely what occurs in this injury, consistent findings have been shown by studies of limb position at the time of non-contact injury. According to Boden et al. (2000) the sagittal position of the concerned knee often shifts between early flexion to hyper extension at the precise moment of injury occurrence. Sakane et al. (1999) suggested that the resulting injury is likely to be from the ACL seeing overstretched tensile composition in its quest to resist the anterior forces and shear within this range of motion. Shimokochi and Shultz (2008) also complimented this theory with their tests on the quadriceps muscle contraction at the time of the early flexion, which they said also resulted in greater ACL tension than is the case with deeper flexion. These findings indicate an elevated level of vulnerability and resulting injury as a result of numerous contractions within this movement range.

The previous studies go some way to informing clinicians of the processes involved with ACL injuries and can explain the effects of rapid deceleration. During these times, the quadriceps movements required to halt the sportsperson become elevated causing the ACL to be put through undue stress and strain. According to video footage by Boden, Torg, Knowles, and Hewett, (2009), the most influential movement was found in the sagittal plane, hip and ankle landing positions. This was more notable when landing flat-footed or with a flexed hip posture. Further assessment of the sagittal knee position between injured athletes compared with a healthy control group however showed no distinguishable difference between how the injured athletes moved compared with the healthy ones; although there did seem to be slightly more extended knee position in the ACL injured group. According to Waldén et al. (2015), in the coronal plane, knee abduction (or valgus), was notably more pronounced for participants with an ACL injury. Video analysis of 39 international football players taken by Waldén et al. in 2015, showed there to be a connection between knee abduction and ACL injury, valgus collapse in these cases however was rare.

In a very recent study by Zhang, Hacke, Garrett, Liu, and Yu (2019), the locations of bone bruises detected by MRI scans resulting from ACL injury were used to determine the mechanism of this injury. The researchers found that the tibial anterior translation was a primary mechanism in most ACL injuries. In addition, they found that knee valgus occurred following anterior translation of the tibia. The locations of the bone bruise also demonstrated hyper-extension of the knee joint, which have been identified to be another mechanism associated with non-contact ACL injury.

## 3.10 Management of ACL injuries

### 3.10.1 Surgical

The treatment of ACL injuries involves surgical reconstructions and rehabilitation or non-surgical interventions with rehabilitation only (Beynnon, Johnson, Abate, Fleming, & Nichols, 2005). Collins, Katz, Donnell-Fink, Martin, and Losina (2013) have reported that the general prevalence rate of ACL-RC is approximately 23 percent within three years of acquiring an injury. The latest systematic review undertaken on the topic indicates that following ACL-RC, 81 percent of patients are likely to return to some form of sport, with 55 percent returning to competitive sport (Ardern, Taylor, Feller, & Webster, 2014). Lohmander et al. (2007) stated that even though ACL-RC is regarded as a gold treatment method for ACL rupture, over 50 percent of patients who underwent this treatment experienced post-traumatic knee OA within twenty years of this surgery. It is known that many variables can impact upon the likelihood of developing post-traumatic knee OA, which include intra-articular pathogenic processes that are instigated when the patient is injured, in conjunction with mechanical failures. Following ACL-RC, abnormal knee kinematics is considered to be a contributor to degenerative processes, as explains Andriacchi, Koo, and Scanlan (2009) and Andriacchi and Mündermann (2006), as a result of the cartilage bearing abnormal loads. Therefore, as it is has become apparent that ACL-RC does not always result in knee OA, understanding how the variations in mechanical factors affect the risk of developing knee OA post-ACL-RC, if at all, will be of significant importance to the field.

In addition, patients who go through ACL-RC tend to have a marginally higher risk of developing symptoms related to knee OA than those who undergo non-surgical therapy (Smith, Postle, Penny, McNamara, & Mann, 2014). Moreover, this evidence is limited, and the frequency of OA among patients treated with surgical reconstructions in comparison to non-surgical therapy is still imprecise (Keays, Newcombe, Bullock-Saxton, Bullock, & Keays, 2010; Smith et al., 2014).

### 3.10.2 Non-surgical (Conservative)

In 2014, Smith et al. conducted a systematic review which uncovered little proof to support the notion that ACL-RC is a better method of recuperation as opposed to conservative management. As a result, many patients decided not to opt for ACL reconstruction; some of this cohort who chose conservative management instead were able to successfully regain useful functioning. Similar findings emerged in a recent study conducted by Ellis, Vite, and Wilson (2018). While not as relevant in the USA, the use of conservative management in ACL injuries could be an ideal treatment method for some patients following ACL rupture.

In 1983,Noyes, Butler, Paulos, and Grood first proposed the idea of differential patient presentation following ACL injury. The ‘rule of thirds’ was discussed, a hypothesis which states that up to one third of patients suffering from ACL deficiency are able to perform well, and display levels of pivoting and cutting activity while avoiding instability, also known as ‘giving way’. This cohort, as a result, demonstrated the possibility of functioning without the need for ACL-RC to help rectify the mechanical instability of the joint. An additional hypothesis from the study proposed that the other two thirds of the patients with ACL deficiencies are only able to function adequately, with the Activity of Daily Living Scale (ADLS) requiring some modification of activity. Otherwise, the patients would not be able to perform any sort of movement due to ACL deficiency as a result of joint instability. Surgical reconstruction would be recommended for these kinds of patients. Essentially, the published document outlines the fact that a range of functional instability may be present in patients following an ACL injury, which could vary from grossly unstable to functionally stable with no giving way. While acknowledging the possibility that some patients are able to recover without the need for ACL-RC, the original work by Noyes et al. (1983) failed to advance any form of method for identifying the most likely patients needing surgical reconstruction before making an attempt at re-engaging with the activity.

A commitment activity modification is a viable solution for those patients who do not wish to undergo surgery following their injury and would rather use conservative treatment. In 1994, Daniel et al., stated that the population focus had previously been on patients who wished to return to level I/II pivoting and cutting activities. Activities which require this level of movement include a variety of sports ranging from football, to basketball, soccer and skiing, all of which call for a high level of dynamic stability in the knee joint, according to Bogunovic and Matava (2013), in order to 'perform adequately'. Those patients who prefer to live a more inactive lifestyle will not require their knees to perform such physically demanding actions, as well as people who choose 'straight-line' sports such as jogging and cycling. These kinds of people are more likely to succeed with this treatment without the need for surgery. Furthermore, the most likely candidates to successfully undergo this treatment are those with no concomitant injuries to the knee, further compromising joint stability and joint health (Bogunovic & Matava, 2013). Patients choosing to modify their activity will have their rehabilitation centred around dealing with post-acute injury impairments, improving joint strength and making sure that the patient will not experience any sort of functional instability or giving way during their activities of choice.

### 3.10.3 Surgical versus Non-surgical (Conservative)

Monk et al. (2016) undertook a systematic review of adults presenting with acute ACL injuries who have had one of two treatment intervention types, namely surgical management, which comprised ACL reconstruction accompanied subsequently by structured rehabilitation, or conservative treatment, which consisted solely of structured rehabilitation. Self-reported knee functioning outcomes at two and five years post-treatment failed to demonstrate any differences between the two treatment types. However, care must be exercised in the interpretation of these results, as a number of respondents whose ACL rupture symptoms were not alleviated on completion of rehabilitation, subsequently availed of ACL reconstruction surgery. Therefore, additional studies need to be conducted in this area, similar to the aforementioned ongoing trials, in order to build a strong evidence base, as the current small-scale research is limited in its nature as it solely focuses on a young, active cohort.

### 3.11 Biomechanical characteristics of ACL-RC

According to Hart et al. (2016), patients with ACL-RC have been shown to walk with greater peak knee flexion angles throughout the gait cycle than controls in the early stages, usually within six months or less, but with less flexion (more extended) one-year post-surgery. Regarding joint loading in the frontal plane, external knee adduction moments are frequently useful in measuring load distribution on this plane (Schipplein & Andriacchi, 1991). Moreover, instances of augmented peak adduction moments in the movement have been linked with advancement (Foroughi, Smith, & Vanwanseele, 2009; Miyazaki et al., 2002) and severity of medial tibiofemoral compartment OA (Sharma et al., 1998). Research on peak adduction moments has reported inconsistent outcomes in participants with ACL-RC, fluctuating from low to high (Patterson, Delahunt, & Caulfield, 2014; Webster & Feller, 2012a). In addition, some patients who have undertaken ACL-RC report exhibiting reduced peak knee flexion, reduced external knee flexion torque and augmented GRF in the course of single-leg landings (SLL) in comparison to healthy individuals (Tsai, McLean, Colletti, & Powers, 2012).

In a very recent study conducted by Lin, Tang, Tan, and Cai (2019), ACL-RC knees demonstrated abnormal kinematics of the patellofemoral joint in comparison to normal knees and ACL-injured knees when ascending the stairs. The researchers used a fluoroscopic imaging system to measure the kinematics of the patellofemoral joint. They also found that during knee flexion, the reconstructed knees demonstrated increased patellar external rotation, with lateral tilt and lateral translation. These biomechanical changes may lead to degeneration of the patellofemoral joint regardless of the reconstruction intervention used.

Furthermore, individuals who have undergone ACL-RC have been reported to demonstrate differences in the kinematic and kinetic motion of their affected knees as compared to the movement of the contralateral knees and those of healthy controls (Shabani et al., 2015). These differences in kinematic and kinetic motion are clear indicators of changes in movement. To understand these changes, it is important to investigate the variability in movement. According to James (2004), Konradsen (2002); Konradsen and Voigt (2002), movement variability (MV) is one variable that affects the development and continuation of injury. McLEAN, Neal, Myers, and Walters (1999) have suggested that increased variability corresponds with the development of other musculoskeletal injuries, which include ACL rupture. Other studies, in particular, by James, Dufek, and Bates (2000), Heiderscheit, Hamill, and van Emmerik (2002), and Hausdorff, Rios, and Edelberg (2001) also have linked increased variability with lower extremity overuse injury, patellofemoral pain and fall injuries, respectively.

## 3.12 Methods

### 3.12.1 Movement variability (MV)

#### 3.12.1.1 Introduction

Intra-individual variation between repeated movement patterns, such as a hopping task or stride cycle, is considered as comprising a core component of all motor tasks (König, Taylor, Baumann, Wenderoth, & Singh, 2016). Stergiou and Decker (2011) noted that such movements can accommodate the adaptation to stress imposed upon the body in a flexible manner. As such it is thought that that MV performs an important function in relation to musculoskeletal injury (Bartlett, Wheat, & Robins, 2007; Stergiou & Decker, 2011). MV has been described as variations that typically emerge in motor performance arising out of activities that are carried out on numerous occasions (Stergiou & Decker, 2011).

Multiple definitions of human MV have been proposed, and, according to Mukherjee and Yentes (2018) most can be classified into two broad types. The first describes variability as a measure of the variance being observed, where the mean is calculated as the middle of the distribution range. Variability is usually determined using several methods which generally includes standard deviation (SD) and coefficient of variation (CV) calculations. The second type focuses on the fluctuations in movement that can occur, as well as examining their pattern and meaning.

In fact, variability within and between measurements in data derived from biological systems cannot be avoided. Kantz and Schreiber (2004) distinguished between two distinct forms of variability, which are noise and variation caused by measurement error and intrinsic dynamics of the system, respectively. Measurement noise has been the main source of variability within the context of motor control and biomechanics studies. Nevertheless, the functional dimensions of variability associated with nonlinear systems have begun to attract increasing attention in recent years, following advances in nonlinear dynamics and chaos theory (Glass, 2001). Furthermore, Latash, Scholz, Danion, and Schöner (2002) noted that various different arrangements of multiple elements or degrees of freedom (e.g. limb segments, joints, muscles and motor units) when engaging in coordination activities can produce comparable results.

#### 3.12.1.2 Movement variability in sports biomechanics

Different levels of movement configuration can all produce MV, which arises both amongst different individuals and within the same individuals (Bartlett, 1997; Bartlett, Wheat, & Robins, 2007; Bates, 1996; Hatze, 1986; James, 2004; Müller & Sternad, 2004; Newell, Deutsch, Sosnoff, & Mayer-Kress, 2006). Irrespective of the efficiency with which a movement is carried out, modifications will inevitably occur each time the movement is made. The field of sports biomechanics has begun to pay ever-increasing attention to the investigation of MV, in light of the significant role played by MV and coordination variability in sports movement examinations, as highlighted by many recent studies in the field (Baida et al., 2018; Mukherjee & Yentes, 2018; Mohr, Meyer, Nigg, & Nigg, 2017; Bates, 2010; Pollard et al., 2005; Preatoni, 2010; Wilson, 2009; Bartlett et al., 2007; Hamill et al., 2007).

Details regarding movement organisation and performance may be crucial in sports, especially in high-performance sports. Movement command is elevated to an exceptionally high degree by elite athletes, whose motor outcomes can be described as repetitive and patterned. Nonetheless, slight dissimilarities can be noted and minor alterations may accumulate in time due to various factors, including environmental factors, training processes, learning mechanisms, hidden pathological conditions or partial recovery. Variability may conceal such factors, which is why MV examination in sports is so important. This issue should be explored in terms of the information it can provide in relation to the biomechanical qualities of athletes, in addition to the aspects of reliability and suitable experimental approaches.

#### 3.12.1.3 Movement variability and noise

The relationship between MV and noise warrants detailed investigation. Even though numerous theoretical and experimental studies over the last twenty years have provided evidence to support the lack of a correlation between variability and noise (K M Newell et al., 2006; Slifkin & Newell, 1998), more recent research conducted by Bradshaw, Maulder, and Keogh (2007) seems to employ the terms 'variability' and 'noise' interchangeably. The standard assumption has been that MV requires minimisation or removal due to being a negative side-product of random noise in the central nervous system (Faisal, Selen, & Wolpert, 2008; Harris & Wolpert, 1998; Van Beers, Haggard, & Wolpert, 2004). Other studies, however, have provided an alternate view, namely, that rather than being a product of undesired noise or error in the movement system, variability in motor output is actually beneficial to the movement system, endowing it with great flexibility and adaptability (Newell & James, 2008; Riley & Turvey, 2002; Slifkin & Newell, 1998). Nonetheless, this does not mean that MV does not arise from noise to a greater or lesser extent or that performance is adversely affected by all noise (Davids, Shuttleworth, Button, Renshaw, & Glazier, 2004). Similarly, it cannot be implied that motor performance is favourably influenced by all variability. These issues call for further research to determine the exact relationship between variability and noise, as well as the extent to which they contribute to task achievement (Glazier, 2011).

#### 3.12.1.4 Assessment of variability

Any study of sports-related skills must be very specific in its description of the motor skills demonstrated by an athlete. To achieve this, the researcher may need to identify either discrete or continuous variables in an attempt to outline the athlete's specific kinematic and kinetic movements. The assessment of variability in discrete and continuous variables should be undertaken independently.

##### 3.12.1.4.1 Discrete measures of variability

According to Dowling and Vamos (1993) and (Bartlett et al., 2007), discrete parameters have long been used as a way of assessing performance in athletes, with a quantitative biomechanical method being applied to extract the parameters from kinematic and kinetic curves. The analysis of these discrete measures is then used to define the composition of a specific motor task in order to understand the characteristics of differing groups. Granata, Marras, and Davis (1999), James et al. (2000) and Nigg and Bobbert (1990), state that these discrete measurements can then also be used to reduce or prevent future injury.

Many different methods have been proposed to measure the variability within kinematic and kinetic parameters. It is known that the application of standard deviation (SD), a method outlined by Kao, Ringenbach, and Martin (2003), Owings & Grabiner (2004) and Fleisig, Chu, Weber, and Andrews (2009) and the coefficient of variation (CV) as proposed by Queen, Gross, and Liu (2006) and Bradshaw et al. (2007) have been used as estimations of spread in conjunction with quantitative motion analysis. There have been a variety of approaches to this in the past, however this method is normally advocated for the estimation of variability within kinematic and kinetic dimensions. Furthermore, Chau and Parker (2004) and Chau, Young, and Redekop (2005) have suggested that non-parametric parameters, such as the inter-quartile range (IQR) or the median absolute deviation (MAD) are also used for variability assessment.

**3.12.1.4.2 Continuous measures of variability**

According to Sutherland, Kaufman, Campbell, Ambrosini, and Wyatt (1996), Queen et al. (2006) and Ryan, Harrison, and Hayes (2006), the taking of single measurements from a continuous variable can result in a significant amount of data being lost or disregarded, some of which may have proven useful for the study. If an individual performs the same motions time after time, soon we will begin to see a family of curves that may differ slightly in their size and duration, and we will not see a set of perfect kinematic or kinetic lines that complement each other in a uniform manner. The application of discrete variables for the assessment of human motion is a strong indicator, however, on its own may not prove significant enough to offer a conclusive assessment of that particular movement. In fact, the outline of kinematic and kinetic curves are often good representations of the way in which a motor task is undertaken and may also assist physicians in the categorisation of whether or not the patient's manifestations can be regarded as physiological or pathological, or for a trainer to be able to assess their motions over time as a way of noticing any changes.

The research into variability within continuous variables is less prevalent than the data which looks at discrete parameters. Kadaba et al. (1989) and Growney, Meglan, Johnson, Cahalan, and An (1997), have suggested that researchers tend to look primarily at the methodological elements of data, repeatability or any variation, which are according to Steinwender et al. (2000) set between 'normal' and pathological patients. Other researchers look at the issue of variability over the curves when they are trying to outline specific patterns of motion, however any analysis like this tends to stop at the point of outlining the main characteristics of a group of curvatures with the assessment of confidence bands, for example, mean curves ^ a multiple of the SD.

It can be seen throughout the existing research, the common usage of repeatability in continuous variables which are the coefficient of multiple correlation (CMC) (Kadaba et al., 1989) and the intra-class correlation coefficient (ICC) (Duhamel et al., 2004; Ferber, McClay Davis, Williams Iii & Laughton, 2002). The assessment range runs from 0, which represents a very low chance of being repeated to 1, for a highly likely chance of perfect repetition. This approach necessitates many tests, if only to validate the intra-session variability. Growney et al. (1997), for example, called upon the usage of three separate trials conducted over a three-day period by Queen et al. (2006) whilst simultaneously referencing two further separate sessions which included as many as six tests each. Alternatively, the ICC can be calculated, according to Duhamel et al. (2004), when data collected from one specific session becomes available, at this stage it can be regarded as the 'proportion of variance due to the time-to-time variability in the total variance'.

#### 3.12.1.4.3 Dynamical systems theory approach to measure variability

In a dynamical system, when in the presence of multiple degrees of freedom, variability in performance is a fundamental condition for optimality and adaptability. Variability patterns, as identified by the gait analysis parameters, include elements such as stride length and frequency and may not necessarily reflect the variability patterns as identified by individual segmental coordination. A study by Van Emmerik, Wagenaar, Winogrodzka, and Wolters (1999) demonstrated this very same notion when he looked at motions in Parkinson's disease. Furthermore, studies by Hamill, Haddad, Heiderscheit, Van Emmerik, and Li (2006) and Hamill et al. (2007) which used a biomechanical research method on injuries sustained through running and many other studies have now successfully demonstrated a link between reduced CV and orthopaedic conditions.

In a dynamical systems approach, the reconstruction of the so-called state space is a fundamental element which is required to pinpoint the main factors behind how a system behaves (Preatoni et al., 2013). The state space relates to the representation of the relevant variables that help to identify the key features of the system. It is demonstrated by firstly the angle-angle plot and secondly with the position-velocity plot. An angle-angle, for example, sagittal plane knee angle versus ankle angle plot, can show areas where coordination adaptations may occur along with other areas of the gait cycle where there is minor variation amongst the pattern of coordination. Heiderscheit et al.(2002) suggested that these coordinative alterations within the angle-angle plots may be quantified even further through a method of ‘vector coding techniques’. Another form of state space can be represented by the placement and velocity of a segment of a joint can be plotted graphically relative to each other. This area is also known as the phase plane, which according to Hamill, van Emmerik, Heiderscheit, and Li (1999) is the first crucial step in the ‘quantification of coordination using continuous relative phase (CRP) techniques’.

According to Hamill et al. (1999), CRP may regularly be regarded as a higher order assessment of the coordination between two segments or two joints. This higher order emerges from the derivation of CRP from the movement dynamics when in the phase plane of the two joints or segments. The usage of CRP analysis stems from its fundamental signification which enables the characterisation of joint or segmental coordination during gait analysis (Hamill et al., 1999; R. E. A. Van Emmerik et al., 1999). Despite the relative ease with which CRP may be seen to be implemented, many key concepts which relate to the methodology and analysis must be more clearly investigated and addressed. Primarily, CRP is, according to Peters, Haddad, Heiderscheit, Van Emmerik, and Hamill (2003), not a higher approach to resolution within the discrete relative phase, it represents the calculated coordination between two oscillators formulated upon their phase plane angles. It should be reiterated that movement of the joints and segments is not actually an oscillator, but it is designed to mimic the behaviour of oscillators.

As CRP is often considered a higher order measure of the coordination between two segments or two joints and not a higher resolution form of discrete relative phase (DRP) (Peters et al., 2003) as well as this thesis mainly focuses on variability between discrete variables; the peak angles and movement at knee joint associated with the increased risks of ACL injuries; CRP was not used and CV was employed. CV is the most popular approach, with the SD normalised to the distribution's mean, and the mean changed to a percentage to show the relative or normalised variability of the scores (Brown, Bowser, & Simpson, 2012; Brown, Padua, Marshall, & Guskiewicz, 2009). Statically, CV is known as 'relative variability' and generally considered as an accurate measure for variability; this explains its popularity. Therefore, for this thesis, CV was used as the main measurement of variability. In addition, the focus of this PhD investigation is on tasks which are commonly associated with screening programmes for ACL injuries.

#### 3.12.1.5 How has variability been investigated?

In the literature, there are several studies that have investigated MV by examining functional motor skills such as sidestep cutting  (Edwards, Brooke, & Cook, 2017a; Pollard et al., 2018), running (Gribbin et al., 2016b; Hamacher, Hollander, & Zech, 2016; Mann et al., 2015; Meardon, Hamill, & Derrick, 2011; Paquette, Milner, & Melcher, 2017), walking (Chau et al., 2005; Gribbin et al., 2016b) and landing (C. Brown et al., 2012; Kipp & Palmieri-Smith, 2012; Kulig, Joiner, & Chang, 2015) however, other researchers have analysed the injury-related aspects (Hamill et al., 2007, 1999), coordinative patterns (Seay, Haddad, Van Emmerik, & Hamill, 2006) or OAs (Chiu, Lu, & Chou, 2010). The variability of discrete  kinematic and kinetic variables during movements related to sporting activities has been less extensively investigated compared to the reliability of common walking variables (Benedetti, Catani, Leardini, Pignotti, & Giannini, 1998; Chau et al., 2005; Dingwell & Cavanagh, 2001; Growney et al., 1997; M P Kadaba et al., 1989; Mrn P Kadaba, Ramakrishnan, & Wootten, 1990; Steinwender et al., 2000; Stolze, Kuhtz-Buschbeck, Mondwurf, Jöhnk, & Friege, 1998; Winter, 1984). The plethora of motor tasks associated with various sports makes this scarcity of research even more poignant. The movements that have attracted the most research attention are landing (James et al., 2000; Rodano & Squadrone, 2002) and running (Bates, Osternig, Sawhill, & James, 1983; DeVita & Bates, 1988; Diss, 2001; Ferber et al., 2002; Lees & Bouracier, 1994; Queen et al., 2006).

All the above-mentioned studies reported conflicting results due to the differences in methods, functional tasks and participants. For example, Preatoni et al. (2010) included seven healthy race walkers in their study, which could be considered a small sample size. Furthermore, they used sample entropy estimation (SampEn) to examine variability in lower limb angles and GRF. The primery focus of the study was the regularity of kinematic and kinetic time-series patterns during race walking using 3D motion analysis. According to the results, variability during race walking was determined to not only be the product of random noise, but may also contained information regarding the inherent propriety of the neuro-musculo-skeletal system as the authors stated. Bradshaw et al. (2007) also investigated variability among ten healthy athletes using 2D motion analysis. They captured data during the start of sprints, and measured variability using the CV. All participants were asked to perform four 10 m sprints. The researchers concluded that the variability in task outcome measures (linear kinematics) was considerably lower than that observed in joint rotation velocities (angular kinematics).

In a recent study conducted by Edwards et al. (2017), MV was investigated during an anticipated sidestep cutting task among healthy players, as well as players with a history of groin pain. The sample comprised ten healthy participants and a further seven with a history of groin pain. Lower limb and trunk kinematics and kinetics data were obtained using 3D motion analysis. Variability in lower limb kinematics and kinetics was measured using CV over ten trials. The results showed that the cohort with a history of groin pain generally demonstrated lower MV than the healthy group at knee, ankle, L5–S1 and T12–L1 joint angles, but not for all variables. The groin pain group also demonstrated lower variability in GRFs and most moment peaks of knee, ankle and hip joints. The researchers concluded that the management of groin pain should clinically involve the thoracic segment along with the lower limbs in order to improve the movement pattern in these patients.

Brown et al. (2012) investigated MV during single-leg landings (SLL) in individuals with and without ankle instability. A total of eighty-eight individuals (39 males, 49 females) participated in this study, and they were subsequently divided into four groups: functionally unstable, copers, mechanically unstable and controls. Lower extremity kinematics and kinetics data were collected using 3D motion analysis. Every participant in this study performed ten successful trials and the MV in these trials was measured using CV. A mixed model of variance (ANOVA) was used to determine the variance between groups. The results showed that the group with ankle instability demonstrated less variability at the knee and hip than healthy individuals during SLL tasks. The researchers were of the belief that less MV at the proximal joints (hip and knee) during landing may result in ankle instability episodes.

The study conducted by Fleisig et al. (2009) investigated inter-individual variability in baseball pitching among participants of various skill levels using 3D motion analysis. In this study, 93 healthy male baseball pitchers were included and divided into five groups based on their respective levels of competition. The SD of the five pitches of each participant was then determined. The researchers then conducted three multiple analyses of variance (MANOVAs) on the SD values to determine the differences among the five levels of competition. The results showed that pitchers who had advanced to higher levels exhibited less variability in their motions.

James et al. (2000) also investigated the effects of lower limb overuse injuries and landing height on variability using 2D motion analysis. The authors recruited twenty participants (both healthy and injury prone) for the study. Each participant was asked to perform ten double landings from different heights. The variability for each variable was determined by calculating the mean absolute difference of the individual trials within a deviation from the central mean. The results showed that the injury-prone group demonstrated greater variability for the peak ankle joint moment magnitudes, whereas the healthy group showed greater variability for time to peak ankle joint moment. Based on these results, the researchers suggested that a possible relationship exists between overuse injuries and variability. Such findings support the notion that MV may play a role in injury. However, this study defined injury-prone individuals as those with a history of lower limb overuse injury, which means that they included participants with different lower extremity injury types across several anatomic locations. This might have affected their results, as different injuries display different characteristics.

Gribbin et al. (2016) examined intersegmental coordination using a hip-knee joint coupling method during walking and jogging in two groups, namely individuals with ACL-RC and a healthy sample. Coordination variability was the variability measure investigated in this study. The research was underpinned by the dynamic system theory (see section 2.11.4.3), where the researchers used a vector coding method to measure hip-knee joint coupling (intersegmental coordination). They chose six joint couples: hip frontal and knee sagittal, hip frontal and knee transverse, hip frontal and knee frontal, hip sagittal and knee transverse, hip transverse and knee frontal planes and hip sagittal and knee frontal planes. They also used an intersegmental variability coefficient, which is also known as a vector coding variability, to calculate the coordination variability between ten full gait cycles. Results showed that the reconstructed group demonstrated lower coordination variability in hip sagittal-knee transverse and hip sagittal-knee frontal joint couples during the mid-stance phase. However, increased coordination variability was found to occur in all the other couples during the stance phase.

In a recent study conducted by Paquette et al. (2017), variability in the foot contact angle was investigated among healthy runners and runners with a history of injury, as well as between habitual rearfoot strike and non-rearfoot strike runners. The researchers divided the participants into two groups: a control group (14 male, 7 female) and those with a history of injury (10 male, 13 female). All participants ran for forty minutes on a treadmill where kinematic data were collected using 3D motion analysis. The data were obtained from five consecutive foot strikes during four different time points (10, 20, 30 and 40 minutes). Variability in the foot contact angle was measured using the standard deviation (SD) method. The findings showed that no differences occurred in the two groups (with and without a history of injury) in both their foot contact angle and variability in their foot contact angle. However, there was a significant difference in the variability of their foot contact angle between non-rearfoot strike runners and rearfoot strike runners ($P < 0.001$). This led the authors to conclude that the lower variability in non-rearfoot strike runners may increase the risk of injury. However, as this study used a treadmill, it does not reflect the real environment in which runners normally perform. As a result, this may have affected the data they generated.

Lees and Bouracier (1994) conducted a longitudinal study to investigate variability for four GRF variables in three separate test sessions. The researchers divided 14 healthy subjects into two groups: experienced runners and inexperienced runners. The results showed that, generally, inexperienced runners demonstrated greater variability than experienced runners, suggesting that they were using a less economical action.

Mohr et al. (2017) have investigated variability during a running task. The authors aimed to study the relationship between the perceived comfort of running footwear and variability in kinematics variables. Thirty-six recreational athletes (18 females and 18 males, age: 25.4 ± 3.5) participated in this study. The authors used the coefficient of variation (CV) method to identify variability. The results demonstrated that an individual's perception of decreased comfort, whether perceived or real, was associated with a reduction in the variability of kinematics variables, particularly during the late swing phase of running. Such findings highlight the effects of footwear on MV from one particular perspective, and this could lead to the exploration of a new investigation topic, namely to establish whether footwear is an appropriate intervention if MV is targeted.

Similar to the previous study, Enders, Maurer, Baltich, and Nigg, (2013) investigated variability using CV. The purpose of this study was to investigate the effects of different biomechanical constraints on the variability of muscle activation during cycling. The researchers studied fifteen male cyclists. The findings demonstrated that during constant load cycling, muscle activity can be divided into high-variability and low-variability after a principal component analysis (PCA) is applied to the entire dataset. In addition, the structure of the variability found that dynamic muscle use is similar to those observations made during static EMG measurements.

#### 3.12.1.6 Movement variability and ACL injury

It has been suggested that MV may increase the likelihood of injury occurring (Baida, Gore, Franklyn-Miller, & Moran, 2018; James, 2004; Konradsen, 2002; Konradsen & Voigt, 2002). Increased variability was suggested to be correlated with risk of further musculoskeletal injuries, such as ACL rupture (McLean et al., 1999), lower extremity overuse injury (James et al., 2000), patellofemoral pain (Heiderscheit et al., 2002) and injuries from falls (Hausdorff et al., 2001). According to studies of MV and the possible role it can play in musculoskeletal injuries, it is important that investigations carried out in this area focus specifically on the knee joint where an ACL injury can occur. This can be achieved by examining all biomechanical variables and, in particular, the variables which have been shown to be risk factors for an ACL injury, such as knee valgus angle, knee valgus moment and the GRF, all of which are worthy of investigation. Variability within these factors should be taken into consideration, as they are all are risk factors and, therefore, any significant changes in movement when the individual performs the same tasks may increase the risk of an ACL injury.

The most frequently investigated tasks in relation to sport injuries are landings, squatting, running and sidestep cutting. This is because the majority of ACL non-contact injuries occur during two of these tasks, namely cutting and landing, while the two other tasks, running and squatting, are important to select in order to draw comparisons between complex and less complex tasks.

### 3.12.2 Screening of knee biomechanics

#### 3.12.2.1 3D motion analysis

3D motion analysis is considered to be a gold standard method by which to evaluate joint kinematics and kinetics, when carrying out an injury risk assessment (Munro, Herrington, & Carolan, 2012). It allows the researcher to measure movement in three planes: sagittal, frontal and transverse. In addition, kinetics data (GRF and joint moments primarily) can be obtained using this system, when incorporated with force platforms, and 3D motion analysis is also commonly applied in studies which investigate lower limb biomechanics (Blackburn & Padua, 2008; Ford et al., 2003; Hewett et al., 2005). Notwithstanding its benefits, in practice, 3D motion analysis tends not to be employed in clinical settings or in population-wide sport screening programmes due to its expensive nature, as well as its associated spatial and temporal costs. Alternatively, 2D analysis methods are used instead, as they are cheaper, and rely upon more user-friendly and transportable equipment. The latter has been applied to measure knee valgus angles in both athletes and those who have experienced an injury.

Despite the widespread adoption of 3D motion analysis, researchers should be aware of its reliability and validity capacity in order to use it appropriately (Rothstein & Echternach, 1993). McGinley, Baker, Wolfe, and Morris (2009) have noted that variations generally occur in movement measures and they attribute these to the margin of error that usually exists. Throughout the literature, a large body of studies have investigated the reliability of 3D motion analysis in terms of either its within-day or between-day effects in different conditions. It has been shown that within-day reliability has demonstrated better reliability than between-day (Kadaba et al., 1989). Researchers believe that this is due to the error in marker placement, which exerts a greater influence on between-day reliability (Ferber et al., 2002; Ford, Myer, & Hewett, 2007; Queen et al., 2006). In addition, skin movement artefact can also be a source of error (Cappozzo, Catani, Leardini, Benedetti, & Della Croce, 1996). Finally, this method is considered to be a gold standard method by which to evaluate joint kinematics and kinetics, when carrying out an injury risk assessment during different sporting tasks (Munro, Herrington, & Carolan, 2012).

### 3.13 Screening tasks in assessing the risk of knee injury

Numerous screening tasks have been utilised in the literature to assess the biomechanical risk factors associated with knee injury. Some of the commonly investigated tasks are Single leg landing (SLL) (Chinnasee, Weir, Sasimontonkul, Alderson, & Donnelly, 2018; Donohue et al., 2015; Grooten, Karlefur, & Conradsson, 2019; Ithurburn, Paterno, Ford, Hewett, & Schmitt, 2017; Jenkins, Williams III, Williams, Hefner, & Welch, 2017; Johnston, McClelland, Feller, & Webster, 2017; P. T. Johnston, McClelland, & Webster, 2018; Jones, Herrington, Munro, & Graham-Smith, 2014; Kajiwara et al., 2019; Kiratisin, Sinsurin, & Vachalathiti, 2018; Lawrence III, Kernozek, Miller, Torry, & Reuteman, 2008; Lessi & Serrão, 2017; A Munro et al., 2017; Shimokochi, Hosaki, Takiguchi, & Ogasawara, 2018; Tamura et al., 2017; Teng et al., 2017), SLS (Garrick et al., 2018; Herrington, 2014; Herrington, Alenezi, Alzhrani, Alrayani, & Jones, 2017; A Munro et al., 2017; Rees, Younis, & MacRae, 2019; Ugalde, Brockman, Bailowitz, & Pollard, 2015; Willson & Davis, 2008; Zeller et al., 2003),; running (Alenezi, Herrington, Jones, & Jones, 2016; Asaeda et al., 2018; Besier, Lloyd, Cochrane, & Ackland, 2001; Bohn, Petersen, Nielsen, Sørensen, & Lind, 2016; K. Markolf, Yamaguchi, Matthew, & McAllister, 2019; Pamukoff, Vakula, Moffit, Choe, & Montgomery, 2017; Perraton et al., 2018; Savage et al., 2018; Tashman, Collon, Anderson, Kolowich, & Anderst, 2004; Tashman, Kolowich, Collon, Anderson, & Anderst, 2007; Thakkar et al., 2018; Thomson, Einarsson, Hansen, Bleakley, & Whiteley, 2018)  and sidestep cutting (Dos'Santos, McBurnie, Thomas, Comfort, & Jones, 2019; Bencke et al., 2013; Benjaminse, Welling, Otten, & Gokeler, 2017; Chang, 2018; Craft, 2017; K. L. Havens & Sigward, 2015; Jones et al., 2014; Kaila & Irwin, 2017; Kristianslund, Faul, Bahr, Myklebust, & Krosshaug, 2014; Mok, 2015; Nagano, Ida, Ishii, & Fukubayashi, 2015; Pace et al., 2015; Pollard et al., 2018; Pollard, Stearns, Hayes, & Heiderscheit, 2015; Samaan, Ringleb, Bawab, Greska, & Weinhandl, 2016; Savage, Lay, Wills, Lloyd, & Doyle, 2018; Schreurs, Benjaminse, & Lemmink, 2017; Sigward, Cesar, & Havens, 2015; Sinclair, Brooks, & Stainton, 2019; Tamura, Akasaka, & Otsudo, 2019; Thakkar et al., 2018)

### 3.13.1 Single Leg Landing (SLL)

Landing tasks, which can be either double-leg or single-leg, have been extensively used in the literature to investigate the risk of knee injury. Hewett et al. (2005) and Padua, Bell, and Clark (2012) have demonstrated that the kinematic and kinetic variables such as knee valgus angle and moment analysed in landing tasks have been previously used to predict ACL injuries, based on our understanding of ACL injury hallmarks and loading mechanisms. Adolescent girls presenting with increased abduction angles and moments of the knee, high vertical GRF and lower maximum knee flexion angles in double-leg landings, tend to have a higher risk of acquiring an ACL injury (Hewett et al., 2005). Dai, Herman, Liu, Garrett, and Yu (2012) have highlighted that injury prevention methods and data to locate at-risk individuals can be generated by identifying such risk factors.

Whilst, double-leg landings have been beneficial to understanding ACL injury risk factors (Hewett et al., 2005; Padua et al., 2012), in order to analyse the biomechanics of the lower extremities linked to ACL loading and risk factors, SLL been analysed, too (Ali, Rouhi, & Robertson, 2013; Lyle, Valero-Cuevas, Gregor, & Powers, 2014). ACL injuries are usually seen in SLL (Boden et al., 2000; Koga et al., 2010; Krosshaug et al., 2007), and, as explained by Wang (2011) and Yeow, Lee, and Goh (2010 & 2011), studies indicate that ACL injuries occur more frequently with a SLL due to the lower knee and hip flexion angles, and secondly, in contrast to double-leg landings, a weaker load dispersion capability. Moreover, according to Hewett et al. (2005), significant correlations were found to occur between knee abduction angle and moment, which are considered to be risk factors for ACL injury and peak GRF. In contrast to stronger and older sports players, adolescent players such as footballers and basketballers who suffered such an injury displayed a twenty percent higher reaction force (Myer, Ford, Palumbo, & Hewett, 2005). These findings suggest that the higher the vertical ground reaction landing force, the greater the risk of ACL damage, thus analysis of SLL lower extremity biomechanics is highly beneficial.

### 3.13.2 Single-leg squat (SLS)

Aside from landing tasks, researchers have also focused on squat tasks when assessing lower-extremity biomechanics, as this allows them to understand biomechanical factors when performing a less complex task. In addition, it permits them to draw effective comparisons between complex tasks, such as cutting and landing, where ACL injuries tend to occur. Double-leg squats (D. R. Bell, Oates, Clark, & Padua, 2013; Padua et al., 2012) and SLS (F Alenezi, Herrington, Jones, & Jones, 2014; Faisal Alenezi, Herrington, Jones, & Jones, 2014; Atkin, Herrington, Alenezi, Jones, & Jones, 2014; Herrington, 2014; Nguyen, Shultz, Schmitz, Luecht, & Perrin, 2011; Stensrud, Myklebust, Kristianslund, Bahr, & Krosshaug, 2011) have both been evaluated. With less external loading and the slower motions of squats compared to landings, it is unusual to see ACL injuries occur in this exercise. Using squats for such analysis makes sense because, in common with landings, they incorporate both descending and ascending motions. The lower external loading associated with squats is regarded as less harmful than landings, particularly for people who have already suffered an ACL injury (Yamazaki et al., 2010). Moreover, the slower motions in squats enable visualisation of inappropriate motions with greater clarity, and they can offer immediate remedial suggestions. As remarked upon by Hewett et al. (2005) and Padua et al. (2012), however, although potential ACL injury risk factors were ascertained in double-leg landings, ascertaining such risk factors in SLL or squats operates under the assumption that the factors are the same for these different tasks.

In addition, biomechanical data obtained during SLS have been shown to be useful for screening purposes, in terms of assessing lower extremity dynamic alignment and the risk of injury, and a strong correlation has also been found to exist between it and knee abduction during jogging (Whatman, Hing, & Hume, 2011). Alenezi et al. (2014) reported similar results, as well as strong correlations between SLS and sidestep cutting and running in knee valgus angles.

### 3.13.3 Running

Running is integral to most sporting activities. Besides, running speed is critical for many sporting activities, whether it is the capability to beat a defender, outrun opponents, or initiate sufficient take-off velocity to attain distance or height on a jump. Tongen and Wunderlich (2010), define running as a gait in which there is an aerial time, a spell when no limbs are touching the ground. The comprehension of lower extremity biomechanics assists during running and crucial for the prevention and therapy of lower extremity injuries. Apart from wind resistance and gravity, there are no external forces functional to the moving body during this aerial phase. Thus, it is the stance phase of running that must be altered to adjust pace. However, Messier et al. (2008) state that some risk factors have been acknowledged to predict the incidence of running-related injuries, including disproportionate ground reaction forces (GRFs) and joint loads.

In addition, Schache, Bennell, Blanch, and Wrigley (1999) has proposed that the exploration of the biomechanics of running could assist in determining those who are more likely to develop running-related injuries. Studies have also revealed that gender-based differences in running biomechanics can occur. For example, Ferber, Davis, and Williams (2003) and Malinzak, Colby, Kirkendall, Yu, and Garrett (2001) both found that women display a higher peak knee valgus, while later studies by Ferber et al. (2003) and Souza and Powers (2009) demonstrated higher female hip-internal rotation and adduction.

A very important element regarding this task is running speed which also can alter some lower limb biomechanics as explained by Schache et al. (2011). The researchers examined the effects of different running speeds on the lower limb kinetics. There were four running speeds applied in this study; 3.5 m/s, 5.0 m/s, 7.0 m/s and maximum sprinting. The results, in general, showed that a significant speed effect ($P < 0.008$) was observed for 29 of the 33 variables, with the absolute magnitude of these 29 variables increasing with faster running. However, the knee joint during stance was unaffected by increasing running speed, whereas ankle joint was. In terms of ACL injury, this means running speed during forward running may not have that effect on the risk of this injury as this type of injuries usually occurs during stance phase.

### 3.13.4 Cutting tasks

Numerous sports require participants to run and alter direction rapidly, known as 'cutting'. These motions are linked to noncontact ACL injuries (Boden et al., 2000; Cochrane et al., 2007; Krosshaug et al., 2007). As Hewett et al. (2005) and Sigward and Powers (2007) report, certain lower limb motions have been found, especially in knee valgus (adductor) loading, to present a greater risk of injury during cutting movements. We have comparatively less knowledge of the mechanisms required for the best cutting movements or which motions are more or less likely to cause damage. This knowledge is required to formulate training exercises that can lessen ACL injury risk but do not compromise the cutting performance skills. Both pivots and cutting incorporate decelerative and accelerative stages to generate a quick directional change. These quick changes, combined with the deceleration–acceleration movements are widely-seen in encounters where ACL injuries have occurred (McLean et al., 2004).

As Hase and Stein (1999) explain, when moving direction rapidly, a person must decelerate towards the initial direction and then begin moving their body towards the desired position.

The position of the overall body's centre of mass (COM) and the horizontal GRF variables suggest bodily deceleration and translation becomes greater the higher the cut angle (Havens & Sigward, 2015; Havens & Sigward, 2013), although, these actions are achieved via a variety of lower extremity and trunk joint patterns when the cuts are between forty-five and ninety degrees. Therefore, different mechanics are probably linked to more rapid cuts at various angles. To accurately comprehend how training exercises can influence outcomes, it is crucial to analyse the entire body and the kinematics and kinetics of joints during the movements. Alluded to earlier, footballing injuries mostly occur through rapid directional changes, for example, during pivoting and cutting. Owing to this, precise analysis is needed to find which sportspeople are exhibiting bad pivoting and cutting mechanics.

Jones et al. (2014) conducted a study to determine whether there is a relationship between SLL, cutting at 90 degrees and pivoting (180 degree turns) in terms of the characteristics of dynamic valgus. Twenty female soccer players were involved in this study. The researchers found strong correlations for peak knee abduction angles between tasks ($R$ = 0.63-0.86, $P$ \ .01). The results indicate that women athletes who demonstrate poor SLL mechanics perform the same during cutting 90 degree and 180 degree turns.

A recent study which aimed to show the different cutting angles in knee kinematics and kinetics was undertaken by Schreurs et al. (2017) and comprised 13 male and 16 female participants. The study included the examination of five different types of movement: running forwards; and cuts of 45, 90, 135 and 180 degrees. The researchers found that different cuts required different knee kinematics and kinetics, with the sharper cutting angles proving to be a much greater risk than the other manoeuvres. Interestingly, differences also emerged in the results between the male and female participants who were shown to respond differently to the movements. This could have wider effects on future studies and in the dissemination of future injury prevention advice.

In terms of MV during a cutting task, Pollard et al. (2015) investigated it within an ACL-RC population. The study comprised a cohort of ten ACL-RC females and ten healthy females. They applied the dynamic systems theory (see Section 2.11.4.3), where the researchers used a vector coding method to measure hip-knee joint coupling (intersegmental coordination). The results showed that female soccer players with ACL-RC who were returning to sport demonstrated altered lower extremity coupling variability during this task. In addition, Edwards, Brooke and Cook (2017b) investigated MV during an unanticipated cutting task between players with and without a history of groin pain. They used the CV method to measure MV. The results showed that the injured cohort demonstrated lower MV than the control group in the majority of knee and ankle biomechanical variables.

**3.14 Gaps in the current literature**

Previous research literature has investigated lower limb biomechanics following ACL injuries, using different methods, aims, populations and tasks. The majority have studied lower limb kinematics and kinetics in healthy individuals and those with ACL impairments (reconstructed/deficient). Furthermore, some authors have compared lower limb kinematics and kinetics with the contralateral limb and with healthy controls when engaging in different tasks. Aside from various comparative analyses, a number of studies have been conducted to identify risk factors for ACL injury.

In addition to experiencing suboptimal return to sport (RTS) rates and rates of subsequent ACL injury, persons who have undergone ACL-RC reveal persistent asymmetries in their movement in comparison to individuals devoid of a history of ACL-RC. Kinetic and kinematic asymmetries have been reported during walking (Bush et al., 2001; Lewek, Rudolph, Axe, & Snyder-Mackler, 2002; Roewer, Di Stasi, & Snyder-Mackler, 2011); double-leg jumping and landing tasks (Paterno et al., 2011; Paterno, Ford, Myer, Heyl, & Hewett, 2007; Schmitt, Paterno, Ford, Myer, & Hewett, 2015); single-leg jumping and landing tasks (King et al., 2018; Pratt & Sigward, 2017; Gokeler et al., 2010; Myer et al., 2011; Oberländer, Brüggemann, Höher, & Karamanidis, 2014; Orishimo, Kremenic, Mullaney, McHugh, & Nicholas, 2010; Ortiz, Olson, Trudelle-Jackson, Rosario, & Venegas, 2011; Palmieri-Smith & Lepley, 2015; Xergia, Pappas, Zampeli, Georgiou, & Georgoulis, 2013); double-leg squats (Donohue et al., 2015); SLS (Donohue et al., 2015); running (Herrington, Alarifi & Jones, 2017; Pairot-de-Fontenay et al., 2019; Pamukoff et al., 2018; Perraton et al., 2018; Bush et al., 2001; Tashman et al., 2007); and sidestep cutting (King et al., 2018; Pollard et al., 2015; Samaan et al., 2016; Stearns & Pollard, 2013). These asymmetries manifest when involved limbs are equated with uninvolved limbs and uninjured controls. Therefore, it is important to investigate MV to gain a better understanding of these differences.

The primary focus of the aforementioned studies presented above was to identify either differences or correlations in knee and hip mechanics during specific tasks, as well as differences in coordination. Few studies have investigated MV in the lower extremities in relation to sports injuries, particularly ACL injury. For example, Pollard et al. (2015) conducted a study to investigate whether ACL-RC female soccer players demonstrated differences in lower limb coupling variability (coordination variability) as compared to healthy controls during a 45 degree sidestep cutting exercise. In line with the dynamic system theory (see Section 2.11.4.3), the researchers used a vector coding method to measure seven lower extremity intralimb couplings (intersegmental coordination). In this study, MV was defined as the variability of coupling angles across trials and was measured using a SD method. The results showed that the ACL-RC group demonstrated increased lower limb variability as compared to individuals with no history of ACL injury. In this study only one functional sporting task was required, namely a 45 degree sidestep cutting exercise. In terms of risk of ACL injury, this angle was shown to be less risky than 90 and 135 degree angles, as previously explained (Schreurs et al., 2017).

Despite the fact that the vector coding technique can offer worthwhile information in relation to intersegmental coordination and variability in coordination, it cannot provide information on a specific point, such as a peak valgus angle (discrete measure) when linked to risk of ACL injury. Another drawback regarding vector coding is that it is employed for angles (kinematics) only and cannot be used for kinetics (moments). Therefore, it is not an appropriate option to use in studies that involve moments as their main outcome measure. Consequently, this study employed the CV to measure variability, as it can measure variability between trials for a specific point, such as peak valgus angle/moment and other discrete measures that are deemed to be risk factors for ACL injury. By using this method, it is possible to link MV with risk of ACL injury. This was highlighted in a previous study undertaken by Pollard et al. (2015), where they recommended that further research be carried out to investigate MV in relation to the risk of acquiring ACL injury.

Similar to the previous study, Gribbin et al. (2016) investigated intersegmental coordination and its variability in ACL-RC recreationally active individuals as compared to a non-injured cohort, using a vector coding method during walking and running. The researchers included six joint couples and defined variability as variability of coupling angles across ten gait cycles. This was measured using an intersegmental variability coefficient method which is similar to the CV. The results showed that the ACL-RC group demonstrated increased variability of intersegmental coordination, which means that less consistency of movement patterns occurred during walking as compared to the healthy controls. Two tasks were included in this study, namely walking and running, which are considered to be non-risky in terms of ACL injury. Therefore, this study aims to include both risky and non-risky tasks. Notably, studies which investigated intersegmental coordination and its variability usually link it with the risk of developing degenerative OA.

However, to date no study has investigated MV within individuals who have undergone ACL-RC and ACL-DF as compared to healthy subjects, using 3D motion analysis, during these four common sporting functional tasks. In addition, no study has provided reference values for MV in a healthy population. The inclusion of both risky and non-risky tasks is useful, especially as it provides a wide range of insightful knowledge and logical justifications for some of the findings. Similarly, including ACL-RC and ACL-DF cohorts, as well as healthy controls is beneficial in terms of investigating conditions that are related to risk of ACL injury, as it provides additional information pertaining to all populations with this condition. In addition, no study to date has been conducted to establish the relationship between MV and ACL injury risk factors. Providing such information would play a very important role in improving our understanding of MV, along with gaining large attention throughout the development ACL injury prevention programmes and ACL reconstruction rehabilitation. Therefore, this study aims to bridge that gap.

### 3.15 Justification for the male-only data in this thesis

The main reason that only males have been included as participants in this thesis is that the data was collected mainly at the 3D motion analysis laboratory at Imam Abdulrahman Bin

Faisal University in Dammam, Saudi Arabia. Due to cultural and religious issues in that country, it would have been difficult to recruit females for such a study conducted by a male researcher. Most Saudi Arabian women would not consent to wearing shorts in the presence of a man or to having markers placed on their bodies, unless the researcher doing so was female. The university would also not consent to such activities, as they contradict the national culture. Another reason is that there are significant differences in biomechanics between males and females performing functional sporting tasks, as reported by several previous studies and as will be discussed in the next chapters. Due to these differences, it is more sensible to investigate each gender separately. The final reason is that the period of data collection for this thesis was short, rendering it difficult to recruit more than the included number of participants, as three groups had to be investigated doing four tasks for both limbs.

# Chapter 4

# 4 Methods

General research methods are considered to be applicable to all of the studies included in this thesis. Each study has its own specification which will be described in detail later in each chapter. In addition, the data collected using two biomechanical laboratories. In addition, this chapter contains participants, instrumentations, 3D motion analysis laboratories, procedures, explanation of the tasks included in this thesis, data processing, main outcome measures, MV measurement, relation between the two laboratories used in this thesis, and at the end of this chapter is the reliability study.

## 4.1 Participants

The demographic profile of all participants involved in this thesis comprised individuals aged between 18–40 years, who were subsequently divided into three groups:

1- Healthy
2- ACL-DF group
3- ACL-RC group

Subjects were solely recruited from this age group, as according to Griffin (2001), the vast majority of sporting athletes are within this average age range. These athletes are at a higher risk of injury and therefore would benefit most from the study we are undertaking. The individuals included in the study were required to have engaged in a moderate level of physical activity, which has been defined as performing any sport or exercise for at least half an hour, three times a week, over the previous six months (Munro & Herrington, 2011). These requirements are in line with the American College of Sports Medicine guidelines (exercise at least 3–5 times a week, at a moderate intensity, for no less than 30 minutes) (Garber et al., 2011). Participants were also able to maintain a normal balance, described by by Bohannon, Larkin, Cook, Gear, and Singer (1984) as the ability to stand on one leg, with eyes closed, for 30 seconds. Ethical approval was acquired from the University of Salford's Research, Innovation and Academic Engagement Ethical Approval Panel (Ethical approval number: HSCR 16-61) (Appendix 12.16). In addition, a consent form and information sheet were completed by each participant (Appendices 12.9 and 12.12). Additional details about the participants assigned to each group, as well as the inclusion and exclusion criteria adopted, will be explained later in the chapters to follow.

## 4.2 Instrumentation

The thesis utilised instrumentation which comprised two different 3D motion analysis systems.

### 4.2.1 3D motion analysis laboratory at the University of Salford

A ten-camera motion analysis system (Qualisys, Gothenburg, Sweden) operating through Qualisys Track Manager software (Version 1.10.282; Qualisys) sampling at 250 Hz, along with three force platforms (AMTI BP400600, USA) embedded into the floor sampling at 1000 Hz was used to collect the kinematic and kinetic data (Figure 4-1). The Brower Timing Gate System (TC-Timing System, USA) was used to monitor running speeds. In 3D motion analysis, the accuracy of calibration is crucial (Richards, 2008). Therefore, the calibration process requires careful set up.

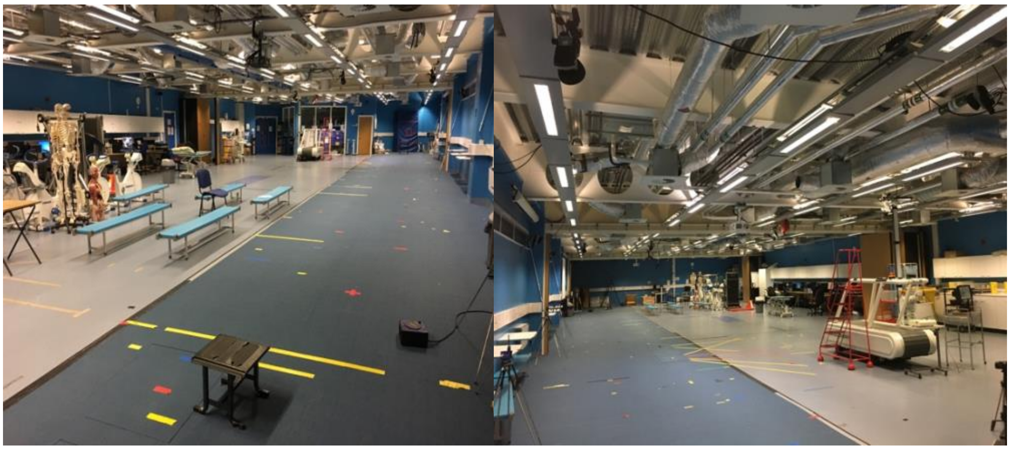


*Figure 4-1 3D motion analysis laboratory at Salford University.*

### 4.2.2 3D motion analysis laboratory at Imam Abdulrahman Bin Faisal University

The 3D motion analysis laboratory at Imam Abdulrahman Bin Faisal University has a ten-camera motion analysis system (Vicon-Bonita cameras, UK) operating through Nexus software (Version 2.5.0.93422h) sampling at 250 Hz, along with two force platforms (Kistler force plate, Type 9286AA, Winterthur, Switzerland) embedded into the floor sampling at 1000 Hz. This was used to collect the kinematic and kinetic data (Figure 4-2). To monitor running speeds, the Brower Timing Gate System (TC-Timing System, USA) was used. In addition, the calibration process was carefully set up.

A set of Brower Timing Systems timing lights were employed for the task of measuring the total time it took a participant to complete the cut and running tasks in both laboratories. As suggested by Yeadon, Kato, and Kerwin (1999), the timing lights were set up at hip height to guarantee that only the lower torso would interrupt the beam.

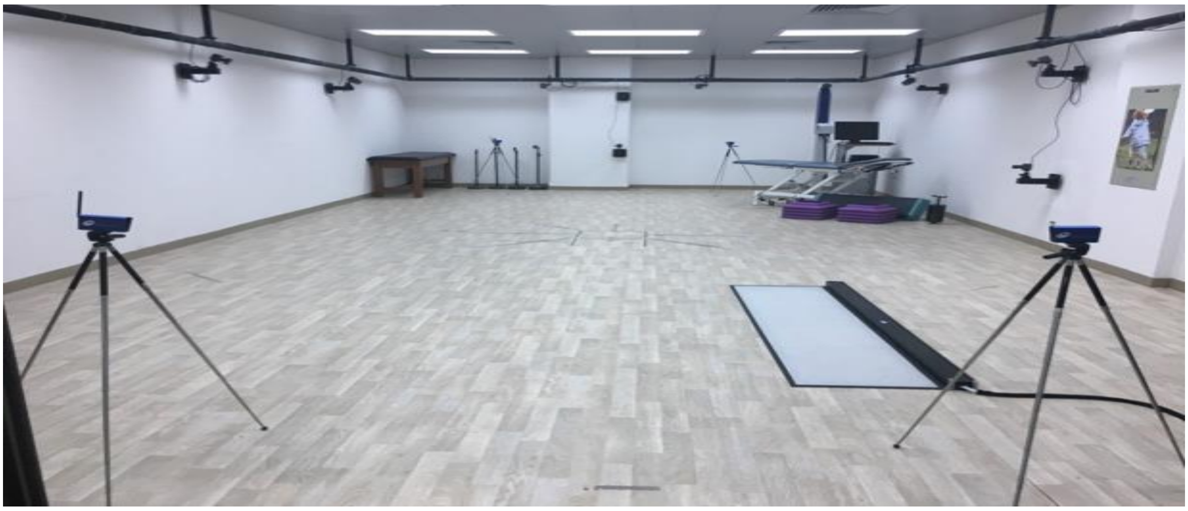


*Figure 4-2 3D motion analysis laboratory at Imam Abdulrahman Bin Faisal University.*

## 4.3 System Calibration

Correct calibration of the system is fundamental in ensuring the acquisition of meaningful kinematic/kinetic data. Calibration was therefore carried out to manufacturing standards, using two separate pieces of equipment to conduct this task. When applying the Qualisys system, two instruments were used for calibration purposes. The first object, depicted in Figure 4-3, is a metal reference frame shaped like an 'L' with four markers attached to the side. The frame should be placed on the force platform, corner down, parallel to the X and Y axes. Winter (2009) stated that the markers' distances from the origin of the force platform's coordinate system were pre-determined and therefore the software can derive the origin from the markers on the shape. The reference frame also allows the software to determine the lateral, posterior and vertical axes, named X, Y and Z respectively. In addition, it comprises a handheld device that takes the shape of a 'T', with two markers placed on it (Figure 4-3). Payton (2007) explained that the examiner should attempt to move the object randomly around the testing area while the first reference object is still in place. This allows the software to determine the position and orientation of the ten cameras within the coordinate system. The calibration of the testing environment took one minute. Upon completion, if the residual result of the cameras and the standard deviation of the length of the handheld reference object were not within 1mm of similarity, the calibration was repeated until the error margin is within 1mm.

*Figure 4-3 3D L-shaped frame and the T-shaped handheld wand for the Qualisys system.*

For the Vicon system, calibration was carried out using only one device known as an active wand (Figure 4-4). This instrument facilitates the attainment of unparalleled accuracy through the use of simultaneous multi-plane video calibration throughout the whole capture volume process. In addition, the active wand can be used to determine the global co-ordinate system, as well as the location and orientation in 3D space of the ten Vicon cameras. The calibration process comprised two key stages, the first of which involved performing a dynamic calibration. To carry out this process, an active wand device was waved throughout the area of the intended 3D capture volume for a period sufficiently long so as to ensure that every camera had an opportunity to clearly view the calibration object. Effective calibration requires that every camera image error must be equal to or less than 0.2mm, and this exercise must be repeated if a higher measurement error is recorded. The second stage in this calibration process involved defining the origin of the global co-ordinate system, whereby the active wand was positioned on the force platform, with its corner facing downward. The axes on the laboratory co-ordinate system were as follows: Y axis anterior-posterior (direction of travel); X axis medio-lateral; and Z axis superior-inferior.

*Figure 4-4 Active wand for the Vicon system.*

### 4.4 Procedure

Having signed their consent form, participants were then asked to select the standard trainers provided in their size and they were also given tight-fitting shorts. Moreover, information regarding their activity levels, age, mass and height measurements was recorded.

#### 4.4.1 Marker placement

Prior to commencement of the tasks, reflective markers (14 mm) were attached with self-adhesive tape to the participants' lower limbs at the following lanmarks: anterior superior spines. They were also attached to their posterior superior iliac spines, iliac crests, greater trochanters, medial and lateral femoral condyles, medial and lateral malleoli, posterior calcanei and the head of the first, second and fifth metatarsals, as shown in Figure 4-5. These markers are used to define the anatomical reference frame and centres of rotations of the joints. Figure 4-5 shows the tag set-up for the static trial.

A total of forty reflective markers were placed as static markers, as listed below:

- Anterior superior iliac spines (two markers)
- Posterior superior iliac spines (two markers)
- Iliac crest (two markers)
- Greater trochanters (two markers)
- Medial and lateral femoral condyles (four markers)
- Medial and lateral malleoli (four markers)
- Posterior calcanei (two markers)
- Four Clusters (16 markers)
- First, second and fifth metatarsals for both limbs (six markers)

The anatomical markers were later removed for data collection leaving the tracking markers in place, as shown in the Figure 4-6. The remaining 28 tracking markers consisted of 16 cluster markers, eight markers for standard shoes and four markers on the ASISs and PSISs. The marker placement was according to a protocol previously established and validated in projects undertaken at the University of Salford (Alenezi, Herrington, Jones & Jones, 2014). The motion capture system (Pro-Reflex, Qualisys, Sweden) was used to track the movement of these markers as the participant performs the tasks. With regard to speed, measurements were taken by using the Brower Timing Systems, where the total time to complete cutting and running tasks was calculated through the utilisation of a set of timing lights.

During movement assessments, anatomical significance and movement of each segment can be determined using the calibration anatomical systems technique (CAST) (Cappozzo et al., 1996). In the static standing trial, subjects were asked to stand on force plates to enable a camera view on all tracking and anatomical markers for the Qualisys software before commencement of the assessments. The anatomical markers created a reference point to recognise bone movement using the tracker markers during the action tests. In addition, Cappozzo, Della Croce, Leardini, and Chiari (2005) stated that as a way of pinpointing the 3D orientation and placement of each marker, two strategically positioned cameras should be in place so that both would capture the marker simultaneously. Furthermore, there should also be three non-collinear markers, visible from each segment, as a way of accurately reading its exact position.

Khan, Nokes, Jones, and Johnson (2007) recommended that in order to reduce artefacts on the skin, a set of rigid plastic cluster plates containing four markers should be located on each thigh and shank. Fabrifoam bandages from the USA were used to secure the shank and thigh clusters, which were tightly fastened to the anterior lateral aspect of each segment situated at mid-segment level. They were fastened with double tape to limit downward movement and rotation of the cluster pads.

*Figure 4-5 Static trial markers.*

*Figure 4-6 Tracking markers.*

### 4.4.2 Marker placement during the reliability study

In the reliability study, the same marker placement process was applied, as outlined in Section 4.4.1. In addition, some techniques were used to improve the marker placement process. As the reliability study required two visits, it was important to place the markers on the same bony landmarks during each visit. This would aid in achieving good reliability. During the data collection process, a tape measurement was used to ensure that the markers were placed in similar positions during both visits. For example, the distance between the anterior superior iliac spine and the marker on the medial malleolus was recorded for limb length. This was also done for every marker from the malleolus on the medial side and from the lateral malleolus on the lateral side during the first visit. This was re-checked during the second visit. Other sources of error during marker placements were considered, such as the calibration procedure, as explained in Section 4.3. In between the trials and tasks, the markers were monitored to confirm whether any movements had affected their positions. A pointed pen was used to establish the exact location of the markers.

### 4.4.3 Standardised shoes

Standardised footwear were provided for all participants in the form of New Balance 525 Running Shoes (Model: ss136668), which were available in different sizes and used for all the tasks undertaken. This approach was adopted to eliminate any potential footwear effects on the biomechanics, which may arise as a result of different shoe usage among participants.

### 4.5 Preparation and Tasks Undertaken

In advance of performing the side-cutting task, all participants were required to undertake a ten-minute general warm-up exercise, followed by a further five-minute warm-up on an exercise bike before beginning the test. The purpose of the warm-up exercises, according to Woods, Bishop, and Jones (2007), is to avoid the risk of further injury to participants. The participants' speeds were meticulously controlled through the use of photocell timing gates in the Brower Timing Systems, produced in Utah in the USA. The timing devices were placed 5m apart and 5m away from the force plates. The course incorporated strategically placed cones in order to ensure that any angles were set at 90 degrees, while also ensuring that all participants followed the same movement processes. To control the performance, the time required to complete the tasks (Run; 1.90 s ± 10%; cut90: 2.0 s ±10%) was measured.

#### 4.5.1 Overground Running

Participants were instructed to run for a distance of 10m. They were also given some practice to be aware of the task. In addition, the participants were advised to halt the process in the event of any difficulty occurring and to take some breaks. Furthermore, they were required to execute at least five successful trials for each movement on the overground running track in the 3D laboratory. A successful trial necessitates touching the force platform while running along a 10m runway (McLean et al., 2004). Every participant started with his dominant or non-dominant, allocated on the basis of a randomised process. Then do it again for the other leg. In order to minimise the effect of fatigue between trials, all participants were allowed to rest for 30 seconds. In addition to controlling the performance, the time required to complete the task (Run; 1.90 s ± 10%) was measured.

### 4.5.2 Single-leg squat (SLS)

Each participant was instructed to stand on one leg (dominant or non-dominant allocated on the basis of a randomised process) on the force platform. They were also asked to squat down as far as possible, to at least a 45-degree knee flexion, and they were given some practice time so as to eliminate any risk of falling. A standard goniometer (Gaiam110 Pro) was used to calculate the degree of knee flexion achieved throughout the practice trials, and the same tester was also involved in the observations carried out during this entire process. This ensured that a uniform and standardised approach was adopted for all participants taking part in the study, thus decreasing the influence of velocity on knee flexion. Each assessment was done over five seconds' duration, as measured using a counter. According to Herrington (2014), the first count initiates the squat, the third count shows the lowest point of the squat and the fifth count indicates the end of the trial. Furthermore, they were required to perform at least five successful trials for both legs. In order to minimise the effect of fatigue between trials, all participants were allowed to rest for 30 seconds. For the trials to be deemed valid, participants had to ensure that their squats complied with the minimum required knee flexion degree angle, while also retaining their balance for the entire duration of the process.

### 4.5.3 Single-Leg Landing (SLL)

The subjects were asked to drop from a 31cm step (Figure 4-7), lean forward and drop to as vertical a position as possible. Herrington and Munro (2010), recommended that participants take a unilateral stance on the contralateral limb and step forward to drop onto the force platform corresponding to the landing leg, thus ensuring that the contralateral leg makes no contact with any other surface. All participants were requested to start with either their dominant or non-dominant foot, allocated on the basis of a randomised process. They were subsequently invited to repeat this task using their other leg. Each participant was asked to perform at least five successful SLL trials for both legs. To minimise the effect of fatigue between trials, all participants were allowed to rest for 30 seconds.

*Figure 4-7 The step (31cm Height).*

### 4.5.4 Cutting to 90 degrees

All subjects were asked to run for 5m, plant their foot (dominant or non-dominant allocated on the basis of a randomised process), and change direction away from their foot at 90 degrees and continue running for a further 3m. They were subsequently asked to repeat this task for the other foot. In addition, all participants took part in at least five successful cutting trials for both legs. A successful trial requires that the contact phase occurs (while cutting) on the force plate. To minimise the effect of fatigue between trials, all participants were allowed to rest for 30 seconds. It was crucially important to ensure that a consistent approach was adopted, whereby every participant performed the cutting task at a 90-degree angle. Therefore, cones were used to guide the participant to perform this activity at the correct angle. In addition to controlling the performance, the time required to complete the tasks (Cut90: 2.0 s ±10%) was calculated.

## 4.6 Randomization

All participants started the four tasks and the two legs in a random order using randomisation blocks via randomization.com.

## 4.7 Data Processing

Qualisys Track Manager software (Version 1.10.282; Qualisys) and Vicon Nexus software (Version 2.5.0.93422h) were used to reconstruct the 3D marker trajectories. Visual3D motion capture software (Version 6.00.16 C-Motion Inc., Rockville, MD, USA) was used to import C3D data files from Qualisys and Nexus software to analyse and evaluate the kinetic and kinematics data. Butterworth $4^{th}$ order bi-directional low-pass filter was utilised to filter motion and the force plate with cut-off of frequencies 12 Hz and 25 Hz for kinematic and force data, respectively. Yu, Gabriel, Noble, and An (1999) developed a procedure to determine the cut-off frequency for the Butterworth low-pass digital filter. This became the basis of the cut-off frequency and was selected in this thesis. The main goal of data filtration is to reduce random noise by smoothing the data with no effect on the signal. Joint kinematics were evaluated using an X-Y-Z Euler rotation sequence (X = flexion-extension, Y = abduction-adduction or varus-valgus and Z = internal-external rotation) (Figure 4-8). Joint kinetic data were calculated using a 3D inverse dynamic; the joint moment data were normalised to body mass and presented as an external moment.

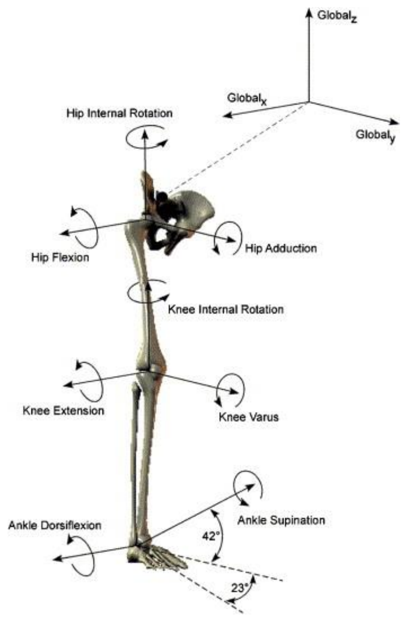



Figure 4-8 Lower extremity segment and joint rotation denotations

(McLean et al., 2005).

The CAST model, referenced by Cappozzo, Catani, Della Croce, and Leardini (1995) was used. Both anatomical and tracking markers (Section 4.4.1) were used in the static trial, which was processed using Qualisys and Nexus before being extracted to Visual3D, a post-processing software. Prior to extraction for post-processing software, a static trial was required in order to provide a reference point for the software during the trial. During the static trial, each participant stood on the force plates with each of the markers in full view of the cameras. The anatomical and tracking markers were then recorded and added to the database for the Qualisys software application to utilise later. The placement of these anatomical markers provided a useful reference position for the software to be able to recognise bone motion by tracking the movements of these specific markers, which had been set during previous movement trials.

Figure 4-9 shows that the model utilised seven rigid segments attached to the joint. Each segment contained six variables used to define its location, three are used to define the position of the origin and three are for the rotation in 3D. To be more precise, three variables are used to define the segment translation based upon the three perpendicular axes, the vertical, the medial-lateral and anterior-posterior. Three further variables are used to explain the rotation movement taking place about each axis of the segment, these being the sagittal, frontal and transverse. Each participant's body mass readings were recorded in kilogrammes, along with their height in metres and then saved into the software package ready for usage in the kinetic calculations. In addition, each segment of the pelvis, thigh, shank and foot was modelled to detect the proximal and distal joint/radius. Table 4-1 below outlines details of each segment, as well as the markers used to identify that specific segment. According to Bell, Brand, and Pedersen (1989), the calculation of the hip-joint centre occurs automatically through the use of the ASIS and PSIS markers, using a regression equation.

*Table:4-1 Lower limb segments and their markers.*

| Segment | Proximal markers | Distal markers | Tracking markers |
|---|---|---|---|
| **Pelvic** | Right and left anterior superior iliac spine | Right and left posterior superior iliac spine markers | Right and left anterior superior iliac spine + Right and left posterior superior iliac spine |
| **Thigh** | Hip joint center | Medial and lateral condyle | Cluster on the thigh (4 markers) |
| **Shank** | Medial and lateral condyle | Medial and lateral malleolus | Cluster on the shank (4 markers) |
| **Foot** | Medial and lateral malleolus | 1$^{st}$ and 5$^{th}$ metatarsal head | Heal marker, 1$^{st}$ metatarsal head marker, 2$^{nd}$ metatarsal head marker and 5$^{th}$ metatarsal head marker |

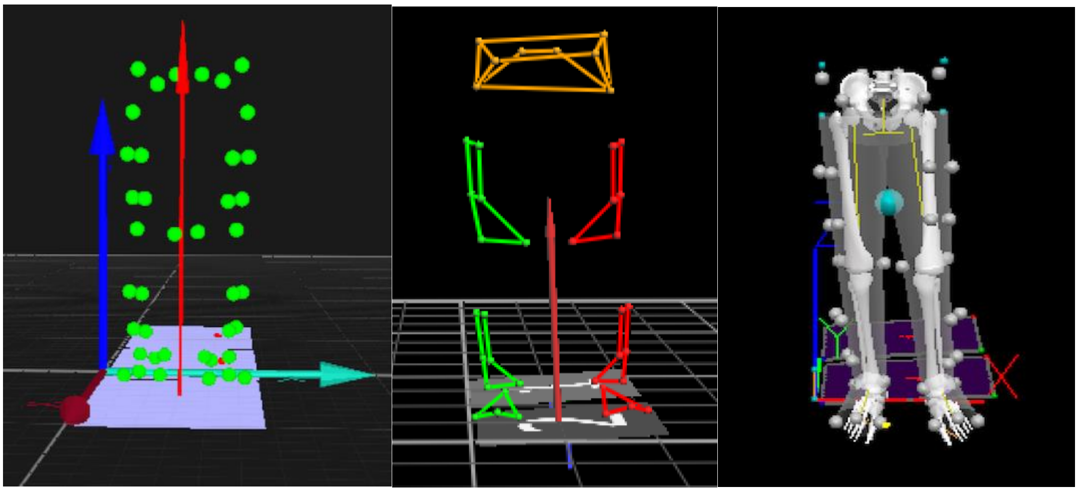


*Figure 4-9 Static models in QTM™ (left), static models in Nexus (middle)and bone model in Visual 3D™ (right).*

In order to form the correct basis for the running and cutting tasks, kinematics and kinetic data were normalised to 101 data points, representative of between 0–100 percent of the stance phase (Kuenze et al., 2014) and 100 percent of the leg contact phase. This definition was deemed to apply from the initial contact of the leg (IC) to toe-off (TO). The initial contact was taken from when the vertical GRF first exceeded 10 newtons (N). The official TO was declared when the VGRF fell under 10 N.

Throughout the SLS task, start time readings were taken as soon as the knee exceeded to a 15-degree flexion, and this was subsequently ended when returning to this point during ascent at the end of the task. Kinematics and kinetic data were normalised to 101 data points, which was representative of between 0–100 percent of the defined phase. To ensure accurate recording of the SLL task, the exercise was defined as commencing from the IC until the knee displayed a 15-degree ascending flexion; this angle was selected so as to ensure that the maximum knee flexion was included in the SLL cycle. Kinematics and kinetic data were normalised to 101 data points, which was representative of between 0–100 percent of the defined phase.

## 4.8 Movement Variability measurement

In this thesis, the within-subject variability was calculated using the coefficient of variation CV (second order) method introduced by Kvålseth (2017), based on the following formula:

$$CV = \left(\frac{SD^2}{SD^2 + M^2}\right)^{1/2}$$

SD indicates to the standard deviation for the peaks in the five successful trials. M indicates to the mean of peaks in the five successful trials. In fact, this CV is the most appropriate version to use for biomechanical data as it overcomes the significant limitations associated with the previous version. Other examples of its advantages over the previous version include: surmounting the challenges encountered in relation to positive and negative datasets; providing more accurate interpretation of coefficient of variation values due to the absence of upper bound constraints; facilitating more insightful and intuitive interpretations; being less sensitive to outliers; and being less likely to be adversely impacted by mean scores and errors or fluctuations in mean values (Kvålseth, 2017). A zero value signals that no variability has occurred while, conversely, a score of one is indicative of the highest levels of variability.

## 4.9 Main outcome measures

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

- The maximum knee joint angles (frontal and sagittal) during the first 60% of the defined phase, following IC.
- The maximum knee joint moments (valgus and extension moment) during the first 60% of the defined phase, following IC.
- The MV for the maximum knee joint angles (frontal and sagittal).
- The MV for the maximum knee joint moments (valgus and extension).

- The maximum GRF.
- The MV for GRF.

These discrete kinematic and kinetic variables were selected because of their association with ACL injury risk (Hewett et al., 2005; Myer et al., 2015; Numata et al., 2018; Hewett, Torg, & Boden, 2009; Cochrane et al., 2007; Ebstrup & Bojsen-Møller, 2000; Kobayashi et al., 2010; Olsen et al., 2004; Markolf et al., 1995; Berns, Hull, & Patterson, 1992; McLean, Lipfert, & Van Den Bogert, 2004; Leppänen et al., 2017) as explained in Section 2.7.2.

### 4.10 Relation between the two laboratories used in the thesis

For the purposes of this research, while two different laboratories, namely the Human Performance Laboratory (HPL) at Salford University and the Imam Abdulrahman Bin Faisal University laboratory (IABFUL) were selected to gather relevant data, it was anticipated that the kinetics and kinematics produced in both sites would be similar in nature. However, to verify this, two participants were invited to visit both aforementioned laboratories, where the same research investigator placed all the markers in position. The participants were then requested to carry out five successful trials on the four identified sporting tasks. The primary aim of this exercise was to ascertain if any differences emerged in relation to the key outcome measures (namely knee valgus angle and moment, knee flexion angle and moment and VGRF) between either laboratory. The findings are presented in Tables 4-2, 4-3, 4-4 and 4-5, which show that no differences were found to occur in terms of outcome measurement scores. This confirms that the kinetics and kinematics scores produced in both laboratories were almost similar for the majority of the variables in the four tasks undertaken, irrespective of the type of system used or its positioning.

*Table:4-2 Lower limb kinematics and kinetics for the first participant during running and cutting 90° in the two laboratories.*

| Variable and task | Biomechanical value (Mean) | | CV of the biomechanical variable | |
|---|---|---|---|---|
| | HPL | IABFUL | HPL | IABFUL |
| Running | | | | |
| Hip flexion (°) | 50 | 47 | 0.03 | .05 |
| Hip abduction (°) | -7 | -5 | 0.22 | 0.27 |
| Knee flexion (°) | 50 | 49 | 0.06 | 0.04 |
| Knee extension moment (Nm/Kg) | 2.24 | 2.21 | 0.07 | 0.08 |
| Knee valgus (°) | -6 | -5.5 | 0.12 | 0.17 |
| Knee valgus moment (Nm/Kg) | 0.26 | 0.31 | 0.20 | 0.32 |
| Ankle dorsiflexion (°) | 32 | 31 | 0.05 | 0.05 |
| VGRF (*BW) | 2 | 2 | 0.02 | 0.02 |
| Cutting 90 | | | | |
| Hip flexion (°) | 53 | 52 | .04 | 0.04 |
| Hip abduction (°) | -21 | -19 | 0.08 | 0.10 |
| Knee flexion (°) | 64 | 65 | 0.06 | 0.05 |
| Knee extension moment (Nm/Kg) | 1.95 | 1.92 | 0.11 | 0.10 |
| Knee valgus (°) | -4 | -6 | 0.40 | 0.28 |
| Knee valgus moment (Nm/Kg) | 0.40 | 0.51 | 0.31 | 0.23 |
| Ankle dorsiflexion (°) | 26 | 28 | 0.06 | 0.13 |
| VGRF (*BW) | 1.3 | 1.4 | 0.03 | 0.04 |

*Table:4-3 Lower limb kinematics and kinetics for the first participant during SLL and SLS in the two laboratories.*

| Variable and task | Biomechanical value (Mean) | | CV for the biomechanical variable | |
|---|---|---|---|---|
| | HPL | IABFUL | HPL | IABFUL |
| SLL | | | | |
| Hip flexion (°) | 61 | 59 | 0.08 | 0.10 |
| Hip abduction (°) | -14 | -16 | 0.04 | 0.15 |
| Knee flexion (°) | 68 | 65 | 0.10 | 0.10 |
| Knee extension moment (Nm/Kg) | 2.76 | 2.74 | 0.12 | 0.11 |
| Knee valgus (°) | -1 | -2 | 0.51 | 0.58 |
| Knee valgus moment (Nm/Kg) | 0.19 | 0.13 | 0.57 | 0.46 |
| Ankle dorsiflexion (°) | 25 | 27 | 0.10 | 0.08 |
| VGRF (*BW) | 2.2 | 2.3 | 0.04 | 0.06 |
| SLS | | | | |
| Hip flexion (°) | 68 | 72 | 0.03 | 0.03 |
| Hip abduction (°) | -19 | -16 | 0.18 | 0.13 |
| Knee flexion (°) | 82 | 85 | 0.08 | 0.03 |
| Knee extension moment (Nm/Kg) | 1.10 | 1.08 | 0.13 | 0.12 |
| Knee valgus (°) | -1.6 | -1.4 | 0.36 | 0.45 |
| Knee valgus moment (Nm/Kg) | 0.09 | 0.05 | 0.51 | .63 |
| Ankle dorsiflexion (°) | 34 | 35 | 0.08 | 0.02 |
| VGRF (*BW) | 1.1 | 1.1 | 0.04 | 0.03 |

*Table:4-4 Lower limb kinematics and kinetics for the second participant during running and cutting 90$^o$ in the two laboratories.*

| Variable and task | Biomechanical value (Mean) | | CV of the biomechanical variable | |
|---|---|---|---|---|
| | HPL | IABFUL | HPL | IABFUL |
| Running | | | | |
| Hip flexion (°) | 52 | 51 | 0.03 | .06 |
| Hip abduction (°) | -5.5 | -7 | 0.22 | 0.13 |
| Knee flexion (°) | 48 | 51 | 0.02 | 0.02 |
| Knee extension moment (Nm/Kg) | 2.93 | 3.01 | 0.09 | 0.07 |
| Knee valgus (°) | -2.6 | -3 | 0.24 | 0.19 |
| Knee valgus moment (Nm/Kg) | 0.13 | 0.14 | 0.28 | .29 |
| Ankle dorsiflexion (°) | 30 | 33 | 0.03 | 0.05 |
| VGRF (*BW) | 2.2 | 2.1 | 0.01 | 0.02 |
| Cutting 90 | | | | |
| Hip flexion (°) | 73 | 77 | 0.03 | 0.03 |
| Hip abduction (°) | -27 | -33 | 0.08 | 0.13 |
| Knee flexion (°) | 78 | 77 | 0.05 | 0.03 |
| Knee extension moment (Nm/Kg) | 2.14 | 2.11 | 0.12 | 0.14 |
| Knee valgus (°) | -3.5 | -4.5 | 0.62 | 0.60 |
| Knee valgus moment (Nm/Kg) | 1.03 | 1.0 | 0.12 | 0.20 |
| Ankle dorsiflexion (°) | 21 | 19 | 0.11 | 0.24 |
| VGRF (*BW) | 2.0 | 2.1 | 0.03 | 0.01 |

*Table:4-5 Lower limb kinematics and kinetics for the second participant during SLL and SLS in the two laboratories.*

| Variable and task | Biomechanical value (Mean) | | CV for the biomechanical variable | |
|---|---|---|---|---|
| | HPL | IABFUL | HPL | IABFUL |
| SLL | | | | |
| Hip flexion (°) | 73 | 67 | 0.08 | 0.10 |
| Hip abduction (°) | -14 | -19 | 0.23 | 0.15 |
| Knee flexion (°) | 87 | 83 | 0.05 | 0.02 |
| Knee extension moment (Nm/Kg) | 3.16 | 3.12 | 0.05 | 0.04 |
| Knee valgus (°) | 3.6 | 3.8 | 0.34 | 0.23 |
| Knee valgus moment (Nm/Kg) | 0.23 | 0.36 | 0.31 | 0.53 |
| Ankle dorsiflexion (°) | 32 | 34 | 0.04 | 0.06 |
| VGRF (*BW) | 2.3 | 2.2 | 0.07 | 0.02 |
| SLS | | | | |
| Hip flexion (°) | 90 | 87 | 0.02 | 0.04 |
| Hip abduction (°) | -3 | -2.6 | 0.32 | 0.21 |
| Knee flexion (°) | 82 | 86 | 0.06 | 0.02 |
| Knee extension moment (Nm/Kg) | 0.93 | 0.96 | 0.19 | 0.21 |
| Knee valgus (°) | 2.5 | 3 | 0.21 | 0.10 |
| Knee valgus moment (Nm/Kg) | 0.02 | 0.03 | 0.52 | .51 |
| Ankle dorsiflexion (°) | 36 | 39 | 0.02 | 0.03 |
| VGRF (*BW) | 1.1 | 1.1 | 0.02 | 0.02 |

### 4.11 Statistical analysis

All statistical evaluations in this thesis were undertaken using the Statistical Package for the Social Sciences (SPSS) (Version 24). Descriptive analysis (means and standard deviations) were provided. Prior to data analysis, the normality distribution of data was observed visually following application of a Shapiro-Wilk test. This test was preferred to other available tests, such as the Kolmogorov-Smirnov (K-S) test, because of previously reported high sensitivity to extreme values (Barton & Peat, 2014). Researchers including Field (2017), Ghasemi and Zahediasl (2012), Razali and Wah (2011), Steinskog et al. (2007) and Thode (2002) have all argued that the Shapiro-Wilk test is the most powerful tool to apply when assessing and testing normality.

The assessment of normality distribution is based upon key assumptions which help to decide if parametric statistical tests are applicable to the particular tests being undertaken. Therefore, we can be certain, through the use of these rigorous checking processes, that the most appropriate statistical tests have been applied. If the data is normally distributed then parametric testing procedures are used, whereas if non-parametric then the most appropriate tests are applied on an individual basis. Each chapter focuses on a specific statistical test, depending on its aims and the hypothesis put forward, which will be explained in further detail at a later stage.

## 4.12 Reliability of lower limb biomechanical outcome measures and CV as an outcome measure of variability among healthy subjects using 3D motion analysis during four specific sporting tasks: SLS, SLL, running and cutting 90.

### 4.12.1 Background

Three-dimensional motion analysis (3DMA) is recognised as an important investigative method used in both research and clinical environments to evaluate and legitimise decision-making and functional diagnoses. Another highly useful measurement, namely repeated functional tasks, can be used to assess the outcome of therapeutic interventions. Variability is, however, observed between measurements taken before and after the intervention, which may be caused by effects from treatment and/or measurement variation (McGinley et al., 2009) . Therefore, the risk of analysing insignificant variances as significant (Schwartz, Trost, & Wervey, 2004) could be minimised by gaining a better understanding of the magnitude of error. This knowledge could also assess the likelihood of a measured intervention effect being significant beyond the margin of error. Variability affecting the testing procedure's error magnitude arises from many possible sources when using 3DMA, for example: instrumental errors, misplacement of anatomical landmarks and STAs or soft tissue artefacts (Cereatti, Della Croce, & Cappozzo, 2006)

It is of paramount importance to understand reliability and minimal detectable change (MDC) values drawn from a healthy population; clinicians and researchers can utilise this knowledge to interpret pathological data. Measurement fluctuations can arise as a result of both intrinsic and extrinsic factors, an example being technical errors (Schwartz et al., 2004). Extrinsic errors can affect the reliability of data procured from one assessor's multiple, resulting in variations in testing sessions, known as inter-session or within-assessor, as well as inter-assessor tests, which describe various tests being conducted by different assessors. One primary reason for this problem is uneven marker placement; however, several other factors could also contribute to data variation. These may include: performance speed, data processing and measurement equipment errors (Monaghan, Delahunt, & Caulfield, 2007). As previously mentioned, Batterham and George (2000) highlighted 'the importance of comprehending the reliability and measurement faults within the screening tools'. One of the most important aspects of 3D motion analysis is to ascertain meaningful and reliable kinematic and kinetic variable data over multiple days. This between-days reliability, as shown by Ferber et al. (2002) and Queen et al. (2006), is most affected by marker-placement error.

Several studies have reported higher reliability rates for sagittal-plane variables as opposed to frontal and transverse variables: Ferber et al. (2002) and Queen et al. (2006), recorded this during running; Alenezi et al. (2014) and Malfait et al. (2014) as well as Ford et al. (2007) all observed this during observation of vertical jumping; finally, Alenezi et al. (2014) and

Nakagawa, Moriya, Maciel, and Serrão (2014) studied SLS. Frontal and transverse movement, notably dynamic-knee valgus, as reported by Hewett et al. (2005) and Myer et al. (2010), is understood to play a key role in dangerous motions that are linked to ACL. As a result, 3D movement analysis methods may be severely hindered in these planes of movement by these measurement errors, possibly leading to misidentification of high-risk injury individuals. With relation to cutting tasks, the reliability or changes of biomechanical factors during these tasks has been investigated in a number of studies, notably Alenezi et al. (2016), Besier et al. (2001), Sankey et al. (2015) and Stephenson et al. (2012).

Within the area of sport biomechanics, MV has been widely investigated. It can be measured using different methods, as discussed in the second chapter. One of the most common methods applied is CV. However, despite it being one of the most frequently used measures of variability, it does have some limitations. For example, problems emerge when the data are both positive and negative and it is also highly sensitive to outliers. Therefore, second order CV has recently been introduced by Kvålseth (2017) in order to address the limitations associated with the previous version. That implies that the newer version is more suitable to use in the case of biomechanical data. However, its reliability has not been tested to date. Therefore, one of the aims of this study is to test the reliability of CV (second order) as a measure of MV.

### 4.12.2 Study aims

1. To examine the between-day reliability of lower extremity biomechanics using a 3D motion analysis during functional tasks.

2. To establish the standard error of measurement (SEM) values for 3D biomechanical measurements during functional tasks.

3. To examine the reliability of CV (second order) as an outcome measure of variability.

### 4.12.3 Study null hypothesis:

1- $H0_1$: All lower extremity kinematics and kinetics are unreliable between days in all four tasks.

2- $H0_2$: There are no differences in reliability between different functional tasks.

3- $H0_3$: The reliability of the GRF measurement is no greater than the reliability of the other kinematic and kinetic data in all tasks.

4- $H0_4$: CV is not a reliable measurement of the variability of biomechanical variables in the lower limbs.

### 4.12.4 Methods

#### 4.12.4.1 Participants

Drawing from the University of Salford's staff and student population, ten fit and moderately active male adults were selected and invited to become volunteer participants, as described in Table 4-6. The 3D motion analysis system available in the human performance laboratory (Section 4.2.1) at Salford University was applied in this study.

*Table:4-6 Participants' demographic profiles*

| | Number (Gender) | Mean | SD |
|---|---|---|---|
| Age (years) | 10 (male) | 27 | 4.4 |
| Height (m) | 10 (male) | 1.7 | 0.1 |
| Body mass (kg) | 10 (male) | 66 | 7.2 |

#### 4.12.4.2 Inclusion and exclusion criteria

Healthy subjects were selected on the basis of being within the 18–40 years age group, with good levels of fitness and health and displaying moderate levels of physical activity, which has been defined as performing any sport or exercise for at least half an hour, three times a week, for the previous six months (Munro & Herrington, 2011). Subjects should be able to maintain normal balance, which was described by Bohannon, Larkin, Cook, Gear, and Singer in 1984 as the ability to stand on one leg for 30 seconds with closed eyes. Additionally, the subjects should not have any record of injury to the lower extremities, pelvis or back in the previous year, which includes surgical operations, as well as being able to perform any given tasks without assistance. Within this context, an injury is regarded as any musculoskeletal discomfort or complaint that may inhibit the subject's ability to perform exercise. Individuals were excluded who were experiencing any kind of pathology or minor pain in their lower limbs which could affect the outcome of testing, as well as those who have shown a tendency towards ACL injuries and those who did not sign the consent form.

#### 4.12.4.3 Tasks undertaken

Each participant completed all four functional tasks: single-leg squat (SLS), single-leg landing (SLL), running and cutting to 90 degrees as described in this chapter (Section 4.5).

### 4.12.5 Data collection

All participants completed two separate sessions, held one week apart. In each session, the subjects completed five successful trials, and the mean of the peaks of the five trials was used to calculate the between-day reliability values. The procedure outlined in this chapter (Section 4.4) was subsequently applied.

### 4.12.6 Number of trials and outlier management

A total of eight trials were collected for each participant in relation to each task. In order to determine the number of trials utilised in the thesis, the average of trials 1–2, 1–3, 1–4 and 1–5 was calculated during the reliability study (n=10) for all four tasks. This was done to allow for the visual inspection of the mean between the different sums of the trials that helped identify when the mean can demonstrate consistent values. A preliminarily inspection of the reliability study data demonstrated that the five trials were suitable and consistent for all four tasks, as shown in: Tables 4–7 for the cutting to 90 degrees task, Tables 4–8 for the running task, Tables 4–9 for the SLL task and Tables 4–10 for the SLS task. Further, the first five trials for each participant were selected. The rationale for collecting eight trials is that it makes provisions for additional trials if the first five trials produced an outlier, incomplete trials occurred, technical errors arose or a trial was not considered to be successful. In addition, a visualisation method was selected to detect the presence of outliers. If a trial produced a significantly different trace or poorer tracking than all of the other trials, it was considered to be an outlier. When this occurred, it was subsequently removed and replaced by trial 6.

*Table 4-7: Average peak values for angles, moments and GRF based on the sum of different trials during the cutting to 90 degrees task (n=10)*

| Variable | Trial 1–2 | Trial 1–3 | Trial 1–4 | Trial 1–5 |
|---|---|---|---|---|
| **Knee flexion angle (°)** | 69.93 | 69.56 | 69.54 | 69.65 |
| **Knee extension moment (Nm/Kg)** | 3.39 | 3.41 | 3.40 | 3.39 |
| **Knee valgus angle (°)** | -7.92 | -7.68 | -7.63 | -7.66 |
| **Knee valgus moment (Nm/Kg)** | 0.73 | 0.74 | 0.74 | 0.73 |
| **GRF (*Body weight)** | 2.17 | 2.19 | 2.18 | 2.18 |

*Table 4-8: Average peak values for angles, moments and GRF based on the sum of different trials during the running task (n=10)*

| Variable | Trial 1–2 | Trial 1–3 | Trial 1–4 | Trial 1–5 |
|---|---|---|---|---|
| **Knee flexion angle (°)** | 50.76 | 50.85 | 50.80 | 50.87 |
| **Knee extension moment (Nm/Kg)** | 3.45 | 3.47 | 3.46 | 3.46 |
| **Knee valgus angle (°)** | -2.03 | -1.94 | -1.98 | -1.90 |
| **Knee valgus moment (Nm/Kg)** | 0.23 | 0.23 | 0.23 | 0.23 |
| **GRF (*Body weight)** | 2.56 | 2.56 | 2.55 | 2.55 |

*Table4-9: Average peak values for angles, moments and GRF based on the sum of different trials during the SLL task (n=10)*

| Variable | Trial 1–2 | Trial 1–3 | Trial 1–4 | Trial 1–5 |
|---|---|---|---|---|
| **Knee flexion angle (°)** | 73.99 | 73.76 | 73.34 | 73.23 |
| **Knee extension moment (Nm/Kg)** | 3.56 | 3.57 | 3.58 | 3.58 |
| **Knee valgus angle (°)** | -1.93 | -2.01 | -2.16 | -2.22 |
| **Knee valgus moment (Nm/Kg)** | 0.37 | 0.40 | 0.38 | 0.38 |
| **GRF (*Body weight)** | 3.26 | 3.23 | 3.24 | 3.28 |

*Table 4-10: Average peak values for angles, moments and GRF based on the sum of different trials during the SLS task (n=10)*

| Variable | Trial 1–2 | Trial 1–3 | Trial 1–4 | Trial 1–5 |
|---|---|---|---|---|
| **Knee flexion angle (°)** | 88.14 | 88.68 | 88.53 | 88.63 |
| **Knee extension moment (Nm/Kg)** | 2.08 | 2.11 | 2.11 | 2.10 |
| **Knee valgus angle (°)** | 0.82 | 0.83 | 0.81 | 0.87 |
| **Knee valgus moment (Nm/Kg)** | 0.08 | 0.07 | 0.08 | 0.08 |
| **GRF (*Body weight)** | 1.12 | 1.12 | 1.12 | 1.12 |

#### 4.12.7 Statistical analysis

All statistical evaluations were undertaken using the Statistical Package for the Social Sciences (SPSS) for Mac (Version 20). According to Batterham and George (2000), descriptive analysis (means and standard deviations) are used for statistical evaluations. In order to show a control, the mean of successful five trials from the first visit and second visit were taken to assess between-days reliability. This ensured that comparisons could be made over time to give a true picture and eliminate external factors from the experiments on different days. Additionally, the Intra-Class Correlation Coefficients (ICCs) and SEM were used to determine intra-rater reliability and the level of agreement. The interpretation of ICC values was as follows: poor <.40; fair .40 to 70; good 70 to 90; and excellent >.90 (Coppieters, Stappaerts, Janssens, & Jull, 2002). The ICC was selected as a specific measurement indicator due to its stringent and uniform criteria, along with its coefficient reliability in relation to all other elements considered to be in the same classification or category. Yaffee (1998) explained this in greater detail by stating that the 'ICC compares the covariance of the scores' against the total variance. The selected ICC (two-way mixed model) was chosen having taken the recommended guidelines produced by Shrout and Fleiss (1979) into consideration.

ICC may seem to be easy to derive from, although used on its own it cannot depict reliability completely as it does not provide any sort of error margin between two measurements. It is therefore also suggested to use the SEM, which is described by Rankin and Stokes in 1998 as 'an important tool that will provide the error interval between two measurements'. Munro et al. in a 2012 publication stated that, for 'practitioners who require a way to discern individual improvements, calculation of the SEM is incredibly valuable'. The SEM provides a value for absolute reliability, with a lower value being more reliable. Baumgartner (1989) stated that this allows the researcher to ascertain an approximation of the real change and also provide an error interval. As recommended by Denegar and Ball (1993), the formula

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

was used to calculate the SEM. Calculation of the SEM can help greatly in discerning the actual change in outcomes, instead of a measurement error. Having a high ICC and a low SEM is considered reliable.

The smallest detectable difference (SDD) was estimated in order to determine the practical measurement error (Portney & Watkins, 2009). SDD play a key role in evaluating previous research, as well as in studies planned for the future, and, in particular, those focusing on intervention strategies. It would be difficult to attribute such changes with certainty to treatment alone, without having also examined the measurement error, so as to eliminate the possibility of other factors impacting upon the results, for example, static alignment or marker placements (Ford et al., 2007; Malfait et al., 2014; Whatman et al., 2011). SDD were estimated using the following formula: $SDD = 1.96 * SEM * \sqrt{2}$ (Kropmans, Dijkstra, Stegenga, Stewart, & De Bont, 1999). In line with other studies, for example Blankevoort, Van Heuvelen, & Scherder, (2013) and Bruton, Conway, and Holgate (2000) the same unit were used to express SDD and SEM, namely the Newton-metre per kilogram for joint moment and degree for the joint angle.

### 4.12.8 Results

#### 4.12.8.1 Reliability of lower limb kinematics and kinetics

Generally, in all tasks, all the variables' ICC values ranged between 0.47 and 0.99, thus reporting fair to excellent reliability. The ICC values for lower limb kinematics and kinetics are presented in Table 4-11 for running, Table 4-13 for cutting 90, Table 4-15 for SLL and Table 4-17 for SLS. Most variables (38 out of 60) reported good reliability (ICC = 0.70 to 0.90). A total of 17 variables (out of 60) reported excellent reliability (ICC > 0.90). However, there were five variables reported fair reliability (ICC = 40 to 70). The highest ICC values were recorded in the sagittal plane, producing an ICC of 0.99 at the hip flexion angle and ankle plantar flexion moment, both of which occurred during SLL. In contrast, the lowest ICC value was reported in the transverse plane at the hip internal rotation angle during SLL. Furthermore, the between-day SEM values for all lower limb biomechanical variables during all tasks are presented in Table 4-11 for running, Table 4-13 for cutting 90, Table 4-15 for SLL and Table 4-17 for SLS. In general, the results show that the SEM values for all kinematic variables (angles) ranged between 0.97 and 3.79 degrees. The highest SEM (3.79°) was found to occur at the hip adduction during the cutting to 90 degrees task. However, the lowest SEM (0.97°) was reported in the knee valgus angle during a SLS.

In terms of kinetic variables, the SEM values of all of them ranged between 0.01 and 0.31 Nm/Kg. Meanwhile, the hip flexion moment during cutting 90 was the highest (0.31 Nm/Kg), but the knee valgus moment during SLS was the lowest (0.01 Nm/Kg).

#### 4.12.8.2 Reliability of CV in lower limb kinematics and kinetics

In general, in all tasks, the between-day ICC values of CV for all lower limb kinematic and kinetic variables ranged between 0.50 and 0.99, reporting fair to excellent reliability ratings. All ICC values for all CVs are presented in Table 4-12 for running, Table 4-14 for cutting 90, Table 4-16 for SLL and Table 4-18 for SLS. Most variables (29 out of 60) reported good reliability (ICC = 0.70 to 0.90). A total of seven variables (out of 60) reported excellent reliability scores (ICC > 0.90). However, 23 variables reported fair reliability levels (ICC = 40 to 70). The highest ICC values for all CVs were first seen in the sagittal plane, with a reported ICC of 0.99 at the ankle plantar flexion moment. Secondly, the VGRF reported the same ICC scores for both, during the running task. However, the lowest ICC values first emerged in the frontal plane at the hip adduction angle (0.50) during SLSs and, secondly, in the transverse plane at the knee internal rotation angle during SLLs (0.51).

In addition, the SEM values for CV of all kinematic and kinetic variables are presented in Table 4-12 for running, Table 4-14 for cutting 90, Table 4-16 for SLL and Table 4-18 for SLS. In general, the results show that the SEM values for all CVs ranged between 0.005 and 0.17. The highest SEM score (0.17) was seen first in the transverse plane at the knee internal rotation angle during SLS, and, secondly, in the frontal plane both at the hip and ankle adduction during cutting 90 tasks. However, the lowest SEM score (0.005) was reported in VGRF during running. In terms of both angles and moments, the lowest SEM value (0.01) was recorded in the sagittal plane at the hip flexion angle (during running and SLS), knee flexion angle (during running, cutting 90 and SLS), dorsiflexion angle (during running and SLS) and plantar flexion moment (during running and cutting 90).

The SDD values are also presented in Table 4-11 for running, Table 4-13 for cutting 90, Table 4-15 for SLL and Table 4-17 for SLS for lower limb kinematic and kinetic variables, while Table 4-12 presents SDD values for running, Table 4-14 for cutting 90, Table 4-16 for SLL and Table 4-18 for SLS for CV of lower limb kinematic and kinetic variables.

*Table:4-11 Between-day ICC (95% CI), mean, SD, SEM and SDD for lower limb kinematics and kinetics during running.*

| Variable | ICC (95% CI) | Mean $_{pooled}$ | SD $_{pooled}$ | SEM | SDD |
|---|---|---|---|---|---|
| Angles (°) | | | | | |
| Hip flexion | 0.85 (0.36-0.96) | 50.05 | 7.02 | 2.76 | 7.66 |
| Hip adduction | 0.84 (0.32-0.96) | 10.43 | 4.00 | 1.61 | 4.46 |
| Hip internal rotation | 0.72 (-0.08-0.93) | 7.26 | 5.48 | 2.92 | 8.08 |
| Knee flexion | 0.94 (0.75-0.99) | 51.98 | 5.13 | 1.27 | 3.51 |
| Knee valgus | 0.80 (0.25-0.95) | -1.60 | 3.35 | 1.50 | 4.17 |
| Knee internal rotation | 0.84 (0.33-0.96) | 3.78 | 4.48 | 1.79 | 4.97 |
| Ankle dorsiflexion | 0.83 (0.35-0.96) | 32.63 | 2.55 | 1.05 | 2.92 |
| Ankle adduction | 0.66 (-0.27-0.91) | 4.52 | 3.09 | 1.81 | 5.02 |
| Moments (Nm/Kg) | | | | | |
| Hip flexion | 0.86 (0.43-0.97) | -2.09 | 0.79 | 0.29 | 0.81 |
| Hip adduction | 0.85 (0.49-0.96) | -2.05 | 0.37 | 0.14 | 0.40 |
| Hip internal rotation | 0.75 (0.01-0.94) | -0.81 | 0.15 | 0.07 | 0.21 |
| Knee extension | 0.75 (-0.08-0.94) | 3.49 | 0.60 | 0.30 | 0.83 |
| Knee valgus | 0.81 (0.30-0.95) | 0.12 | 0.08 | 0.03 | 0.10 |
| Ankle plantar flexion | 0.82 (0.32-0.95) | -2.71 | 0.43 | 0.19 | 0.52 |
| GRF (*Body Weight) | | | | | |
| Vertical GRF | 0.88 (0.55-0.97) | 2.48 | 0.29 | 0.10 | 0.28 |

*Table:4-12 Between-day ICC (95% CI), mean, SD, SEM and SDD for CV of lower limb kinematics and kinetics during running.*

| Variable | ICC (95% CI) | Mean $_{pooled}$ | SD $_{pooled}$ | SEM | SDD |
|---|---|---|---|---|---|
| CV of Angles | | | | | |
| Hip flexion | 0.73 (0.06-0.93) | 0.04 | 0.02 | 0.01 | 0.03 |
| Hip adduction | 0.68 (-0.44-0.92) | 0.14 | 0.15 | 0.08 | 0.24 |
| Hip internal rotation | 0.71 (-0.25-0.93) | 0.29 | 0.26 | 0.14 | 0.39 |
| Knee flexion | 0.76 (0.09-0.94) | 0.03 | 0.01 | 0.01 | 0.02 |
| Knee valgus | 0.81 (0.19-0.95) | 0.27 | 0.18 | 0.08 | 0.22 |
| Knee internal rotation | 0.81 (0.27-0.95) | 0.27 | 0.24 | 0.10 | 0.29 |
| Ankle dorsiflexion | 0.86 (0.46-0.97) | 0.04 | 0.02 | 0.01 | 0.02 |
| Ankle adduction | 0.65 (-0.24-0.91) | 0.36 | 0.28 | 0.16 | 0.45 |
| CV of Moments (Nm/Kg) | | | | | |
| Hip flexion | 0.83 (0.18-0.96) | 0.09 | 0.05 | 0.02 | 0.06 |
| Hip adduction | 0.58 (-0.79-0.90) | 0.06 | 0.03 | 0.02 | 0.05 |
| Hip internal rotation | 0.62 (-0.75-0.91) | 0.09 | 0.03 | 0.02 | 0.06 |
| Knee extension | 0.93 (0.58-0.98) | 0.05 | 0.03 | 0.01 | 0.02 |
| Knee valgus | 0.80 (0.29-0.95) | 0.32 | 0.28 | 0.12 | 0.33 |
| Ankle plantar flexion | 0.99 (0.98-0.99) | 0.05 | 0.08 | 0.01 | 0.02 |
| CV of GRF (*Body Weight) | | | | | |
| GRF | 0.99 (0.95-0.99) | 0.03 | 0.04 | 0.005 | 0.01 |

*Table:4-13 Between-day ICC (95% CI), mean, SD, SEM and SDD for lower limb kinematics and kinetics during cutting 90.*

| Variable | ICC (95% CI) | Mean $_{pooled}$ | SD $_{pooled}$ | SEM | SDD |
|---|---|---|---|---|---|
| Angles (°) | | | | | |
| Hip flexion | 0.74 (-0.14-0.94) | 53.91 | 6.45 | 3.29 | 9.12 |
| Hip adduction | 0.60 (-0.28-0.89) | -6.77 | 5.99 | 3.79 | 10.49 |
| Hip internal rotation | 0.77 (0.13-0.94) | 10.19 | 4.20 | 2.03 | 5.63 |
| Knee flexion | 0.89 (0.57-0.97) | 69.07 | 5.43 | 1.80 | 4.99 |
| Knee valgus | 0.76 (-0.14-0.94) | -4.73 | 4.45 | 2.17 | 6.03 |
| Knee internal rotation | 0.74 (0.06-93) | 5.38 | 5.41 | 2.76 | 7.66 |
| Ankle dorsiflexion | 0.81 (0.02-0.96) | 27.08 | 8.18 | 3.56 | 9.88 |
| Ankle adduction | 0.78 (0.07-0.95) | 15.47 | 4.49 | 2.08 | 5.78 |
| Moments (Nm/Kg) | | | | | |
| Hip flexion | 0.82 (0.27-0.96) | -2.18 | 0.75 | 0.31 | 0.87 |
| Hip adduction | 0.62 (-0.28-0.90) | -0.97 | 0.26 | 0.16 | 0.44 |
| Hip internal rotation | 0.96 (0.84-0.99) | -0.77 | 0.30 | 0.06 | 0.16 |
| Knee extension | 0.91 (0.62-0.98) | 3.01 | 0.45 | 0.14 | 0.38 |
| Knee valgus | 0.90 (0.60-0.98) | 0.65 | 0.28 | 0.09 | 0.25 |
| Ankle plantar flexion | 0.95 (0.78-0.99) | -2.15 | 0.49 | 0.11 | 0.31 |
| GRF (*Body Weight) | | | | | |
| GRF | 0.87 (0.57-0.97) | 2.14 | 0.41 | 0.14 | 0.39 |

*Table:4-14 Between-day ICC (95% CI), mean, SD, SEM and SDD for CV of lower limb kinematics and kinetics during cutting 90.*

| Variable | ICC (95% CI) | Mean $_{pooled}$ | SD $_{pooled}$ | SEM | SDD |
|---|---|---|---|---|---|
| CV of Angles | | | | | |
| Hip flexion | 0.64 (-0.61-0.91) | 0.05 | 0.03 | 0.02 | 0.05 |
| Hip adduction | 0.71 (-0.02-0.93) | 0.43 | 0.32 | 0.17 | 0.48 |
| Hip internal rotation | 0.94 (0.77-0.99) | 0.26 | 0.18 | 0.04 | 0.12 |
| Knee flexion | 0.67 (-0.18-0.92) | 0.04 | 0.02 | 0.01 | 0.03 |
| Knee valgus | 0.96 (0.85-0.99) | 0.29 | 0.27 | 0.05 | 0.15 |
| Knee internal rotation | 0.67 (-0.47-0.92) | 0.28 | 0.25 | 0.14 | 0.40 |
| Ankle dorsiflexion | 0.60 (-0.28-0.90) | 0.09 | 0.03 | 0.02 | 0.05 |
| Ankle adduction | 0.67 (-0.47-0.92) | 0.15 | 0.14 | 0.08 | 0.22 |
| CV of Moments | | | | | |
| Hip flexion | 0.61 (-0.53-0.90) | 0.12 | 0.06 | 0.03 | 0.10 |
| Hip adduction | 0.85 (0.44-0.96) | 0.15 | 0.07 | 0.03 | 0.08 |
| Hip internal rotation | 0.70 (-0.30-0.93) | 0.15 | 0.08 | 0.04 | 0.12 |
| Knee extension | 0.69 (-0.30-0.92) | 0.06 | 0.03 | 0.01 | 0.04 |
| Knee valgus | 0.61 (-0.45-0.90) | 0.22 | 0.14 | 0.09 | 0.25 |
| Ankle plantar flexion | 0.95 (0.78-0.99) | 0.07 | 0.04 | 0.01 | 0.02 |
| CV of GRF | | | | | |
| GRF | 0.78 (0.06-0.95) | 0.08 | 0.05 | 0.02 | 0.06 |

*Table:4-15 Between-day ICC (95% CI), mean, SD, SEM and SDD for lower limb kinematics and kinetics during SLL.*

| Variable | ICC (95% CI) | Mean $_{pooled}$ | SD $_{pooled}$ | SEM | SDD |
|---|---|---|---|---|---|
| Angles (°) | | | | | |
| Hip flexion | 0.99 (0.96-0.99) | 55.49 | 11.31 | 1.07 | 2.97 |
| Hip adduction | 0.82 (0.25-0.96) | 0.61 | 3.89 | 1.65 | 4.58 |
| Hip internal rotation | 0.47 (-0.77-0.86) | 7.49 | 3.42 | 2.50 | 6.92 |
| Knee flexion | 0.97 (0.88-0.99) | 71.76 | 13.59 | 2.35 | 6.52 |
| Knee valgus | 0.75 (0.08-0.94) | -1.42 | 3.80 | 1.90 | 5.27 |
| Knee internal rotation | 0.62 (-0.61-91) | 4.07 | 4.31 | 2.68 | 7.42 |
| Ankle dorsiflexion | 0.90 (0.59-0.97) | 30.77 | 5.92 | 1.91 | 5.29 |
| Ankle adduction | 0.72 (0.01-0.93) | 3.10 | 4.57 | 2.44 | 6.77 |
| Moments (Nm/Kg) | | | | | |
| Hip flexion | 0.83 (0.28-0.96) | -1.90 | 0.64 | 0.27 | 0.73 |
| Hip adduction | 0.88 (0.53-0.97) | -1.80 | 0.26 | 0.09 | 0.25 |
| Hip internal rotation | 0.90 (0.39-0.96) | -1.04 | 0.25 | 0.10 | 0.26 |
| Knee extension | 0.76 (0.05-0.94) | 3.69 | 0.42 | 0.21 | 0.58 |
| Knee valgus | 0.72 (-0.02-0.93) | 0.16 | 0.11 | 0.06 | 0.16 |
| Ankle plantar flexion | 0.99 (0.96-0.99) | -1.62 | 0.67 | 0.07 | 0.19 |
| GRF (*Body Weight) | | | | | |
| GRF | 0.92 (0.70-0.98) | 3.20 | 0.42 | 0.12 | 0.32 |

*Table:4-16 Between-day ICC (95% CI), mean, SD, SEM and SDD for CV of lower limb kinematics and kinetics during SLL.*

| Variable | ICC (95% CI) | Mean $_{pooled}$ | SD $_{pooled}$ | SEM | SDD |
|---|---|---|---|---|---|
| CV of Angles | | | | | |
| Hip flexion | 0.79 (0.08-0.95) | 0.08 | 0.03 | 0.02 | 0.04 |
| Hip adduction | 0.72 (-0.07-0.93) | 0.49 | 0.28 | 0.15 | 0.41 |
| Hip internal rotation | 0.81 (0.16-0.95) | 0.26 | 0.14 | 0.06 | 0.17 |
| Knee flexion | 0.81 (0.21-0.95) | 0.06 | 0.02 | 0.01 | 0.03 |
| Knee valgus | 0.83 (0.35-0.96) | 0.41 | 0.29 | 0.12 | 0.33 |
| Knee internal rotation | 0.51 (-0.42-0.87) | 0.28 | 0.17 | 0.12 | 0.33 |
| Ankle dorsiflexion | 0.67 (-0.47-0.92) | 0.07 | 0.05 | 0.03 | 0.07 |
| Ankle adduction | 0.86 (0.49-0.97) | 0.40 | 0.27 | 0.10 | 0.27 |
| CV of Moments | | | | | |
| Hip flexion | 0.86 (0.45-0.97) | 0.25 | 0.12 | 0.04 | 0.12 |
| Hip adduction | 0.60 (-0.83-0.90) | 0.10 | 0.04 | 0.03 | 0.08 |
| Hip internal rotation | 0.63 (-0.27-0.90) | 0.12 | 0.07 | 0.04 | 0.12 |
| Knee extension | 0.77 (0.16-0.94) | 0.06 | 0.02 | 0.01 | 0.03 |
| Knee valgus | 0.80 (0.18-0.95) | 0.68 | 0.26 | 0.12 | 0.32 |
| Ankle plantar flexion | 0.73 (-0.15-0.93) | 0.16 | 0.10 | 0.05 | 0.14 |
| CV of GRF | | | | | |
| GRF | 0.84 (0.34-0.96) | 0.06 | 0.03 | 0.01 | 0.04 |

*Table:4-17 Between-day ICC (95% CI), mean, SD, SEM and SDD for lower limb kinematics and kinetics during SLS.*

| Variable | ICC (95% CI) | Mean $_{pooled}$ | SD $_{pooled}$ | SEM | SDD |
|---|---|---|---|---|---|
| Angles (°) | | | | | |
| Hip flexion | 0.93(0.70-0.98) | 65.93 | 13.14 | 3.58 | 9.91 |
| Hip adduction | 0.86 (0.42-0.97) | 5.44 | 3.08 | 1.15 | 3.17 |
| Hip internal rotation | 0.80 (0.27-0.95) | 7.90 | 5.29 | 2.35 | 6.52 |
| Knee flexion | 0.93 (0.70-0.98) | 89.09 | 9.70 | 2.64 | 7.31 |
| Knee valgus | 0.87 (0.50-0.97) | 1.47 | 2.66 | 0.97 | 2.70 |
| Knee internal rotation | 0.74 (0.07-93) | 3.51 | 4.57 | 2.35 | 6.51 |
| Ankle dorsiflexion | 0.82 (0.27-0.96) | 43.07 | 3.44 | 1.45 | 4.01 |
| Ankle adduction | 0.84 (0.39-0.96) | 1.43 | 3.80 | 1.51 | 4.18 |
| Moments (Nm/Kg) | | | | | |
| Hip flexion | 0.93 (0.74-0.98) | -0.82 | 0.48 | 0.13 | 0.35 |
| Hip adduction | 0.88 (0.53-0.97) | -0.98 | 0.12 | 0.04 | 0.12 |
| Hip internal rotation | 0.93 (0.74-0.98) | -0.42 | 0.11 | 0.03 | 0.08 |
| Knee extension | 0.87 (0.49-0.97) | 2.11 | 0.16 | 0.06 | 0.17 |
| Knee valgus | 0.97(0.87-0.99) | 0.01 | 0.03 | 0.01 | 0.02 |
| Ankle plantar flexion | 0.87 (0.44-0.97) | -1.01 | 0.18 | 0.07 | 0.18 |
| GRF (*Body Weight) | | | | | |
| Vertical GRF | 0.92 (0.66-0.98) | 1.14 | 0.08 | 0.02 | 0.07 |

*Table:4-18 Between-day ICC (95% CI), mean, SD, SEM and SDD for CV of lower limb kinematics and kinetics during SLS.*

| Variable | ICC (95% CI) | Mean $_{pooled}$ | SD $_{pooled}$ | SEM | SDD |
|---|---|---|---|---|---|
| CV of Angles | | | | | |
| Hip flexion | 0.73 (-0.18-0.94) | 0.06 | 0.02 | 0.01 | 0.03 |
| Hip adduction | 0.50 (-1.33-0.88) | 0.31 | 0.19 | 0.14 | 0.38 |
| Hip internal rotation | 0.71 (-0.26-0.93) | 0.29 | 0.22 | 0.12 | 0.33 |
| Knee flexion | 0.93 (0.73-0.98) | 0.03 | 0.02 | 0.01 | 0.01 |
| Knee valgus | 0.68 (-0.07-0.92) | 0.38 | 0.24 | 0.14 | 0.38 |
| Knee internal rotation | 0.60 (-0.25-0.89) | 0.46 | 0.27 | 0.17 | 0.47 |
| Ankle dorsiflexion | 0.92 (0.67-0.98) | 0.03 | 0.01 | 0.00 | 0.01 |
| Ankle adduction | 0.64 (-0.58-0.91) | 0.52 | 0.28 | 0.17 | 0.47 |
| CV of Moments | | | | | |
| Hip flexion | 0.79 (0.24-0.95) | 0.17 | 0.12 | 0.05 | 0.15 |
| Hip adduction | 0.61 (-0.67-0.91) | 0.08 | 0.05 | 0.03 | 0.09 |
| Hip internal rotation | 0.69 (-0.07-0.92) | 0.13 | 0.07 | 0.04 | 0.11 |
| Knee extension | 0.87 (0.48-0.97) | 0.04 | 0.02 | 0.01 | 0.02 |
| Knee valgus | 0.79 (0.24-0.95) | 0.70 | 0.23 | 0.11 | 0.30 |
| Ankle plantar flexion | 0.64 (-0.37-0.91) | 0.12 | 0.05 | 0.03 | 0.09 |
| CV of GRF | | | | | |
| Vertical GRF | 0.77 (0.12-0.94) | 0.02 | 0.01 | 0.01 | 0.02 |

### 4.12.9 Discussion

Conducted among a healthy population, the aims of this study were to examine the between-day reliability of the lower extremity biomechanics using 3D motion analysis in four functional tasks and to establish the SEM for 3D biomechanical measurements during these four functional tasks. As well as to assess the reliability of CV (second order) as an outcome measure of variability in lower limb biomechanics during these four functional tasks.

The findings of this study show that all 3D lower limb kinematic and kinetic variables for all tasks reported fair to excellent between-day reliability rates, ranging in ICC values from 0.47 to 0.99. The majority of variables reported an ICC value of between 0.70 and 0.99, which is considered to be within the good to excellent range. However, the lowest ICC value was found to occur in the transverse plane of the hip internal rotation angle during SLL, where an ICC value of 0.47 was recorded. Furthermore, the hip internal rotation angle and knee internal rotation angle in the SLL task reported a low ICC value of 0.47 and 0.62 respectively, as compared to all the remaining variables when performing the same task. In general, transverse plane variables (hip and knee internal rotation angles) are less reliable when compared to other planes of movement, an outcome also reported in previous investigations (Alenezi et al., 2014; Ferber et al., 2002; Ford et al., 2007; Malfait et al., 2014; Nakagawa et al., 2014; Queen et al., 2006).

Moreover, several investigators demonstrated sagittal plane variables in particular as being more reliable compared to frontal and transverse variables in different tasks, such as Alenezi et al. (2016), Ferber et al. (2002) and Queen et al. (2006), who recorded this during running. Similar findings were recorded during vertical jumping by Alenezi et al. (2014), Ford et al. (2007) and Malfait et al. (2014). In addition, Alenezi et al. (2014) and Nakagawa et al. (2014) studied the SLS task and reported the same trend. The findings of this study in terms of the running, SLS and SLL tasks are in line with all previously mentioned studies. However, in the cutting 90 task, the sagittal plane variables were not shown to have a high level of reliability as compared to the frontal and transverse variables (Alenezi et al., 2016). In contrast, this study reported that the sagittal plane variables in the cutting to 90 degrees task recorded higher reliability scores than was the case for frontal and transverse variables in terms of kinematic variables. A recent study by Sankey et al. (2015) investigated the reliability of the cutting to

90 degrees task, but it did not use the ICC measurement.  This study focused on the inter-trial, inter-session and inter-observer variability using two modelling approaches; direct kinematic (DK) and inverse kinematic (IK); and task execution variables. The reliability data were quantified using the procedures outlined in Schwartz et al. (2004) which was mainly based on SEM.

A very recent study by Mok, Bahr, and Krosshaug (2018) examined reliability levels during two sport-specific sidestep cutting tasks. This current study's findings in relation to between-day reliability scores in the cutting to 90 degrees task are virtually similar to the results of the aforementioned study, excluding some exceptions. However, Mok, Bahr, and Krosshaug (2018) reported lower ICC values for sagittal planes, in particular the knee flexion angle and the plantar dorsiflexion angle (0.63 and 0.69, respectively) as compared to this study's findings (0.89 and 0.81, respectively). This may be attributable to differences in the cutting angle tasks between the two studies, as the former study did not clearly define the exact cutting angle they chose to use. In addition, they have pooled the results of two different groups of sporting activities (handball and football) which may have affected the results obtained, as football players are more susceptible to cutting-related injuries than handball players.

The reliability of kinematic and kinetic variables can be influenced by various factors. Marker-placement error exerts the greatest influence on reliability (Ferber et al., 2002; Queen et al., 2006). Meanwhile, other factors, such as skin-marker movement and task difficulty, can also affect reliability, especially in jumping and cutting tasks, as these require greater control and balance. Between-day variability has been shown to be associated with marker reapplication (Kadaba et al., 1989), but only one investigator attached markers to participants. To minimise the error, the CAST model was used, considered one of the best available models, as it can reduce skin movement artefacts by attaching markers to the centre of the segment.

Regarding GRF, the findings of this study showed that GRF in all tasks reported high ICC values (between 0.87 and 0.92) compared to other kinematic and kinetic variables. In addition, the ICC value of vertical GRF was more consistent than other variables. The reason for this was reported in the literature by Ferber et al. (2002), who stated that no markers are needed to gather GRF data, so there was no marker placement error.

SEM is a highly important measurement of reliability, as it allows the researcher to ascertain an approximation of the real change and to provide an error interval (Denegar & Ball, 1993). In addition, SEM provides an accurate assessment of between-test changes, and it can determine the sources of the changes (Domholdt, 2005; Fletcher & Bandy, 2008; Munro et al., 2012). In the current study, the SEM values of all variables were calculated. In relation to kinematic variables, highest SEM values were recorded in the frontal plane, in particular at the hip adduction angle during the cutting to 90 degrees task, where a SEM value of 3.79 degrees was recorded. However, the lowest SEM (0.97°) was reported at the knee valgus angle during a SLS. Similar results were reported for the running and cutting 90 tasks by Alenezi et al. (2016). According to (McGinley et al., 2009), an error of two degrees or less is highly acceptable. In addition, an error of between two and five degrees is considered to be reasonable but may need to be taken into consideration during the data interpretation process. However, the authors suggested that errors exceeding five degrees should raise concern and may mislead clinical interpretation if it is too large.

The kinetic variables for all tasks showed SEM values between 0.01 and 0.31 Nm/Kg. The hip flexion moment during cutting 90 was the highest (0.31 Nm/Kg), but the knee valgus moment during SLS was the lowest (0.01 Nm/Kg). Previous studies that investigated the reliability of lower limb biomechanical outcome measures in different tasks (Malfait et al., 2014; Nakagawa et al., 2014) all calculated SEM values for kinematics, but they did not provide these values for kinetic variables. However, two studies by Alenezi et al. (2016 & 2014) only provided SEM values for the kinetic variables during these four functional tasks. Therefore, this study provides SEM values for both kinematic and kinetic variables, as presented in the results tables.

In the current study, the reliability of the CV of peak angles and moments was tested. CV was used frequently in the literature to measure the variability between variables (Brown, Bowser, & Simpson, 2012; Brown et al., 2009; Queen et al., 2006; Bradshaw et al., 2007). It was stated that CV is unreliable between days as one of the null hypotheses of the study. The results of this study showed that the ICC values of the CV of all the kinematic and kinetic variables ranged between 0.50 and 0.99, reporting fair to excellent between-day reliability. The lowest ICC values initially appeared in the frontal plane at the hip adduction angle (0.50) during SLS and, secondly, in the transverse plane at the knee internal rotation angle during SLL (0.51). Depending on the results, the null hypothesis mentioned previously was rejected. Therefore, this study demonstrated that CV (second order) is a reliable measure of the variability of peak angles and moments. This result supports the selection of CV (second order) to measure variability in the next studies of this thesis.

The SEM values for CV of all kinematic and kinetic variables are also calculated and it ranged between 0.005 and 0.17. The highest SEM value (0.17) initially appeared in the transverse plane at the knee internal rotation angle during SLS and, secondly, in the frontal plane at the hip and ankle adduction, with both occurring during the cutting to 90 degrees task. However, the lowest SEM score (0.005) was reported in the VGRF during the running task. In terms of angles and moments, the lowest SEM score (0.01) was found to occur in the sagittal plane at the hip flexion angle (during running and SLS), knee flexion angle (during running, cutting 90 and SLS), dorsiflexion angle (during running and SLS) and plantar flexion moment (during running and cutting 90).

Finally, despite the results shown, this study has some limitations. The first limitation is that this study used specific laboratory settings and a specific model, as described previously. Therefore, the generalisability of the study could be affected for different laboratory settings and models. In addition, only healthy subjects were included in the study. As these functional tests are commonly used for other injured populations, further consideration should be given to testing reliability levels for these populations. Finally, the sample size was relatively small, being restricted to ten males, as the primary focus of this thesis is on male recreational athletes. Therefore, the findings may not applicable to a female population group.

#### 4.12.10 Conclusion

The findings to emerge from this study demonstrates that all 3D lower limb kinematic and kinetic variables are reliable in terms of the running, SLS, SLL and cutting to 90 degrees tasks. However, the levels of reliability were found to range between fair and excellent. In addition, all variables reported low SEM values, which means that the change was accepted as accurate. Finally, the current study demonstrates that CV (second order) is a reliable measure of variability in lower limb kinematic and kinetic variables.

# Chapter 5

## 5 Movement Variability and Biomechanics at the Knee Joint During Four Functional Sporting Tasks: Reference Values, Effect of Leg Dominance and Comparisons Between Tasks

### 5.1 Introduction

Functional sporting tasks, namely running, cutting to 90 degrees, single leg squat (SLS) and single leg landing (SLL), are all worthy of further investigation in terms of knee sport injuries, as many non-contact sport injuries occur during some of these activities. In the literature, multiple studies have investigated these functional tasks in order to achieve different aims. From a sport biomechanics perspective, it is very important to understand all biomechanical characteristics during these tasks in relation to their impact on either performance or prevention of sport injury. One of the essential requirements in understanding knee biomechanics is to identify the reference values for knee kinematics and kinetics during these functional tasks in healthy individuals. Acquiring these values will allow clinicians and researchers to carry out biomechanical assessments, having gained a good understanding of how these variables typically present in healthy individuals. It will also assist them to differentiate between normal and abnormal characteristics, in particular, in relation to the biomechanical variables that are considered to be risk factors for knee injury. Another very important consideration is to gain an understanding of MV during these functional tasks, as it has been linked to non-contact knee sport injuries (Baida, Gore, Franklyn-Miller, & Moran, 2018; James, 2004; Konradsen, 2002; Konradsen & Voigt, 2002).

In terms of non-contact ACL injury, the exact role that MV could potentially play in this type of injury continues to remain somewhat unclear. In order to carry out further investigations, several steps should be undertaken. The first of these include providing MV reference values during these functional tasks among a healthy population. Despite several studies providing normative values for a number of lower limb biomechanics, such as the knee valgus angle during specific functional tasks, no study has recorded MV reference values during these tasks. For example, Herrington and Munro (2010) provided normative values for the knee valgus angle during step landing and drop jump landing, using 2D motion analysis, where they tested 100 physically active individuals (50 males and 50 females); however, this was only at the point of the lowest knee flexion angle during the landing phase. In addition, in their research comprising 234 adolescents, McNair and Prapavessis (1999) provided normative values for GRF during double leg landing. However, the study was very limited in that investigations were carried out on one variable only. Furthermore, as it was conducted in 1999, not all risk factors for lower limb injury were identified at that time.

Holden, Boreham, Doherty, Wang and Delahunt (2015) also provided normative values for knee valgus and knee flexion in a cohort of 181 adolescents during countermovement jump landings, using 2D motion analysis. When undertaking large-scale normative value studies, 2D motion analysis is deemed to be the best method to adopt, as it is both a time and cost-effective tool. However, there are some disadvantages associated with this type of analysis as some very important data, such as kinetics, will be missed, as it does not require a fully equipped laboratory as is the case when conducting 3D motion analysis. In addition, another source of error in 2D motion analysis is known as crosstalk between planes, where increased knee flexion may resemble an increased valgus if out of plane (Piazza & Cavanagh, 2000). However, it is believed that an ACL injury occurs before achieving the maximum knee flexion. Therefore, this may not present as a significant issue unless maximum knee flexion is targeted.

Leg dominance has also attracted some researchers' attention as it is thought that it may affect lower limb biomechanics. It has been studied in various earlier research during different tasks: for example, running for females (Brown, Zifchock, & Hillstrom, 2014); cutting tasks (Greska et al., 2017; Pollard et al., 2018; Bencke et al., 2013; Marshall et al., 2014); and SLL (Mueske et al., 2019; Hughes, 2019; Ludwig, Simon, Piret, Becker, & Marschall, 2017; Mokhtarzadeh et al., 2017; McPherson et al., 2016). However, there is a lack of evidence to demonstrate the effect of leg dominance during SLS task. All the aforementioned studies reported no significant difference between the dominant and non-dominant legs in knee biomechanical variables, with one exception, namely Ludwig et al. (2017) who reported a significant difference during SLL. However, it is important to note that the latter authors conducted their study on a cohort of adolescent male soccer players and investigated only one variable (knee valgus angle). In addition, they used 2D motion analysis, whereas all the other studies used 3D motion analysis. These factors could explain why their findings differ from all other studies. Investigating limb-dominance differences during functional sporting tasks provides further information on non-contact ACL risk-related injury.

In addition, differences in knee biomechanics between different sporting tasks have been reported in several studies. For example, Chinnasee et al. (2018), Kristianslund and Krosshaug (2013) and O'Connor, Monteiro and Hoelker (2009) all identified significant differences in knee biomechanics between sidestep cutting and landing. In addition, Donnelly et al. (2012) reported significant differences between cutting tasks and running. Donohue et al. (2015) also presented similar results when they compared SLL to SLS. In fact, such investigations are very important in identifying differences in knee biomechanics between common sporting tasks, as they can assist in developing a greater understanding of the mechanisms of non-contact ACL injury and could help to address the biomechanical risk factors associated with this injury.

Whilst there are studies which document reference values for the aforementioned tasks and limbs, to the best of the author's knowledge no research to date has provided MV reference values for knee kinematics and kinetics for dominant and non-dominant legs during the four common sporting tasks, using 3D motion analysis. Therefore, this study primarily aims to provide reference values for MV during the aforementioned tasks, as well as to provide reference values for knee kinematics and kinetics associated with the risk of ACL injury during these tasks. In addition, it aims to compare knee biomechanics and MV between dominant and non-dominant legs during these four tasks. Finally, it also aims to compare knee biomechanics and MV between the four tasks.

## 5.2 Null hypothesis

1- $H0_1$: There are no significant differences in knee kinematics and kinetics between dominant and non-dominant legs during the four functional sporting tasks.

2- $H0_2$: There are no significant differences in knee kinematics and kinetics between the four functional sporting tasks.

3- $H0_3$: There are no significant differences in MV between dominant and non-dominant legs during the four functional sporting tasks.

4- $H0_4$: There are no significant differences in MV between the four functional sporting tasks.

## 5.3 Methods

### 5.3.1 Participants

Power analysis was performed using post hoc G*Power (Version 3.1.9.3), based on the peak knee valgus angle obtained in the results of this study, which specifically compared values between the cutting to 90 degrees and running tasks (Es = 0.55). The rationale for this decision was that the peak knee valgus angle was the main variable and it has been shown to be a risk factor in non-contact ACL injury as explained in chapter 3. The sample size (N=45) used in this study showed a statistical power of 97%, according to G*Power post hoc analysis, with the type 1 error alpha level set at 5% ($\alpha = 0.0$) (Faul, Erdfelder, Lang, & Buchner, 2007) (See Appendix 12.4). The participants' demographic profile is presented in Table 5-1. All participants completed a consent form prior to engaging in the test.

*Table:5-1 Participants' demographic profile*

| | Number (Gender) | Mean | SD |
|---|---|---|---|
| Age (years) | 45 (male) | 25.8 | 6.6 |
| Height (m) | 45 (male) | 1.71 | 0.06 |
| Body mass (kg) | 45 (male) | 65.7 | 10.3 |

### 5.3.2 Inclusion and exclusion criteria

Healthy subjects were selected on the basis of being within the 18–40 years age group, displaying good levels of health and fitness and engaging in moderate levels of physical activity, which has been defined as performing any sport or exercise for at least half an hour, three times a week, for the previous six months (Munro & Herrington, 2011). Subjects should be able to maintain normal balance, which was described by Bohannon, Larkin, Cook, Gear, and Singer (1984) as the ability to stand on one leg, with eyes closed, for 30 seconds. Additionally, the subjects should not have any record of injury to the lower extremities, pelvis or back in the previous year, which includes surgical operations, as well as being able to perform any given tasks without assistance. Within this context, an injury is regarded as any musculoskeletal discomfort or complaint that may inhibit the subject's ability to perform exercise. Individuals were excluded who were experiencing any kind of pathology or minor pain in their lower limbs, which could affect the outcome of testing, as well as those who had shown a tendency towards ACL injuries and those who did not sign the consent form.

### 5.3.3 Preparation and tasks undertaken

All participants were required to do some general warm-up exercises of ten minutes' duration, followed by a further five-minute warm-up on an exercise bike prior to beginning the test. The preparation process was explained in detail in Chapter 4 (Section 4.5). In terms of the tasks undertaken, all participants were invited to complete all four functional sporting tasks: SLS, running, cutting to 90 degrees and SLL with their two legs (dominant and non-dominant). They were required to perform these tasks, as outlined in Chapter 4 (Section 4.5.1 for running, Section 4.5.2 for SLS, Section 4.5.3 for SLL and Section 4.5.4 for cutting to 90 degrees). The randomisation process used for the tasks and legs was explained in Chapter 4 (Section 4.6).

### 5.3.4 Procedure

The biomechanical data were collected for this study using 3D motion analysis at two different laboratories. Firstly, the 3D motion analysis laboratory at the University of Salford was described in detail in Chapter 4 (Section 4.2.1). Secondly, the 3D motion analysis laboratory conducted at Imam Abdulrahman Bin Faisal University was outlined in detail in Chapter 4 (Section 4.2.2). We demonstrated in the previous chapter that there were no significant differences between the two data capture environments.

### 5.3.5 Outcome measures

The outcome measures in this chapter includes all those explained in Chapter 4 (Section 4.9).

### 5.3.6 Statistical analysis

All statistical analysis in this study was carried out using the Statistical Package for the Social Sciences (SPSS) (Version 24). Descriptive analysis (means and standard deviations) were performed. Prior to undertaking the data analysis, the normality distribution of the data was observed visually following application of a Shapiro-Wilk test. Differences between dominant and non-dominant legs were tested using a paired t-test for parametric variables (normally distributed) and a Wilcoxon Rank test for nonparametric variables (not normally distributed). The significance level was set at 5% (significant level when the p-value $< 0.05$, non-significant level when the p-value $> 0.05$). The effect size for each comparison was also provided, along with p-values, in order to be able to fully interpret the results. The effect size was calculated using Cohen's d Index (Cohen, 2013). Table 5-2 demonstrates the levels and classifications of effect size according to Cohen (2013).

*Table 5-2: Levels and classification of effect size*

| **Effect size (Es) value** | **Classification** |
|---|---|
| Es $< 0.2$ | Weak |
| $0.2 \leq$ \| Es \| $< 0.5$ | Small |
| $0.5 \leq$ \| Es \| $< 0.8$ | Moderate |
| \| Es \| $\geq 0.8$ | Large |

(Cohen, 2013)

In addition, one-way repeated-measures analysis of variance (ANOVA) with post hoc tests were used to assess differences between tasks for normally distributed data. However, the Friedman test was used, in combination with post hoc tests, in cases where the normality assumption was violated or outliers were detected. Epsilon (**ε**) correction was applied using the Huynh-Feldt method when the sphericity assumption was violated (Maxwell, Delaney, & Kelley, 2017). The Bonferroni correction was applied for the alpha level ($\alpha = 0.05$) for all comparisons carried out in this study (corrected $\alpha = 0.005$). Comparisons between tasks were conducted on the dominant leg for all variables. In terms of MV values, a zero value signals that no variability has occurred while, conversely, a score of one is indicative of the highest levels of variability.

## 5.4 Results

### 5.4.1 Comparisons between legs

#### 5.4.1.1 Overground running

All participants (n=45) performed forward overground running on the force plate with their two legs (dominant/non-dominant). Knee kinematic and kinetic reference values are shown in Table 5-3. MV reference values for each variable are also presented in Table 5-3. In general, no significant differences were found to occur between dominant and non-dominant legs in all biomechanical variables. The biggest difference was found in the peak knee flexion angle, where the non-dominant leg demonstrated greater knee flexion than the dominant one (Es =0.32); however, this was found not to be significant following Bonferroni correction (corrected $\alpha = 0.005$).

In addition, four figures were provided to illustrate knee motion and moment curves, which includes the 95% confidence interval for both legs in the sagittal and frontal planes throughout the entire (100%) stance period of the running task. They are as follows: Figure 5-1 displays the knee sagittal plane motion; Figure 5-2 displays the knee sagittal plane moment; Figure 5-3 displays the knee frontal plane motion; and Figure 5-4 displays the knee frontal plane moment. The V-GRF is also illustrated in Figure 5-5. All graphs presented in these figures represent average scores obtained for all participants (n=45). In terms of MV, no significant differences were found to occur between dominant and non-dominant legs, in all variables.

*Table 5-3: Knee kinematics, kinetics (peaks) and MV in each variable during running.*

| Variable | Dominant leg | | Non-dominant leg | | P-value | Effect size |
|---|---|---|---|---|---|---|
| | Mean | SD | Mean | SD | | |
| Knee flexion angle (°) | 45.93 | 5.74 | 48.00 | 6.81 | 0.03 | 0.32 |
| MV for knee flexion angle | 0.05 | 0.03 | 0.06 | 0.08 | 0.82 | 0.17 |
| Knee extension moment (Nm/Kg) | 2.99 | 0.53 | 3.07 | 0.53 | 0.21 | 0.15 |
| MV for knee extension moment | 0.09 | 0.12 | 0.09 | 0.04 | 0.49 | 0.00 |
| Knee valgus angle (°) | -1.16 | 5.17 | -0.17 | 4.42 | 0.18 | 0.21 |
| MV for knee valgus angle | 0.31 | 0.27 | 0.35 | 0.25 | 0.43 | 0.15 |
| Knee valgus moment (Nm/Kg) | 0.20 | 0.30 | 0.14 | 0.16 | 0.62 | 0.25 |
| MV for knee valgus moment | 0.54 | 0.23 | 0.55 | 0.29 | 0.97 | 0.04 |
| V-GRF (*Body weight) | 2.31 | 0.36 | 2.30 | 0.32 | 0.72 | 0.03 |
| MV for V-GRF | 0.04 | 0.10 | 0.03 | 0.06 | 0.36 | 0.08 |

*Negative (-) represents knee valgus angle and varus moment.*

* *Significant difference between dominant and non-dominant legs after Bonferroni correction ($p < 0.005$).*

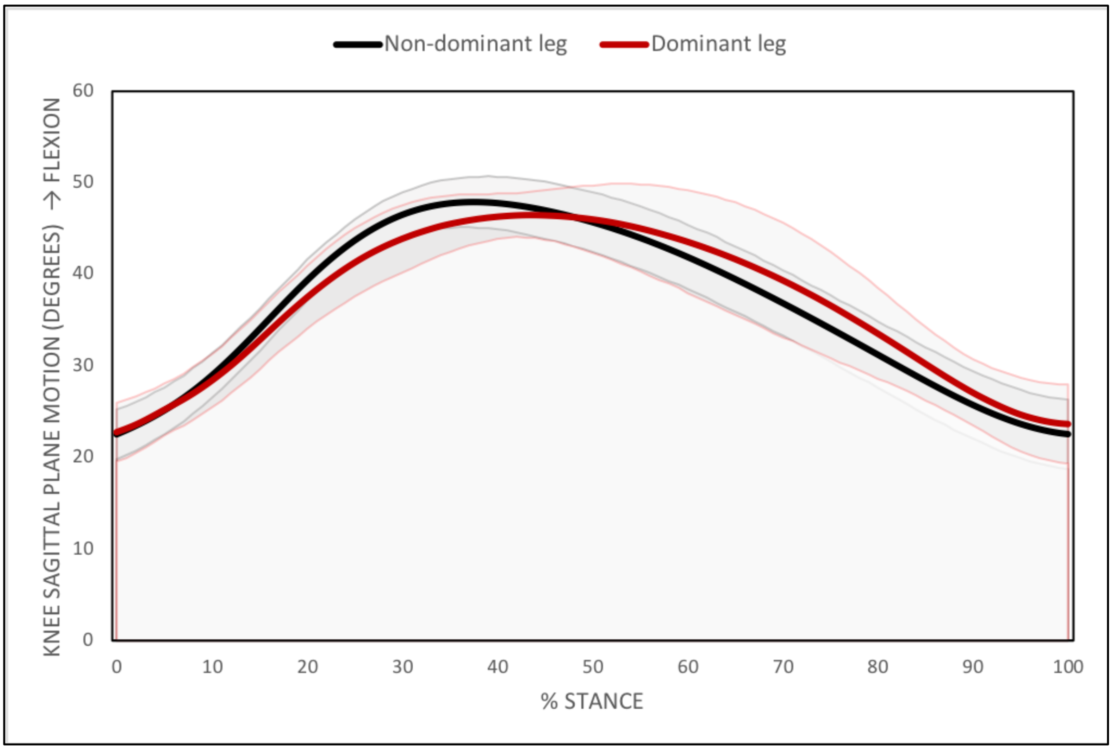



*Figure 5-1: Knee sagittal plane motion plus 95% confidence interval (shaded areas) for both legs during running task.*

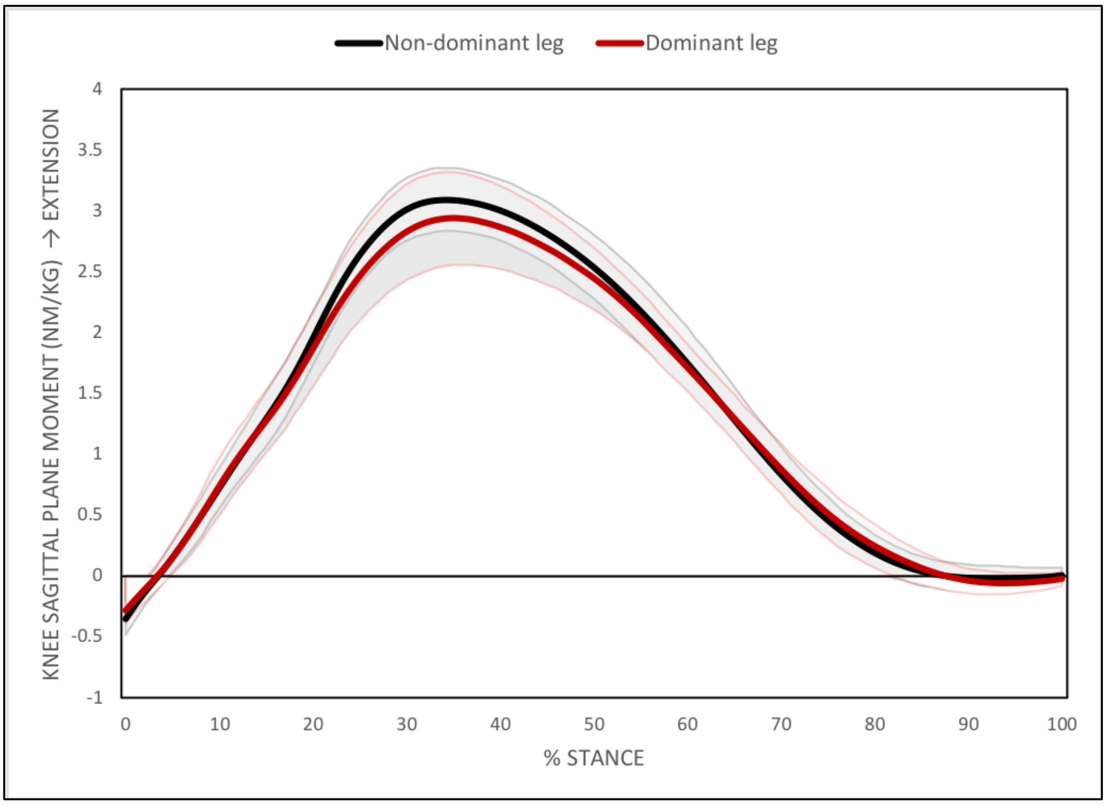



*Figure 5-2: Knee sagittal plane moment plus 95% confidence interval (shaded areas) for both legs during running task.*

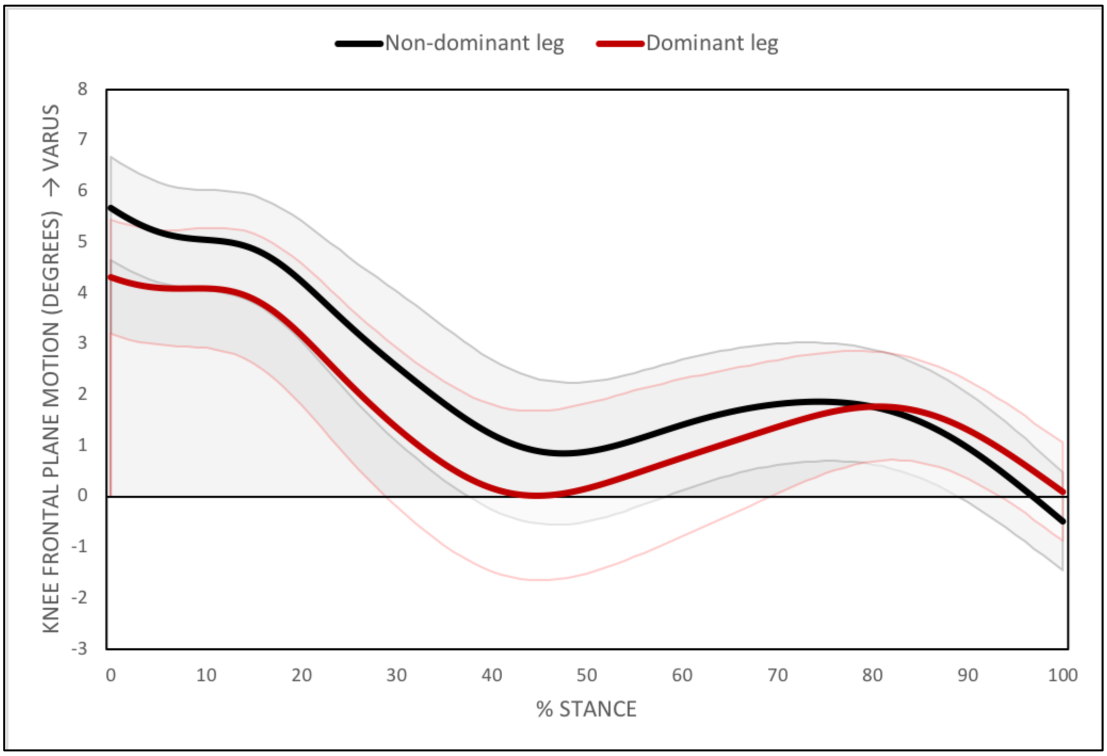



*Figure 5-3: Knee frontal plane motion plus 95% confidence interval (shaded areas) for both legs during running task.*

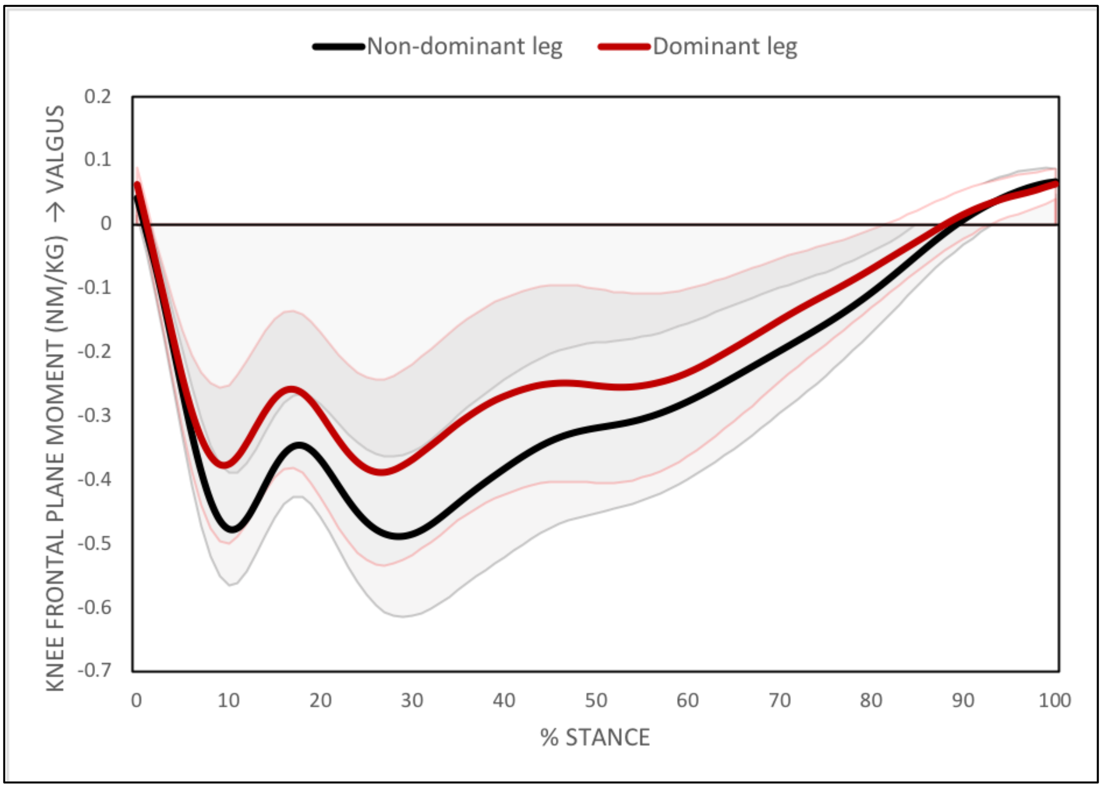



*Figure 5-4: Knee frontal plane moment plus 95% confidence interval (shaded areas) for both legs during running task.*

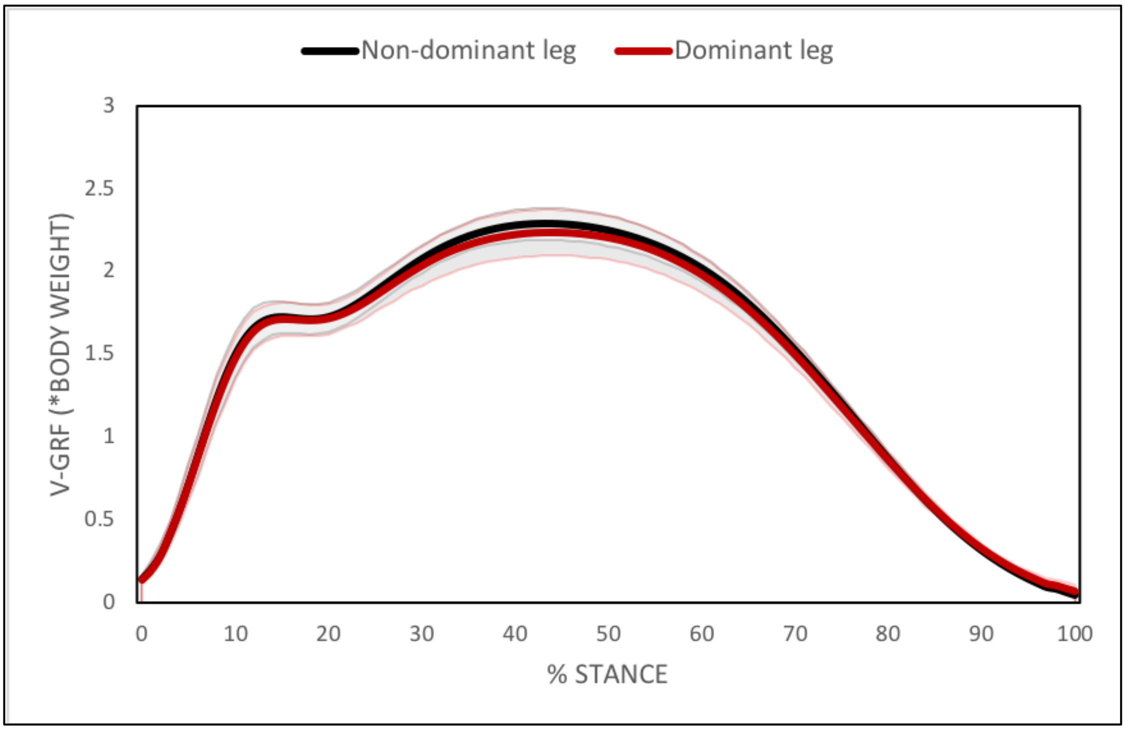



*Figure 5-5: Vertical ground reaction force plus 95% confidence interval (shaded areas) for both legs during running task.*

### 5.4.1.2 Cutting to 90 degrees

All participants (n=45) performed cutting to 90 degrees task on the force plate with their two legs (dominant/non-dominant). Knee kinematic and kinetic reference values are shown in Table 5-4. MV reference values for each variable are also presented in Table 5-4. In general, no significant differences were reported between dominant and non-dominant legs in all variables. In fact, the greatest differences between the two legs were in the peak knee flexion angle and the peak V-GRF. Similar to the running task, the non-dominant leg demonstrated greater knee flexion than the dominant one (Es = 0.36). In addition, the dominant leg demonstrated higher peak V-GRF than the non-dominant one (Es = 0.22). However, these differences were not significant following Bonferroni correction (corrected $\alpha = 0.005$).

Furthermore, four figures were provided to illustrate the knee motion and moment curves, which includes the 95% confidence interval for both legs in the sagittal and frontal planes throughout the entire (100%) stance period of the cutting to 90 degrees task. They are as follows: Figure 5-6 displays the knee sagittal plane motion; Figure 5-7 displays the knee sagittal plane moment; Figure 5-8 displays the knee frontal plane motion; and Figure 5-9 displays the knee frontal plane moment. The V-GRF is also illustrated in Figure 5-10. All graphs presented in these figures represent the average scores obtained for all participants (n=45). In terms of MV, no significant differences were found to occur between dominant and non-dominant legs, in all variables.

*Table 5-4: Knee kinematics, kinetics (peaks) and MV in each variable during cutting to 90 degrees task.*

| Variable | Dominant leg | | Non-dominant leg | | P-value | Effect size |
|---|---|---|---|---|---|---|
| | Mean | SD | Mean | SD | | |
| Knee flexion angle (°) | 63.60 | 8.06 | 66.35 | 7.41 | 0.04 | 0.36 |
| MV for knee flexion angle | 0.07 | 0.06 | 0.06 | 0.04 | 0.83 | 0.10 |
| Knee extension moment (Nm/Kg) | 2.49 | 0.89 | 2.52 | 0.65 | 0.50 | 0.04 |
| MV for knee extension moment | 0.13 | 0.08 | 0.12 | 0.07 | 0.44 | 0.13 |
| Knee valgus angle (°) | -4.09 | 5.37 | -3.23 | 4.99 | 0.33 | 0.17 |
| MV for knee valgus angle | 0.40 | 0.29 | 0.42 | 0.29 | 0.87 | 0.07 |
| Knee valgus moment (Nm/Kg) | 1.19 | 0.65 | 1.16 | 0.70 | 0.88 | 0.04 |
| MV for knee valgus moment | 0.28 | 0.14 | 0.30 | 0.14 | 0.49 | 0.14 |
| V-GRF (*Body weight) | 2.23 | 0.37 | 2.15 | 0.37 | 0.05 | 0.22 |
| MV for V-GRF | 0.09 | 0.07 | 0.09 | 0.06 | 0.54 | 0.02 |

*Negative (-) represents knee valgus angle and varus moment.*

** Significant difference between dominant and non-dominant legs after Bonferroni correction ($p < 0.005$).*

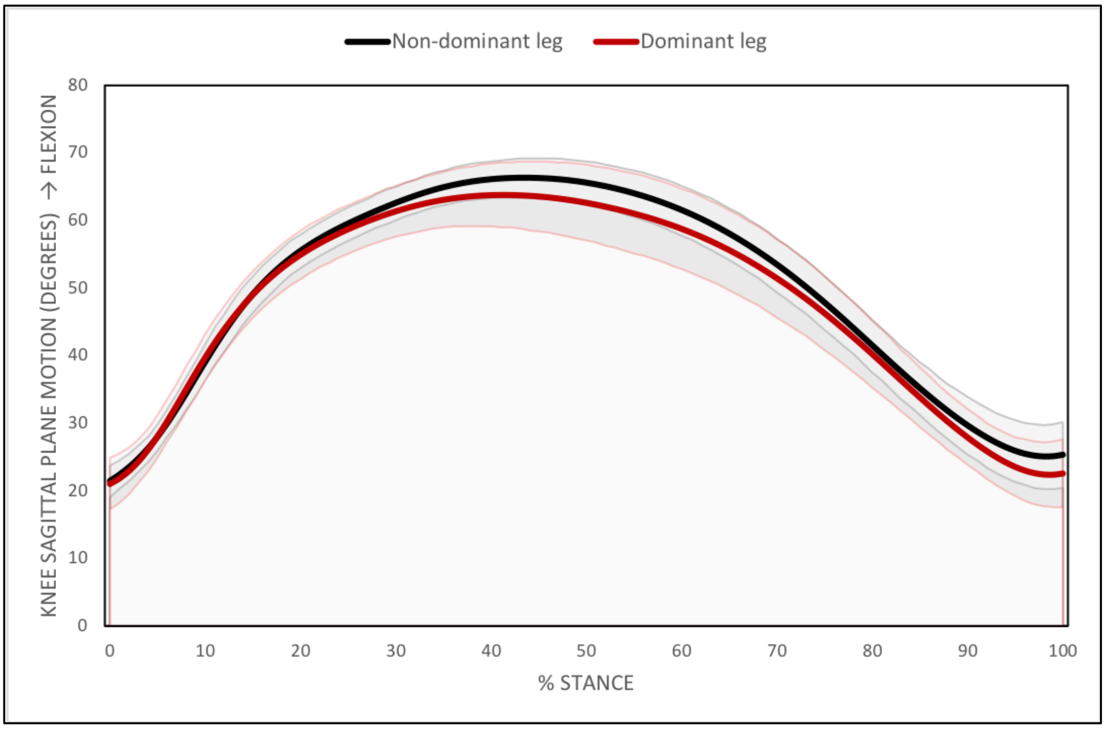



*Figure 5-6: Knee sagittal plane motion plus 95% confidence interval (shaded areas) for both legs during cutting to 90 degrees task.*

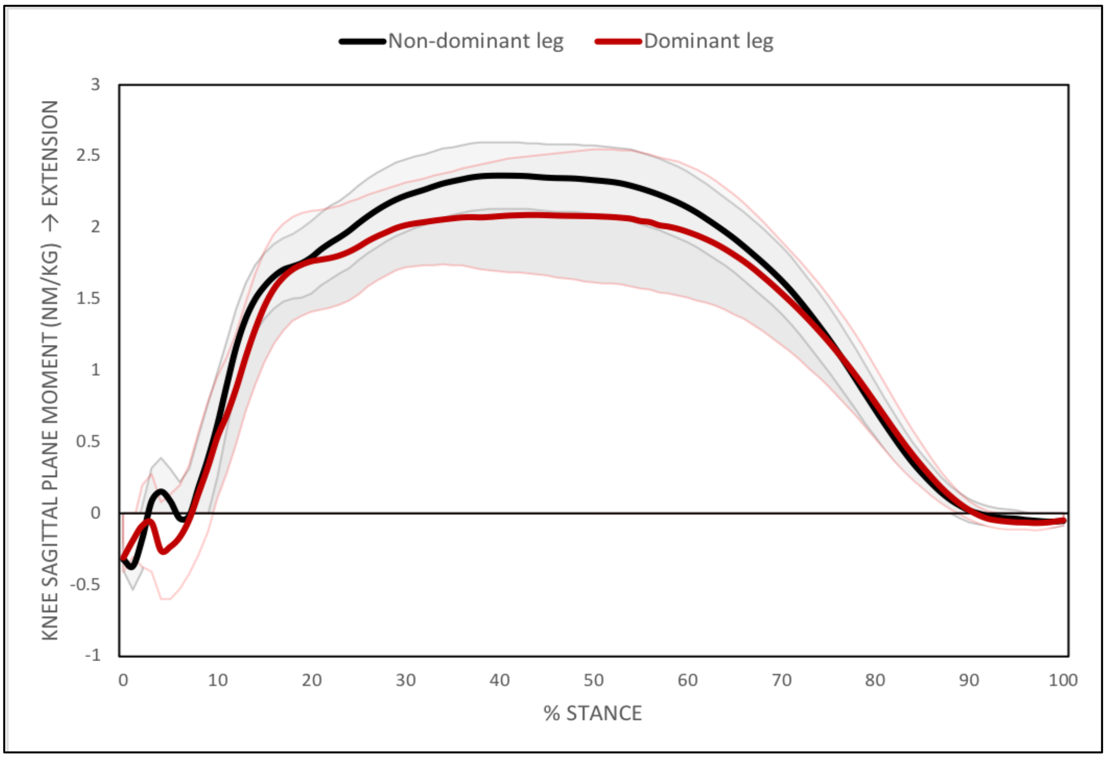



*Figure 5-7: Knee sagittal plane moment plus 95% confidence interval (shaded areas) for both legs during cutting to 90 degrees task.*

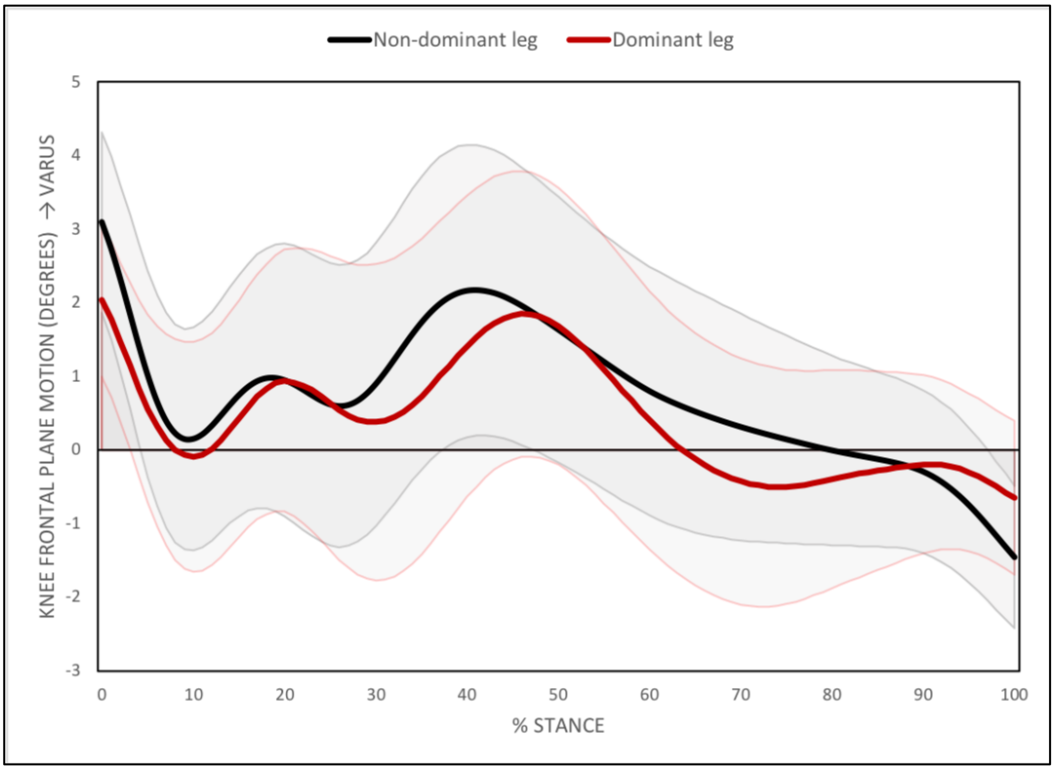



*Figure 5-8: Knee frontal plane motion plus 95% confidence interval (shaded areas) for both legs during cutting to 90 degrees task.*

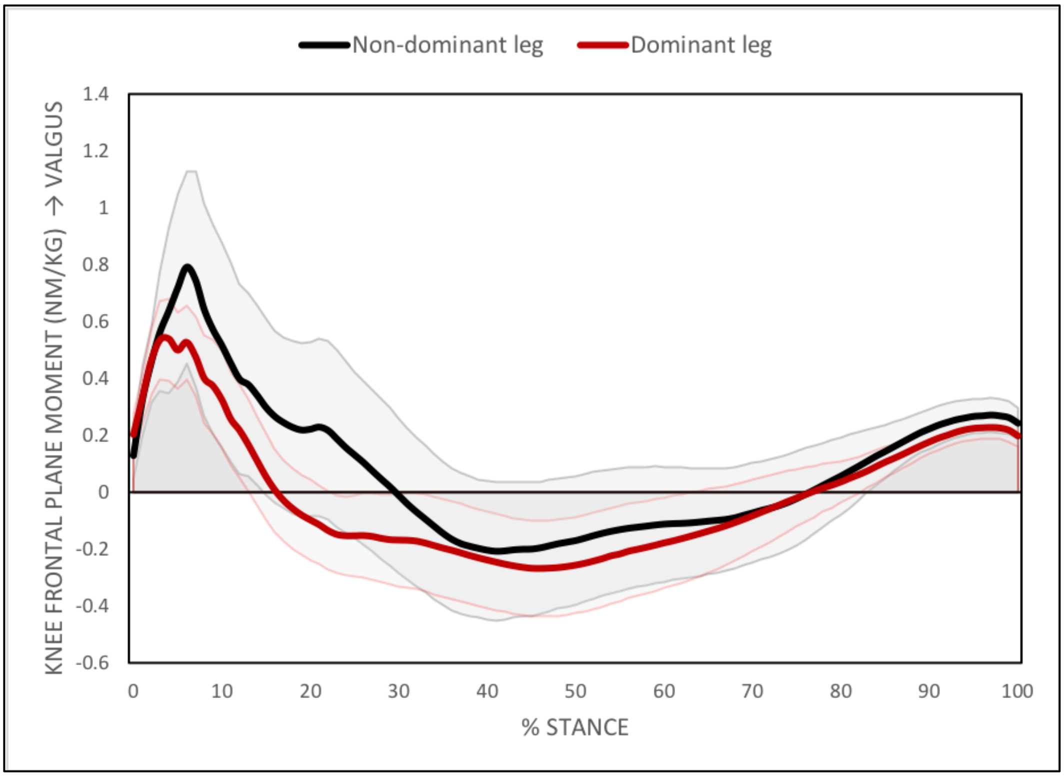



*Figure 5-9: Knee frontal plane moment plus 95% confidence interval (shaded areas) for both legs during cutting to 90 degrees task.*

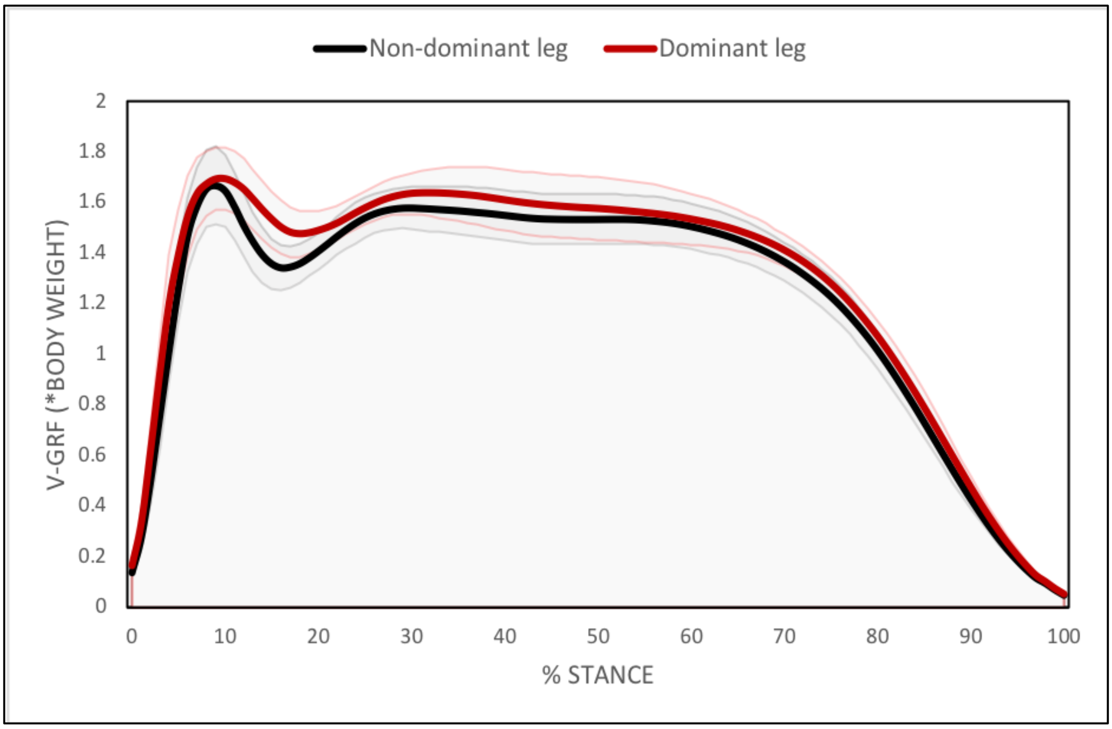



*Figure 5-10: Vertical ground reaction force plus 95% confidence interval (shaded areas) for both legs during cutting to 90 degrees task.*

**5.4.1.3 SLL**

Each participant in this study (n=45) performed SLL on the force plate with their two legs (dominant/non-dominant). Knee kinematic and kinetic reference values for this task are shown in Table 5-5. MV reference values for each variable are also presented in Table 5-5. Overall, in all the variables, no significant differences were reported between dominant and non-dominant legs.

In addition, four figures were provided to illustrate the knee motion and moment curves, which includes the 95% confidence interval for both legs in the sagittal and frontal planes throughout the entire (100%) defined phase of the SLL task. They are as follows: Figure 5-11 displays the knee sagittal plane motion; Figure 5-12 displays the knee sagittal plane moment; Figure 5-13 displays the knee frontal plane motion; and Figure 5-14 displays the knee frontal plane moment. The V-GRF is also illustrated in Figure 5-15. In terms of MV, no significant differences were found to occur between the dominant and non-dominant legs in all variables during this task. All graphs presented in these figures represent the average scores obtained for all participants (n=45).

*Table 5-5: Knee kinematics, kinetics (peaks) and MV in each variable during SLL*

| Variable | Dominant leg | | Non-dominant leg | | P-value | Effect size |
|---|---|---|---|---|---|---|
| | Mean | SD | Mean | SD | | |
| Knee flexion angle (°) | 65.81 | 13.02 | 66.13 | 13.38 | 0.79 | 0.02 |
| MV for knee flexion angle | 0.056 | 0.02 | 0.06 | 0.03 | 0.37 | 0.16 |
| Knee extension moment (Nm/Kg) | 3.45 | 0.46 | 3.49 | 0.44 | 0.45 | 0.09 |
| MV for knee extension moment | 0.06 | 0.03 | 0.06 | 0.03 | 0.86 | 0.00 |
| Knee valgus angle (°) | -0.66 | 4.39 | 0.09 | 4.70 | 0.14 | 0.16 |
| MV for knee valgus angle | 0.40 | 0.26 | 0.46 | 0.31 | 0.35 | 0.21 |
| Knee valgus moment (Nm/Kg) | 0.45 | 0.39 | 0.48 | 0.43 | 0.81 | 0.07 |
| MV for knee valgus moment | 0.50 | 0.23 | 0.50 | 0.23 | 0.71 | 0.00 |
| V-GRF (*Body weight) | 2.71 | 0.63 | 2.69 | 0.61 | 0.47 | 0.03 |
| MV for V-GRF | 0.03 | 0.03 | 0.03 | 0.03 | 0.32 | 0.03 |

*Negative (-) represents knee valgus angle and varus moment.*

** Significant difference between dominant and non-dominant legs after Bonferroni correction ($p < 0.005$).*

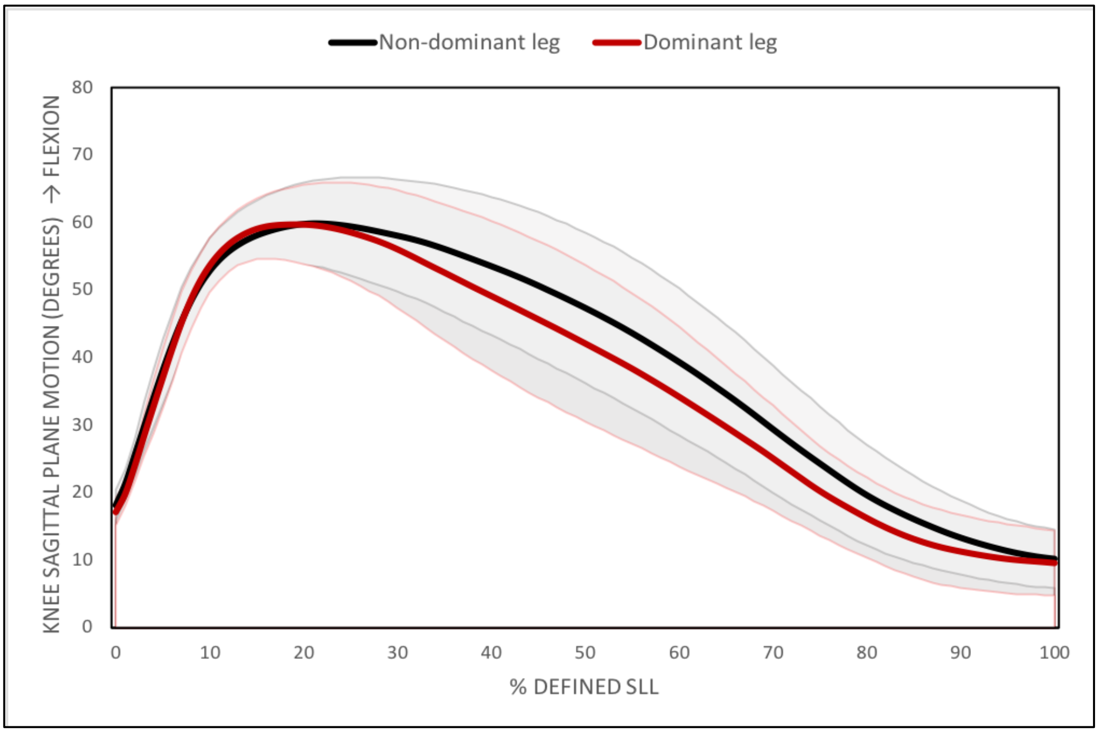



*Figure 5-11: Knee sagittal plane motion plus 95% confidence interval (shaded areas) for both legs during SLL task.*

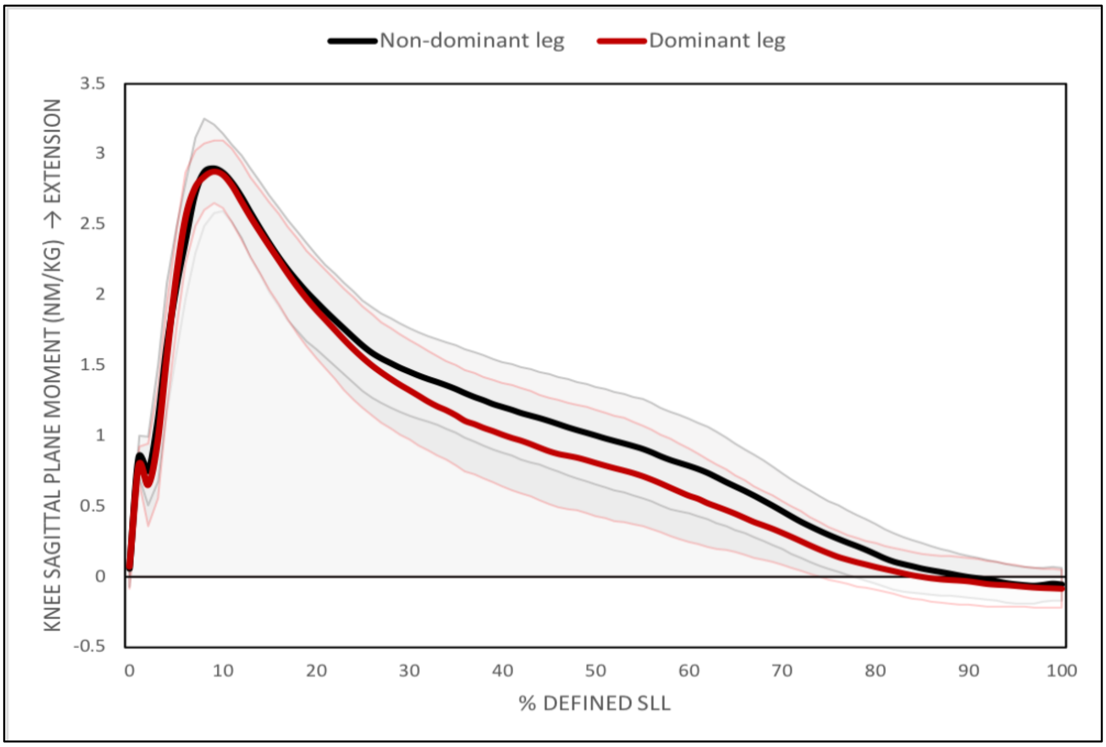



*Figure 5-12: Knee sagittal plane moment plus 95% confidence interval (shaded areas) for both legs during SLL task.*

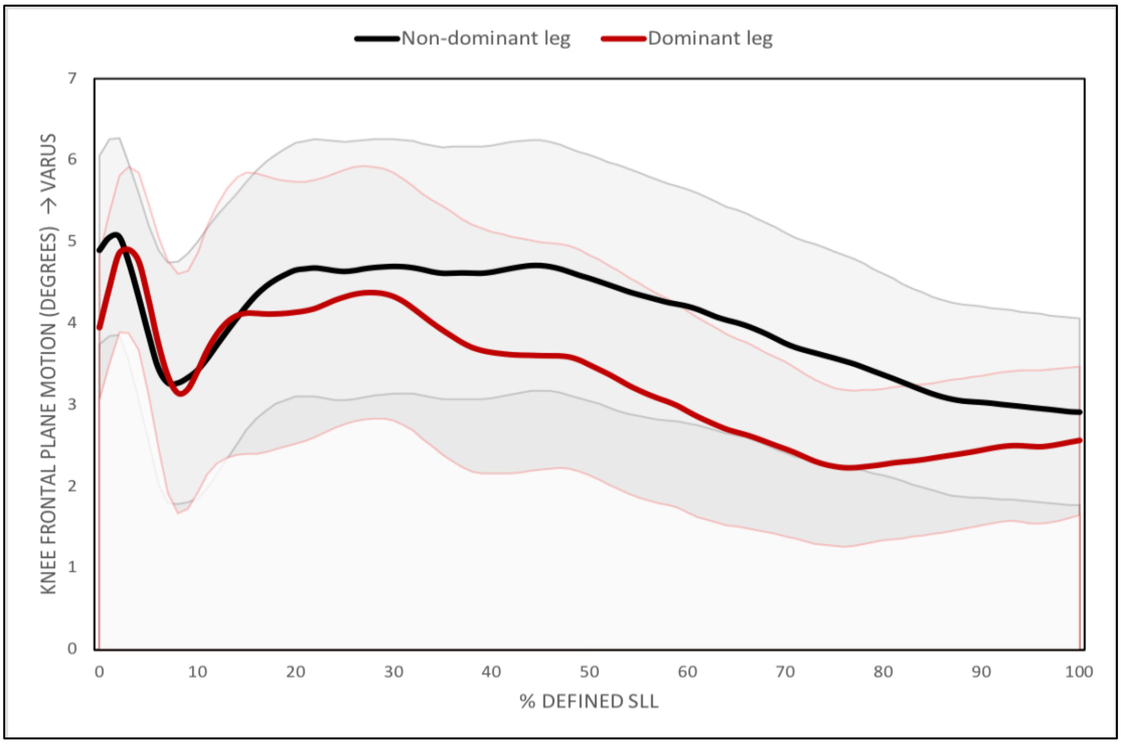



*Figure 5-13: Knee frontal plane motion plus 95% confidence interval (shaded areas) for both legs during SLL task.*

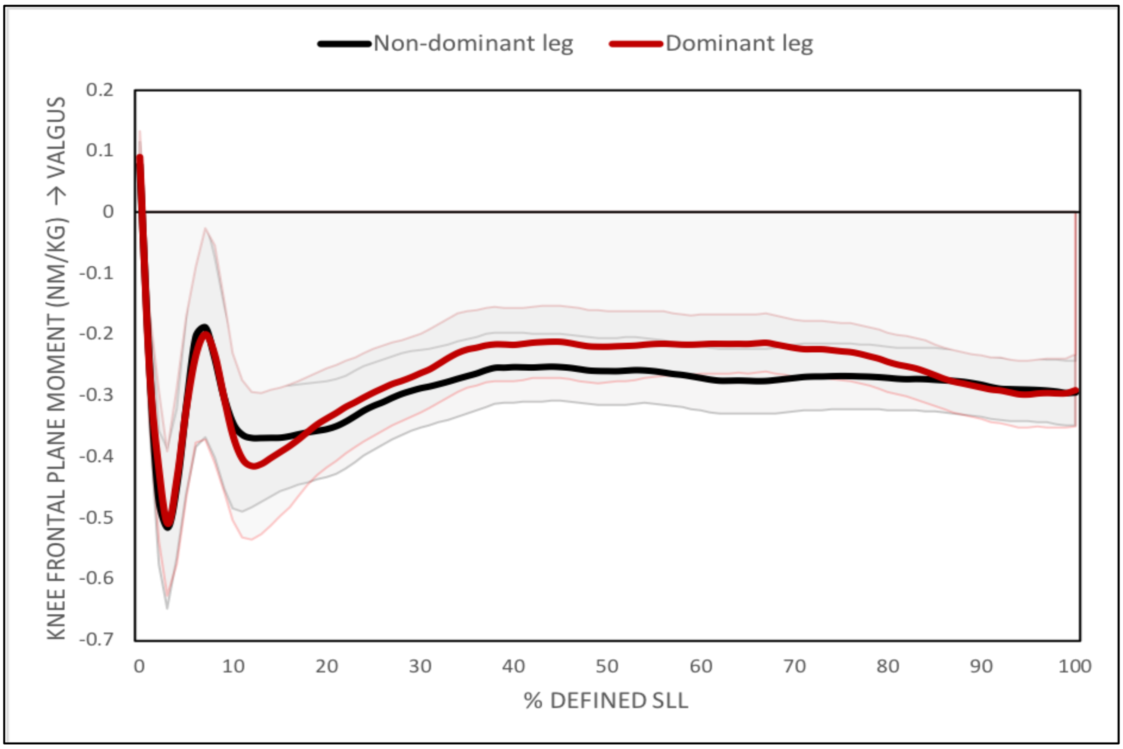



*Figure 5-14: Knee frontal plane moment plus 95% confidence interval (shaded areas) for both legs during SLL task.*

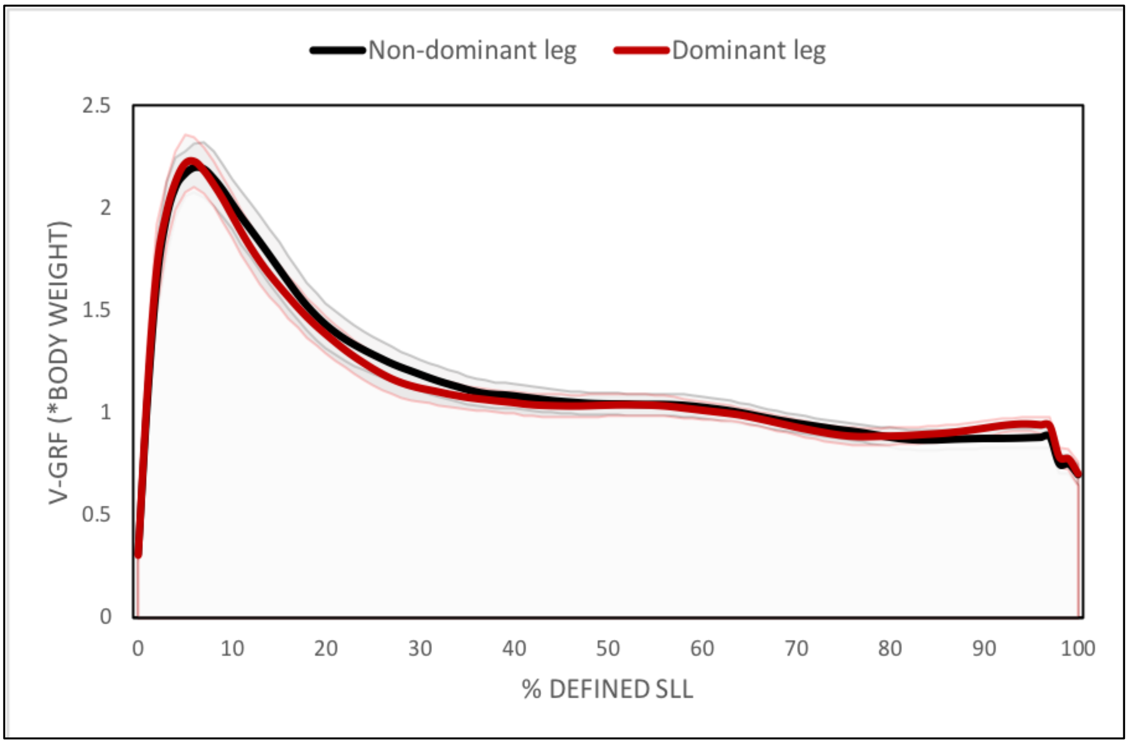



*Figure 5-15: Vertical ground reaction force plus 95% confidence interval (shaded areas) for both legs during SLL task.*

### 5.4.1.4 SLS

A total number of 45 participants performed SLS on the force plate with their two legs (dominant/non-dominant). Knee kinematics and kinetics values are outlined in Table 5-6. MV reference values for each variable are also presented in Table 5-6. During this task, all of the biomechanical variables reported no significant differences between the dominant and non-dominant legs. In addition, the dominant leg showed a higher knee extension moment than the non-dominant one (Es = 0.21). However, this difference was not significant following Bonferroni correction (corrected $\alpha = 0.005$).

In terms of MV, no significant differences occurred between dominant and non-dominant legs in all variables. Furthermore, four figures were provided to illustrate the knee motion and moment curves, which includes the 95% confidence interval for both legs in the sagittal and frontal planes throughout the entire (100%) defined phase of the SLS task. They are as follows: Figure 5-16 displays the knee sagittal plane motion; Figure 5-17 displays the knee sagittal plane moment; Figure 5-18 displays the knee frontal plane motion; and Figure 5-19 displays the knee frontal plane moment. The V-GRF is also illustrated in Figure 5-20. All graphs presented in these figures represent the average scores obtained for all participants (n=45).

*Table 5-6: Knee kinematics, kinetics (peaks) and MV in each variable during SLS*

| Variable | Dominant leg | | Non-dominant leg | | P-value | Effect size |
|---|---|---|---|---|---|---|
| | Mean | SD | Mean | SD | | |
| Knee flexion angle (°) | 85.81 | 11. 61 | 83.55 | 11.50 | 0.07 | 0.20 |
| MV for knee flexion angle | 0.05 | 0.04 | 0.04 | 0.03 | 0.52 | 0.32 |
| Knee extension moment (Nm/Kg) | 1.22 | 0.45 | 1.12 | 0.49 | 0.09 | 0.21 |
| MV for knee extension moment | 0.21 | 0.12 | 0.19 | 0.10 | 0.43 | 0.18 |
| Knee valgus angle (°) | -0.78 | 2.90 | -1.46 | 3.27 | 0.11 | 0.22 |
| MV for knee valgus angle | 0.43 | 0.32 | 0.38 | 0.26 | 0.45 | 0.17 |
| Knee valgus moment (Nm/Kg) | 0.06 | 0.06 | 0.05 | 0.08 | 0.36 | 0.14 |
| MV for knee valgus moment | 0.47 | 0.27 | 0.56 | 0.27 | 0.04 | 0.33 |
| V-GRF (*Body weight) | 1.15 | 0.12 | 1.16 | 0.12 | 0.07 | 0.08 |
| MV for V-GRF | 0.03 | 0.02 | 0.03 | 0.02 | 0.88 | 0.00 |

*Negative (-) represents knee valgus angle and varus moment.*

** Significant difference between dominant and non-dominant legs following Bonferroni correction ($p < 0.005$).*

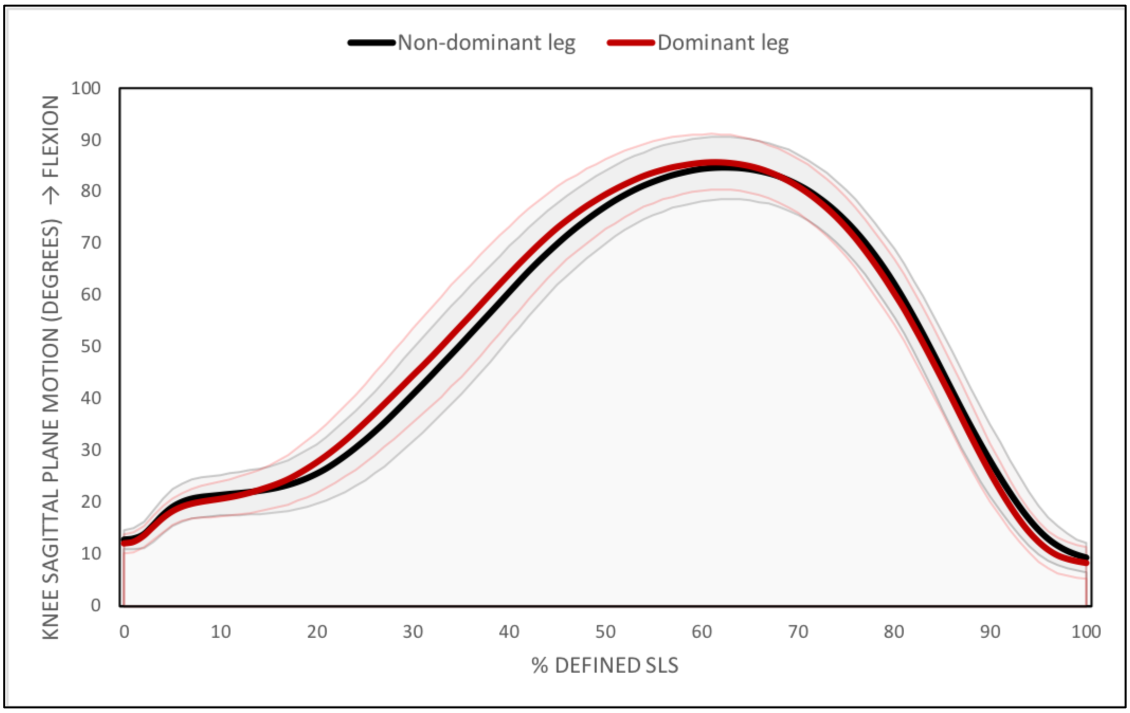



*Figure 5-16: Knee sagittal plane motion plus 95% confidence interval (shaded areas) for both legs during SLS task.*

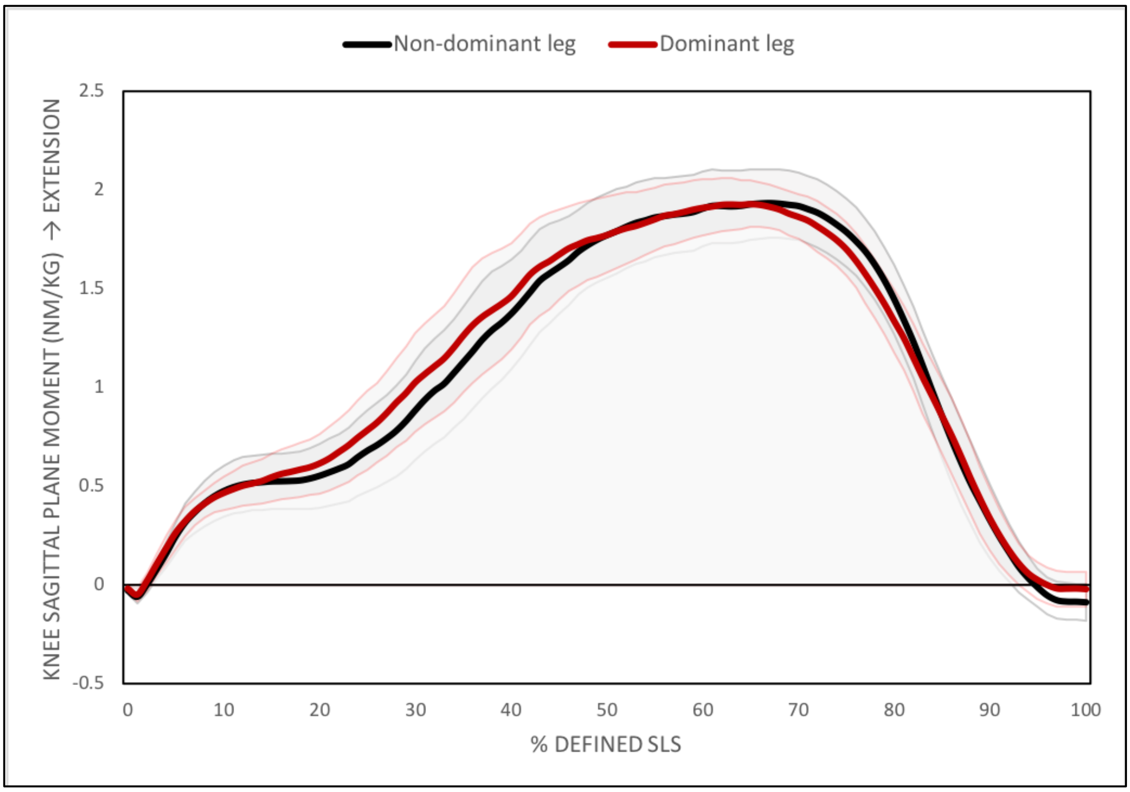



*Figure 5-17: Knee sagittal plane moment plus 95% confidence interval (shaded areas) for both legs during SLS task.*

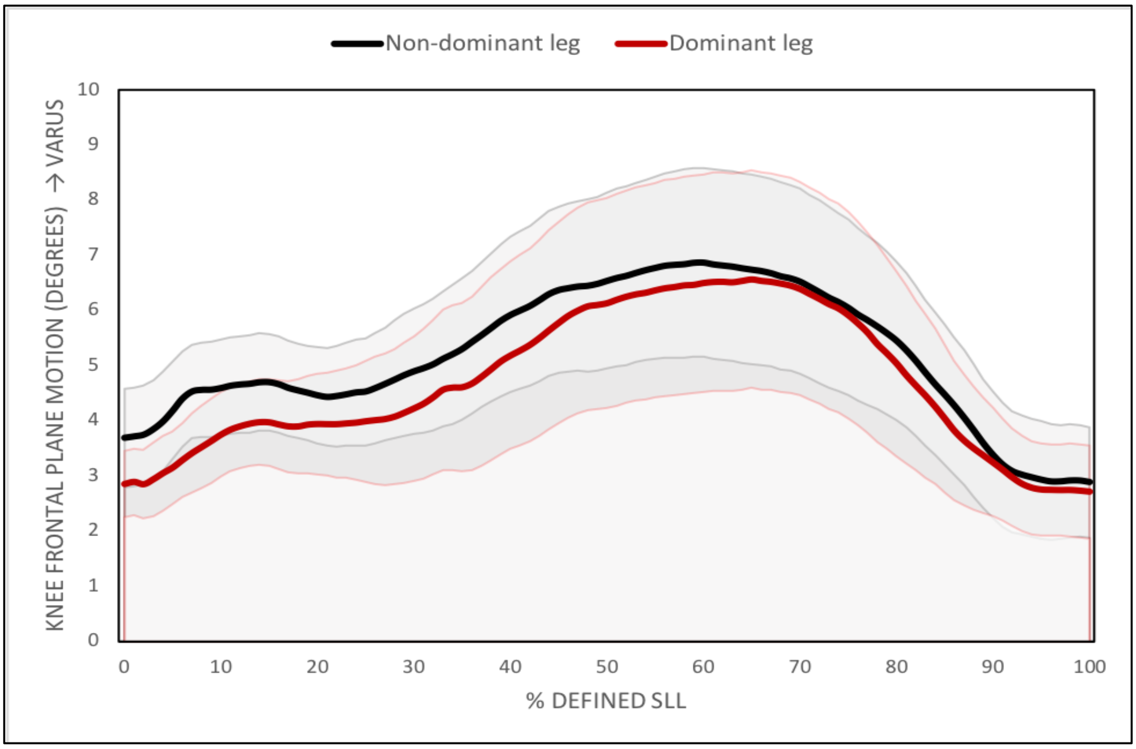



*Figure 5-18: Knee frontal plane motion plus 95% confidence interval (shaded areas) for both legs during SLS task.*

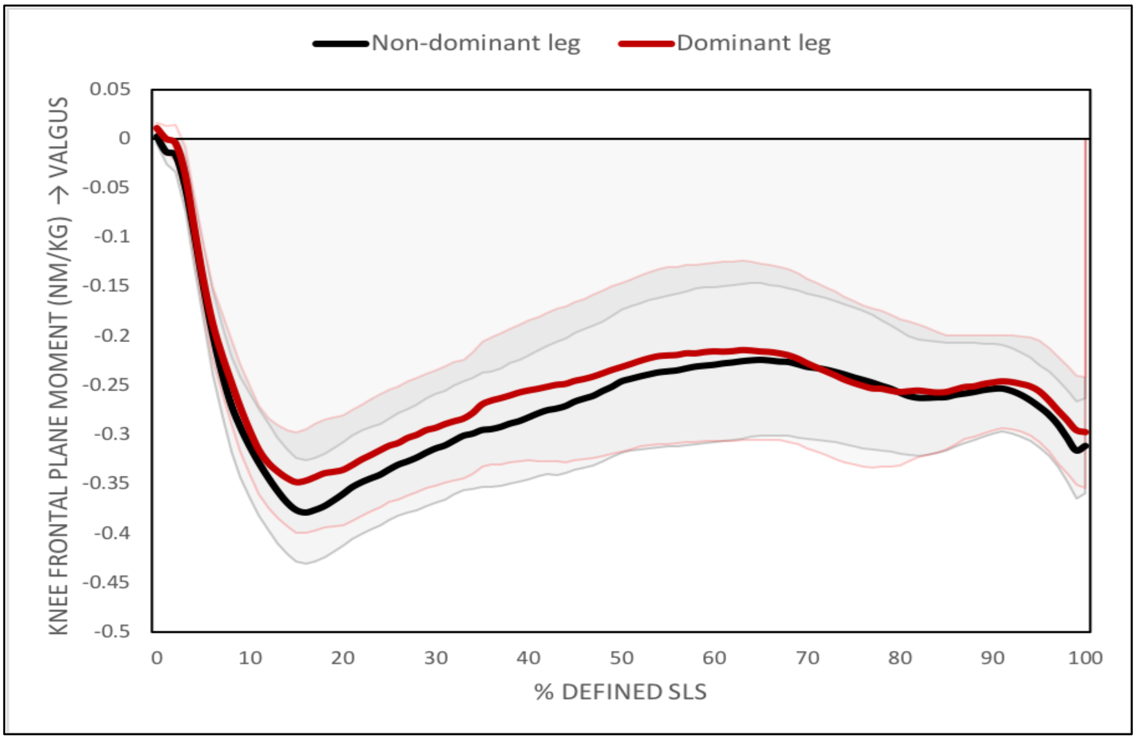



*Figure 5-19: Knee frontal plane moment plus 95% confidence interval (shaded areas) for both legs during SLS task.*

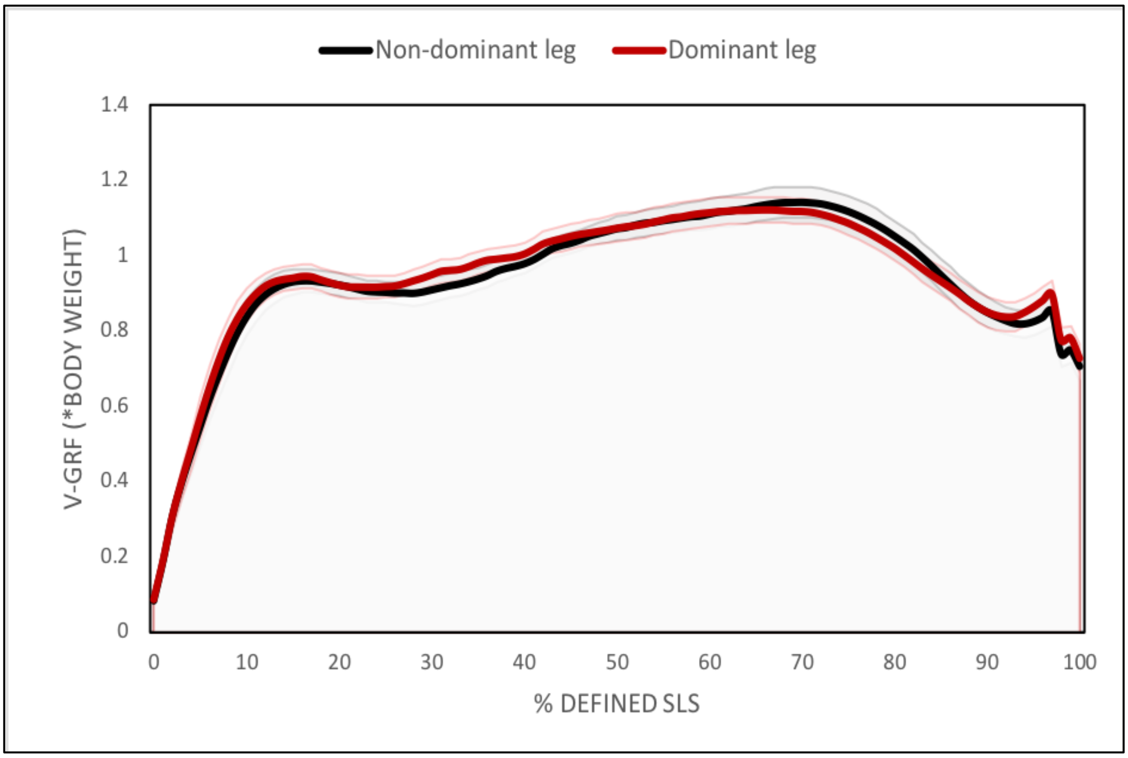



*Figure 5-20: Vertical ground reaction force plus 95% confidence interval (shaded areas) for both legs during SLS task.*

### 5.4.2 Comparisons between tasks

The same sample size (n=45), comprising the same subjects who underwent the comparison between legs, were also involved in the comparisons between tasks. As comparisons drawn between the two legs showed no significant differences in the case of all the variables, the between-task comparisons were conducted using data for the dominant leg. The kinematic and kinetic values, as well as the MV values are all presented in the following tables: Table 5-3 for running; Table 5-4 for cutting to 90 degrees; Table 5-5 for SLL; and Table 5-6 for SLS. The results of all post hoc tests (pairwise comparisons between tasks) are displayed in Table 5-7.

*Table 5-7: Results of repeated measures ANOVA*

| Variable | Pairwise comparisons | | | | | |
|---|---|---|---|---|---|---|
| | Cut90 vs Running | Cut90 vs SLL | Cut90 vs SLS | Running vs SLL | Running vs SLS | SLL vs SLS |
| Knee flexion angle | < 0.0005*† | 0.28 | < 0.0005* | < 0.0005* | < 0.0005* | < 0.0005* |
| MV for knee flexion angle | 0.01 | 0.15 | < 0.0005* | 0.29 | 0.11 | 0.008 |
| Knee extension moment | 0.01 | < 0.0005* | < 0.0005* | 0.02 | < 0.0005* | < 0.0005* |
| MV for knee extension moment | 0.002* | < 0.0005* | 0.007 | 0.41 | < 0.0005* | < 0.0005* |
| Knee valgus angle | < 0.0005* | < 0.0005* | < 0.0005* | 0.46 | 0.01 | 0.09 |
| MV for knee valgus angle | 0.03 | 0.37 | 0.87 | 0.002* | 0.02 | 0.46 |
| Knee valgus moment | < 0.0005* | < 0.0005* | < 0.0005* | 0.007 | 0.03 | < 0.0005* |
| MV for knee valgus moment | < 0.0005* | < 0.0005* | 0.009 | 0.57 | 0.03 | 0.09 |
| V-GRF | 0.46 | < 0.0005* | < 0.0005* | 0.002* | < 0.0005* | < 0.0005* |
| MV for V-GRF | < 0.0005* | < 0.0005* | < 0.0005* | 0.81 | 0.13 | 0.13 |

** Significant difference following Bonferroni correction ($p < 0.005$).*

*† < 0.0005 indicates that the SPSS reading was 0.000.*

In general, the results show that there was a statistically significant difference between tasks in at least three pairwise comparisons in each biomechanical variable and at least one pairwise comparison for each MV variable. The post hoc analysis shows that the number of pairwise comparisons demonstrating significant differences between tasks was higher in the biomechanical variables than in the MV variables. For example, in the peak knee flexion angle, five out of the six pairwise comparisons reported significant differences ($p < 0.0005$) between tasks. However, in the case of MV for the same variable (knee flexion angle), only one out of the six pairwise comparisons reported a significant difference and that was between cutting to 90 degrees and SLL ($p < 0.0005$).

## 5.5 Discussion

The main aims of this study were to:

1. Provide reference values for knee kinematics and kinetics during four common functional sporting tasks.
2. Provide reference values for MV for knee kinematics and kinetics during the aforementioned tasks.
3. Compare knee kinematics and kinetics between dominant and non-dominant legs during these four tasks.
4. Compare knee MV between dominant and non-dominant legs during each task.
5. Compare knee kinematics and kinetics between the four sporting tasks.
6. Compare MV between the four sporting tasks.

### 5.5.1 Running task

Providing reference values for knee kinematics and kinetics and for MV in a healthy population is very important, as these values can help researchers and clinicians to understand MV and biomechanics at the knee joint in this cohort. In addition, it allows clinicians and researchers to conduct biomechanical assessments, having gained a good understanding of how these variables typically present in healthy individuals. The reference values provided in this study for knee kinematics and kinetics for both dominant and non-dominant legs during forward running are presented in Table 4-3 (Section 4.4.1.1). However, these reference values only pertain to a healthy male cohort of recreational athletes. In the literature, several studies have investigated knee biomechanics among healthy males during running activities. The findings of this study are consistent with some earlier studies in relation to most of the variables (Bennett, Fleenor, & Weinhandl, 2018; Alenezi et al., 2016; Almonroeder & Benson, 2017; Ferber et al., 2003; Sakaguchi et al., 2014; Sinclair & Selfe, 2015). However, the peak knee valgus angle reported by Alenezi et al. (2016) (7.04°) and Bennett et al. (2018) (4.8°± 4.4) were higher than all others cited in studies which focused specifically on males. This has been attributed to the fact that they did not segregate the female data from the male, as females demonstrate higher knee valgus angles than males.

Several points need to be taken into consideration regarding these studies. Firstly, all studies failed to include both legs in their investigations. Furthermore, they all, with one exception, involved the right leg only, without providing a clear rationale for this decision. The only exception to this was the study conducted by Bennett et al. (2018), where the authors did not specifically state which leg was involved in their analysis, a practice which is questionable. In addition, only two studies, namely Ferber et al. (2003) and Sakaguchi et al. (2014) undertook power calculations to determine the sample size, while the remaining three authors failed to do so. A further consideration is that not all authors provided values for all knee kinematic and kinetic variables associated with risk of ACL injuries, except Bennett et al., (2018) and Alenezi et al. (2016). However, as previously mentioned, these authors did not provide values for both genders separately, as gender differences in biomechanics during this task have been reported by Sakaguchi et al. (2014). Hence, the current study is the first of its kind to involve both legs and to provide reference values for all knee biomechanics associated with the risk of ACL injuries for both dominant and non-dominant legs during running among a healthy male cohort of recreational athletes.

In terms of MV during running, the reference values obtained in this study for knee kinematics and kinetics for both dominant and non-dominant legs during forward running are as presented in Table 4-3 (Section 4.4.1.1). Unfortunately, in the literature, to the best of the author's knowledge, to date no study found has investigated MV for knee kinematics and kinetics during running using 3D motion analysis. Therefore, this study is the first of its kind to bridge this gap by including both legs. In addition, no statistical difference was found to occur between the dominant and non-dominant legs, in either the case of the biomechanical variables or in the MV variable. Similar results were reported for the knee biomechanical variables by Brown, Zifchock and Hillstrom (2014). However, the latter study only included females. In fact, the largest difference between limbs was found to occur in the knee flexion angle (Es =0.32); however, it was not statistically different following correction (corrected $\alpha = 0.005$).

### 5.5.2 Cutting task

Regarding the cutting task, the reference values provided by this study for knee kinematics and kinetics for both dominant and non-dominant legs during a cutting to 90 degrees task are as outlined in Table 4-4 (Section 4.4.1.2). However, these reference values only pertain to a healthy male cohort of recreational athletes. Cutting tasks have gained significant attention in the literature, as a large number of non-contact knee injuries occur during this task. The findings from this study are consistent with a number of earlier studies which investigated a cutting task at the same angle (90°) for most of the variables (Alhammad, Herrington, Jones, & Jones, 2019; Mcburnie et al., 2019; Schreurs et al., 2017; Alenezi et al., 2016; Suzuki, Ae, Takenaka, & Fujii, 2014; Havens & Sigward, 2015a). However, the peak knee valgus angle reported by Alenezi et al. (2016) (Mean = 11.8°) was higher than what emerged in this study (Mean = 4.09°), when males were investigated. This finding has been attributed to the fact that the former authors did not segregate male and female data, and females demonstrate higher valgus angles than males. Another possible reason could be the running speeds recorded, as they had a lower limit of 3 m/s, while the higher limit was not provided. This could mean that they ran faster than in other studies, as speed can affect knee biomechanics during this task (Dos'Santos, Thomas, Comfort, & Jones, 2018).

Researchers and clinicians need to take a number of points into consideration regarding these studies. Firstly, all studies failed to include both legs in their investigations. In addition, with the exception of one, all others involved the right leg only, without providing a clear rationale for this decision. Suzuki et al. (2014) failed to provide information regarding the leg involved in their analysis, which is a weakness of the study. In addition, two studies did not perform power calculations to determine the sample size (Alenezi et al., 2016; Suzuki et al., 2014). Another point worthy of consideration is that not all authors provided values for all knee biomechanical variables associated with the risk of ACL injuries, except Alenezi et al. (2016). The latter authors did not provide biomechanical values for both genders separately, and significant differences in knee biomechanics between genders during cutting tasks have been reported in a number of studies (Sinclair et al., 2019; Weinhandl, Irmischer, Sievert, & Fontenot, 2017; James et al., 2004; Ford, Myer, Toms, & Hewett, 2005). Another issue identified in the Mcburnie et al. (2019) study is that they defined the cutting angle as ranging between 70–90 degrees, while an earlier study has reported that differences in angles can change knee biomechanics (Schreurs et al., 2017). This could indicate that some participants might perform a cutting task with a difference of 20 degrees, which may be an issue. Hence, the current study is the first of its kind to include both dominant and non-dominant legs and to provide reference values for knee kinematics and kinetics during cutting to 90 degrees among a healthy male cohort of football players.

In terms of MV during cutting to 90 degrees, the reference values provided in this study for knee kinematics and kinetics for both dominant and non-dominant legs during this task are shown in Table 4-3 (Section 4.4.1.2). To date, to the best of the author's knowledge no study found has investigated MV for knee kinematics and kinetics during a cutting to 90 degrees task on healthy males with regard to ACL injury, using 3D motion analysis. Therefore, this study is the first of its kind to address this gap, by investigating both legs.

The current study also demonstrated that no significant differences occurred between the dominant and non-dominant legs, neither in terms of biomechanical variables nor the MV variable. However, the largest differences were reported in the peak knee flexion angle (Es = 0.36) and the peak GRF (Es = 0.22); however, this was not statistically different following Bonofroni correction (corrected $\alpha = 0.005$). In the literature, leg dominance has been investigated during cutting tasks but at different angles (Greska et al., 2017; Pollard et al., 2018; Bencke et al., 2013; Marshall et al., 2014). All these studies reported no significant differences between dominant and non-dominant legs at the knee joint, which concurs with our findings.

### 5.5.3 SLL task

With regard to the SLL task, the reference values provided in this study for knee kinematics and kinetics for both dominant and non-dominant legs are presented in Table 4-5 (Section 4.4.1.3). However, these reference values only pertain to a healthy male cohort of recreational athletes. Similar to the cutting task, landing tasks have attracted significant attention as many non-contact knee injuries occur during this type of activity. The findings of the current study are consistent with a number of previous studies which examined landing in the form of a SLL task for most of the variables (Bennett et al., 2018; Hong, Yoon, Kim, & Shin, 2014; Teng et al., 2017; Alenezi et al., 2014; Mokhtarzadeh et al., 2017; McPherson et al., 2016; Ford et al., 2006). However, the peak knee valgus angle reported by Herrington and Munro (2010) (4.9°±5.7 for left and 4.9°±6.7 for right) and Alenezi et al. (2016) (11.8°) was higher than was recorded in this study and the other aforementioned studies (mean = 0.09°– -1.6°), for a male cohort. This can be accounted for by the latter authors' failure to split the data based on gender, as females demonstrate higher valgus angles than males. With regard to the study by Herrington and Munro (2010), the contributory factor could be the different method they adopted, where they used 2D motion analysis at the point of maximal flexion angle during the landing phase so that cross-talk between planes could exist (Piazza & Cavanagh, 2000), whereas all other studies performed 3D motion analysis.

Readers should take some points regarding the aforementioned SLL studies into account. From the outset, it is important to note that not all studies included both legs in their investigations. The two exceptions to this were the study carried out by Herrington and Munro (2010) (which included both right and left legs) and McPherson et al. (2016) (which comprised both dominant and non-dominant legs). All the remaining authors, with one exception, focused on either the right leg or the dominant leg, without providing a rationale for this decision. Only one study, Bennett et al. (2018) did not provide information regarding the leg involved in the analysis, which is a deficit of the research. In addition, only one study by Teng et al. (2017) carried out power calculations to determine the sample size, while the remaining authors did not do so. Another salient point is that not all authors provided values for all knee biomechanical variables associated with the risk of ACL injuries, with the exception of Bennett et al. (2018) and Alenezi et al. (2014). However as previously, mentioned they did not provide values for both genders separately and this may have affected the values recorded, as gender-related biomechanical differences during this task have been reported in several recently conducted studies (Hughes, 2019; Weinhandl, Irmischer, Sievert, & Fontenot, 2017; Holden, Boreham, & Delahunt, 2016; Jenkins et al., 2017). Hence, the current study included both legs, dominant and non-dominant, in order to provide reference values for knee kinematics and kinetics associated with risk of ACL injuries during a SLL task among a healthy male population of recreational athletes.

In terms of MV during a SLL task, the reference values obtained in this study for knee kinematics and kinetics for both dominant and non-dominant legs during this task are shown in Table 4-5 (Section 4.4.1.3). Variability during the SLL task has been investigated in a recent study by Hughes (2019). However, the latter author aimed to investigate inter-limb coordination variability, using CRP (see Section 2.11.4.3), which differs from the method used in this study. Hughes found that no significant differences in coordination variability occurred between the dominant and non-dominant legs, and this concurs with the findings of this study. However, significant differences in coordination variability between males and females were reported. Another study by Brown et al. (2012) also investigated MV in lower limb kinematics and kinetics using the same method adopted in this study (CV). However, their focus was on ankle joint instability. In addition, as they performed data transformation and presented MV values in the form of transformed values, their findings were not easily comparable with ours. They found that the healthy group demonstrated greater variability at the knee than the functionally unstable and copers groups.

Another recent study by Nordin and Dufek (2017) also investigated MV in a healthy population using the CV method. However, their aim was to compare within subject variability during a SLL task between three different landing heights. They found that the CV decreased at greater landing heights ($p \leq 0.016$). However, they did not subdivide the data based on gender and they did not provide CV values, instead only showing the comparisons. Furthermore, they restricted their investigation to focus solely on the sagittal plane. Unfortunately, none of the two studies investigating MV based on the CV method used the second order version. As explained in Chapter 3 (Section 3.9), the older version of the CV contained some drawbacks, which may affect the data, especially biomechanical data. Thus, there was a need to gather reference values for MV in knee kinematics and kinetics using CV (second order). Therefore, one of the aims of this study is to provide MV reference values for knee kinematics and kinetics associated with ACL injury during a SLL task in healthy males using CV (second order) and 3D motion analysis.

The current study also demonstrated that no significant differences occurred between dominant and non-dominant legs, in relation to biomechanical variables or the MV variable during a SLL task. However, a difference between dominant and non-dominant legs was found in MV for the peak knee valgus angle, with an effect size of 0.21; however, it was not statistically significant ($p > 0.05$). In the literature, leg dominance have been investigated during a SLL task in several studies (Mueske et al., 2019; Hughes, 2019; Ludwig, Simon, Piret, Becker, & Marschall, 2017; Mokhtarzadeh et al., 2017; McPherson et al., 2016). All of the latter studies, with the exception of one, reported no significant differences between dominant and non-dominat legs at the knee joint, which is similar to the findings of this study. Only Ludwig et al. (2017) reported a significant difference between dominant and non-dominant legs. However, they conducted their study on a sample of adolescent male soccer players and investigated only one variable (knee valgus angle). In addition, they used 2D motion analysis, whereas all the other studies used 3D motion analysis. These factors could explain why their findings differ from all other studies.

### 5.5.4 SLS task

With regard to SLS, the reference values provided in this study for knee kinematics and kinetics for both dominant and non-dominant legs during this task are displayed in Table 4-6 (Section 4.4.1.4). However, these reference values are only applicable to a cohort of healthy male recreational athletes. SLS has also gained attention in the literature by many researchers as it is considered to be a very important screening tool, especially in relation to balance and pain. The findings of the current study are consistent with some previous research which studied leg squats in the form of SLS for most of the variables (Nakagawa, Maciel, & Serr, 2015; Alenezi et al., 2014; Graci, Van Dillen, & Salsich, 2012; Khuu, Foch, & Lewis, 2016; Yamazaki et al., 2010; Zeller et al., 2003). However, the peak knee valgus angles reported by Nakagawa et al. (2015) of -6.8°± 5.3, Alenezi et al. (2014) of -6.6° and Graci et al. (2012) of +7.0°± 4.12 differ from this study and what the other aforementioned studies have found (mean = -0.78°–-5.1°). The reason why these findings emerged in the Nakagawa et al., (2015) and Alenezi et al. (2014) studies is that they did not segregate the data by gender, and females demonstrate higher valgus angles than males. Surprisingly however, the study by Graci et al. (2012) reported that males demonstrate no valgus pattern whatsoever, where a mean peak knee valgus value of +7.0°± 4.12 was found to occur. This finding contradicts all previous study outcomes. This has been attributed to their small sample size, as only 10 males were included in the study.

Several points should be taken into consideration regarding these studies. Firstly, not all studies included both legs in their investigations. While Khuu et al. (2016) did, they found no significant difference between them, and, as a result, they carried out their investigations on one limb only. In addition, all but one of the remaining authors included either the dominant leg or the right leg, without providing a rationale for this decision. Only one study by Nakagawa et al. (2015) did not provide information regarding the leg involved in the analysis, a practice which is questionable. Unfortunately, none of the aforementioned studies undertook power calculations to determine the sample size, except Lewis et al. (2015). Furthermore, all studies comprised a small sample size (n = 7–10), except Yamazaki et al. (2010) (n=32). Another notable point is that not all authors provided values for all knee kinematics and kinetics associated with the risk of ACL injuries, except Alenezi et al. (2014).

Furthermore, while some studies comprised both genders (Nakagawa et al., 2015; Lewis et al., 2015; Alenezi et al., 2014), they did not provide values for each separately. Instead, they combined the data for both genders, which may have affected the real values attained, as significant gender differences in knee biomechanics, while performing tasks, have been reported in several studies (Dwyer et al., 2010; Graci et al., 2012; Weeks et al., 2015; Yamazaki et al., 2010; Zeller et al., 2003). It should be noted that there is a need for studies to provide gender-specific reference values for both kinematics and kinetics of the knee joint, which are associated with the risk of ACL injuries. Hence, the current study included both legs, so as to provide reference values for knee kinematics and kinetics associated with risk of ACL injury for dominant and non-dominant legs during SLS among a healthy male cohort of recreational athletes.

In terms of MV during SLS, the reference values obtained in this study for knee kinematics and kinetics for both dominant and non-dominant legs during this task are shown in Table 4-6 (Section 4.4.1.4). MV during SLS has also been investigated in a recent study (Severin, Burkett, McKean, Wiegand, & Sayers, 2017). However, its primary aim was to investigate differences in kinematics and MV between land and water during three different types of squats. They used inertial sensors for biomechanical analyses and CV to measure MV. Although they carried out kinematic investigations, they did not provide MV values for knee angles, but rather instead only reported values for variability of the movement in the three segments in general. Thus, there is a need to compile MV reference values for knee kinematics and kinetics. In fact, only very limited evidence is available in relation to MV during SLS tasks. Therefore, one of the aims of this study is to provide MV reference values for knee kinematics and kinetics associated with ACL injury during SLL in healthy males, using 3D motion analysis.

The findings of this study also showed no significant differences between dominant and non-dominant legs either in relation to the biomechanical variables or the MV variable during SLS. Leg dominance has not been widely investigated to date in the literature during SLS tasks. Khuu et al. (2016) only investigated leg differences and found no significant differences between them. However, the latter authors did not reveal the values recorded for each leg, and only briefly alluded to them in their methods section to justify their leg selection. Fortunately, this study now has clarified this issue.

On examination of MV reference values during running, cutting to 90 degrees and SLL tasks, we note that the MV values for the peak valgus angle and the peak valgus moment are higher than those in the peak knee flexion angle and the peak GRF. Interestingly, variables with higher MV values are considered to be risk factors for ACL, particularly if they increase, as explained in Chapter 2 (Hewett et al., 2005; Numata et al., 2018; McLean, Lipfert, & Van Den Bogert, 2004; Leppänen et al., 2017). This could imply that a relationship exists between MV and biomechanical risk factors.

One explanation as to why no significant differences emerged between knee biomechanics and MV in the dominant and non-dominant legs could be that the participants displayed almost similar neuromechanical and dynamic control in both legs. However, in the literature, leg dominance has been suggested to be one of several factors associated with non-contact ACL injury (Hewett et al., 2005; Pappas, Shiyko, Ford, Myer, & Hewett, 2016; Brophy, Silvers, Gonzales, & Mandelbaum, 2010; Maulder, 2013; Pappas & Carpes, 2012; Paterno et al., 2010), as this type of injury is multifactorial (Hewett, 2017; Quatman, Quatman-Yates, & Hewett, 2010).

### 5.5.5 Between task comparisons

One of the main aims of this study was to compare knee biomechanics and MV in four common sporting tasks. In general, the results showed that there were significant differences between tasks in at least three pairwise comparisons, in each biomechanical variable and at least one pairwise comparison in each MV variable. Post hoc analysis showed that the number of pairwise comparisons reporting significant differences between tasks was higher in biomechanical variables than in MV variables. Beginning with the peak knee valgus angle, there were significant differences between the cutting 90 degrees and the other three tasks ($p < 0.0005$). Similarly, in relation to the peak knee valgus moment, there were significant differences between the cutting task and all three other tasks ($p < 0.0005$). In fact, these findings suggest that cutting to 90 degrees puts the knee joint in a greater valgus than the other tasks, and this could lead to an increased risk of ACL injury. Previous research by Chinnasee et al. (2018), Kristianslund and Krosshaug (2013) and O'Connor, Monteiro and Hoelker (2009) all reported similar results when they compared sidestep cutting to landing. In addition, Donnelly et al. (2012) compared the cutting task with running. However, all these studies adopted a lower cutting angle (69° and 45°) than this specific research.

In addition, there was a significant difference in the peak knee valgus moment between SLL and SLS ($p < 0.0005$). Donohue et al. (2015) also reported similar results when they compared SLL to SLS. Munro et al. (2017) also undertook a comparison between SLL and SLS tasks, where they reported that the knee valgus angle was greater in SLS than SLL. However, the current study found that there was no significant difference in peak knee valgus angle between SLL and SLS ($p = 0.09$). One reason to explain this different finding could be because the focus of the study by Munro et al. (2017) was on females, whereas this study drew upon males. In addition, they used 2D motion analysis, whereas this study used 3D motion analysis. However, no significant differences were found in either the peak knee valgus angle or the peak knee valgus moment between running and SLS. Accordingly, these findings can assist in placing the four common tasks in rank order of riskiness. Cutting to 90 degrees was identified as most risky, as it places the knee joint at the greatest valgus angle, as compared to the three other tasks. The SLL is the second most risky task, while running and SLS were ranked lowest in terms of risk, demonstrating no significant difference between one another. Having the capacity to rank these four common tasks in order of riskiness is very important for clinicians, so as to inform and build a strong evidence base to develop either prevention or rehabilitation programmes.

Another important variable is the peak GRF. Post hoc pairwise comparisons showed that there were significant differences in all comparisons between the tasks, with one exception. There was no significant difference in the peak GRF between the cutting task and running.

Importantly, the peak GRF during SLL was significantly greater than all three other tasks. This is because this task requires landing from a higher height than the other three tasks. Having a greater peak GRF means that a greater load is placed on the knee joint. Clinicians should be aware of this when developing either prevention or rehabilitation programmes especially in the recovery phase from an injury and in terms of the stiffening of joints on loading. Accordingly, these findings can assist us in ranking four common tasks in order in terms of the greatest load they exert on the knee joint. SLL ranks in highest position, as it places a significantly greater load than all three other tasks. No significant difference was found to occur between the cutting task and running. Finally, lowest load levels were attributed to the SLS, being significantly lower than all three other tasks primarily as this is a less dynamic action.

The results also showed that there were significant differences in the peak knee flexion angle in all pairwise comparisons between tasks ($p < 0.0005$), except between cutting 90 degrees and SLL tasks. Similar results were reported by Donnelly et al. (2012) who compared a cutting task with running, Chinnasee et al. (2018) who compared cutting with landing and Donohue et al. (2015) who compared SLL and SLS. In fact, the peak knee flexion angle in SLS was significantly greater than all three other tasks. This was jointly followed by SLL and cutting tasks. Of all four tasks, the lowest peak knee flexion angle was reported during running, as this task is less complex than the other three and, unlike the others, it does not require additional neuromechanical and dynamic control. Furthermore, there were significant differences in the peak knee extension moment in all pairwise comparisons between tasks, with two exceptions. There was no significant difference in the peak knee extension moment between running and cutting to 90 degrees. In addition, no difference was found to occur in the same variable between running and SLL. These findings are in line with previous studies: Donnelly et al., (2012) who compared the peak knee extension moment between the cutting task and running;

and Kristianslund and Krosshaug (2013) who compared the cutting task and landing. Importantly, the peak knee extension moment in the SLL task was greater than all three other tasks. In fact, the increased peak external knee flexion moment was identified to be a risk factor for non-contact ACL injury in a recent prospective study (Leppänen, Pasanen, Krosshaug et al., 2017b). Therefore, clinicians should be cognisant of this when developing ACL injury prevention or rehabilitation programmes.

In terms of MV for the peak knee valgus angle, post hoc pairwise comparisons showed that there were no significant differences in all comparisons between tasks, with one exception. MV for the peak knee valgus angle during SLL was significantly greater than running ($p = 0.002$). In general, running showed the lowest MV for the peak knee valgus angle between all four tasks. In addition, the results demonstrated that there were significant differences in MV for the peak knee valgus moment in only two pairwise comparisons between tasks. Firstly, between cutting to 90 degrees and running and, secondly, between cutting to 90 degrees and SLL. Interestingly, MV for the peak knee valgus moment during cutting to 90 degrees was significantly lower than running and SLL. However, the peak knee valgus moment was greater than the other tasks. This may indicate that a relationship exists between the peak knee valgus moment and MV for the peak knee valgus moment, which will be investigated further in the next chapter.

Unfortunately, no studies compared MV between these tasks, thus comparisons cannot be drawn between the different study findings. MV is particularly important, as this variable has been identified as a risk factor for ACL injury, as previously highlighted.

The results of this study also showed that there were no significant differences in MV for the peak knee flexion angle in all pairwise comparisons between tasks, with one exception. The MV for the peak knee flexion angle during cutting to 90 degrees was significantly higher than SLS. This is likely because cutting to 90 degrees is regarded as a more complex task than SLS, with the result that there will be higher variability with repeated movements. Additionally, the results have shown that there were significant differences in MV for the peak knee extension moment in the majority of pairwise comparisons between tasks. Only two pairwise comparisons showed no significant differences, namely between cutting to 90 degrees and SLS and between running and SLL. Finally, the results showed that there were significant differences in MV for the peak GRF in the pairwise comparisons that included cutting to 90 degrees. In particular, MV for the peak GRF in this task was significantly greater than occurred in all the other tasks. This could be due to the complexity of this task, as compared to the other three tasks, as it is ranked highest in terms of risk among all four tasks, as previously explained. However, there were no significant differences found in the pairwise comparisons between running, SLL and SLS.

Significant differences in knee biomechanics and MV between tasks could indicate that the type of task and its level of complexity can significantly affect lower limb biomechanics, especially those identified as risk factors for non-contact ACL injury. In addition, it enables us to rank them in order of risk. Such information is important in terms of understanding sporting tasks from both a biomechanical and ACL risk perspective.

### 5.6 Clinical implications

There are several clinical implications in relation to this study. Firstly, clinicians and researchers are not currently aware of MV reference values, as well as the knee kinematics and kinetics associated with the risk of ACL injury among a healthy male population, when conducting biomechanical assessments. In addition, it is important to note that there were no significant differences in either the knee biomechanics or in MV between dominant and non-dominant legs for a healthy male population, during all four common sporting tasks. Consequently, clinicians could deal with both legs in the same way, when assigning exercises. Comparisons between tasks showed that cutting to 90 degrees is significantly different from all three other tasks, as it places the knee joint in a greater valgus pattern. Therefore, it is better to not begin with this sporting task in either prevention or rehabilitation programmes for ACL injuries and instead leave it to a more advanced stage in order to avoid any complications that may occur. Furthermore, of the four tasks, SLL demonstrated the greatest GRF. Having a greater peak GRF means there is an increased load on the knee joint. Clinicians should take this into consideration when developing either prevention or rehabilitation programmes.

## 5.7 Study novelty

As highlighted throughout this chapter, this is a novel study in that it is first of its kind to provide MV reference values for dominant and non-dominant legs in males during four common sporting tasks. In addition, it compares knee kinematics and kinetics between dominant and non-dominant limbs during SLS. Another novel point is that it is first study to compare knee kinematics and kinetics between dominant and non-dominant limbs during running. It also compares knee kinematics and kinetics between cutting to 90 degrees and SLS and between running and SLS. Furthermore, it is first study to compare GRF between four common sporting tasks, as well as to compare MV between dominant and non-dominant legs during four common sporting tasks. The final novel point is that this study is the first to compare MV between four common sporting tasks.

## 5.8 Limitations of this study and future work

The present study has several limitations. Firstly, the findings of this study are only generalisable to males and would not be accurate if applied to females, as significant gender differences in knee biomechanics during these four specific sporting tasks have been reported in several studies: cutting task (Weinhandl, Irmischer, Sievert, & Fontenot, 2017; James et al., 2004; Ford, Myer, Toms, & Hewett, 2005); landing task (Hughes, 2019; Weinhandl, Irmischer, Sievert, & Fontenot, 2017; Holden, Boreham, & Delahunt, 2016; Jenkins et al., 2017); and SLS (Dwyer et al., 2010; Graci et al., 2012; Weeks et al., 2015; Yamazaki et al., 2010; Zeller et al., 2003). A further limitation is that the cutting task findings are solely restricted to cutting to 90 degrees and it would not be accurate to apply them to cutting to different angles, as previous research has reported significant differences when cutting to different angles (Alhammad et al., 2019; Schreurs et al., 2017). The findings of this study are also confined to a healthy population and may not be applicable to unhealthy individuals, as they demonstrate different biomechanical characteristics. Future studies are required to provide reference values for both MV and knee biomechanics associated with risk of ACL injury during these four common sporting tasks in females. New research is also needed to compare knee biomechanics and MV between dominant and non-dominant legs in females during these tasks. Furthermore, at this point new research questions should be developed in order to investigate whether a relationship exists between MV and knee biomechanical risk factors, a topic which is the main focus of the next chapter.

## 5.9 Conclusion

The current study has established reference values for MV and knee kinematics and kinetics associated with the risk of ACL injury in dominant and non-dominant legs among a healthy male population during four common sporting tasks. There were no significant differences in either MV or in knee biomechanics between dominant and non-dominant legs during all four tasks. However, significant differences have been found to occur in the between-task pairwise comparisons for most of the biomechanical variables and for a number of the MV variables. In addition, of all four tasks, cutting to 90 degrees demonstrated the highest values in terms of knee biomechanical risk factors. However, this same task demonstrated lower MV, as compared to the other tasks.

Having conducted our investigations, we have gained a full understanding of MV and knee biomechanics, as we have now obtained reference values in a healthy population. However, new research questions need to be developed, for example, in order to investigate whether a relationship exists between MV and knee biomechanical risk factors, a topic which will be the main focus of the next chapter. Another research question needs to investigate MV and knee biomechanics in other populations in relation to ACL injury (ACL-RC/ACL/DF), an area which will also be explored in this thesis.

# Chapter 6

## 6 Relationship Between Movement Variability and Knee Biomechanical Risk Factors of ACL Injury During Four Common Sporting Tasks

### 6.1 Introduction

This chapter seeks to investigate whether a relationship exists between MV and knee biomechanical risk factors. As highlighted in the previous chapter, it is important to conduct such an investigation, as it will enhance our knowledge of MV and ACL biomechanical risk factors. Intra-individual variation between repeated movement patterns, such as a landing task, is regarded as comprising a core component of all motor tasks (König et al., 2016). Kantz and Schreiber (2004) have classified variability into two broad types, the first of which comprises noise resulting from measurement error, while the second describes variation arising from systems-related intrinsic dynamics. Motor control and biomechanics research contends that measurement noise primarily leads to variability. However, according to Glass (2001), the functional dimensions of variability associated with nonlinear systems have begun to attract increasing attention in recent years, following advances in nonlinear dynamics and chaos theory (Glass, 2001).

It is very important to investigate the relationship between MV and knee biomechanics in order to understand the role of MV in biomechanical risk factors for ACL injury. Importantly, a number of studies (for example, James, 2004; Konradsen, 2002; Konradsen & Voigt, 2002) have suggested that MV may play a role in sustaining injury. Furthermore, earlier research by McLean et al. (1999) has suggested that a relationship exists between MV and increased risk of acquiring musculoskeletal injuries, which includes ACL rupture. The previous chapter investigated MV from the perspective of providing reference values, as well as conducting comparisons both between the two legs and between the four different sporting tasks. This chapter will investigate the relationship between MV and biomechanical risk factors for non-contact ACL injury. Such investigations will help clinicians and researchers to establish whether a relationship exists between MV and knee biomechanical risk factors for ACL.

## 6.2 Null hypothesis

H0: There is no significant relationship between biomechanical risk factors for ACL injury and MV.

## 6.3 Methods

### 6.3.1 Participants

No study to date has investigated the relationship between MV and the biomechanical risk factors associated with ACL injury. Therefore, power analysis was performed using post hoc G*Power (Version 3.1.9.3) based on the peak knee valgus moment obtained in this study (Es = 0.48). The sample size (N=45) drawn upon in this research showed a statistical power of 71%, according to G*Power post hoc analysis, with the type 1 error alpha level set at 5% ($\alpha$ = 0.0) (Faul, Erdfelder, Buchner, & Lang, 2009) (See appendix 12.5). Participant demographics are outlined in Table 6-1. The inclusion/exclusion criteria and participant characteristics were explained in detail in Chapter 4 (Section 4.12.4.2). All participants completed a consent form prior to engaging in the testing process.

*Table 6-1: Participants' demographic profiles*

| | Number (Gender) | Mean | SD |
|---|---|---|---|
| Age (years) | 45 (male) | 25.8 | 6.6 |
| Height (m) | 45 (male) | 1.71 | 0.06 |
| Body mass (kg) | 45 (male) | 65.7 | 10.3 |

### 6.3.2 Preparation and tasks undertaken

All participants were required to do some general warm-up exercises of ten minutes' duration, followed by a further five-minute warm-up on an exercise bike, prior to beginning the test. The preparation process was explained in detail in Chapter 4 (Section 4.5). In terms of the tasks undertaken, all participants were invited to complete all four functional sporting tasks: SLS, running, cutting to 90 degrees and SLL with their two legs (both dominant and non-dominant). Details of the requirements involved in performing these tasks were outlined in Chapter 4 (Section 4.5.1 for running, Section 4.5.2 for SLS, Section 4.5.3 for SLL and Section 4.5.4 for cutting to 90 degrees).

### 6.3.3 Procedure

The biomechanical data were collected for this study using 3D motion analysis at two different laboratories. First, the 3D motion analysis laboratory at the University of Salford which was described in details in Chapter 4 (Section 4.2.1). Second, the 3D motion analysis laboratory at Imam Abdulrahman Bin Faisal University which was described in details in Chapter 4 (Section 4.2.2).

### 6.3.4 Outcome measures

This chapter includes all outcome measures, which were explained in Chapter 4 (Section 4.9).

### 6.3.5 Statistical analysis

All statistical analysis in this study was carried out using the Statistical Package for the Social Sciences (SPSS) (Version 24). Descriptive analysis (means and standard deviations) were provided. Prior to undertaking data analysis, the normality distribution of the data was observed visually following application of a Shapiro-Wilk test. For the purposes of this test, a variable is normally distributed if the p value is greater than 0.05; however, it is not normally distributed if the p value is equal to or less than 0.05. Establishing whether an association exists between the variables was conducted using Pearson's correlation (r) for the normally distributed variables. The strength of the correlation was determined based on the general guidelines provided by Cohen (2013), as shown in Table 6-2. However, the Spearman rank-order correlation coefficient ($r_s$) was used for the non-normally distributed variables. If the value of $r_s$ equals zero, this indicates that no association exists. Unlike Pearson's correlation, Spearman's correlation provides no guidelines for determining the strength of the association. However, the closer the correlation coefficient is to 0, the weaker the correlation is between the ranks and the closer the correlation coefficient is to +1 or -1, the stronger the correlation is between the ranks. For both statistical measures, the significance level was set at 5% in order to determine whether the correlations were statistically significant (significant correlation when the p-value is $< 0.05$, non-significant correlation when the p-value is $> 0.05$).

In addition, in order to decrease the risk of a type 1 error occurring, the Bonferroni correction was applied at the alpha level ($\alpha = 0.05$) for all correlation tests carried out, as there were multiple correlations conducted in this study (corrected $\alpha = 0.01$). The effect size for each correlation was also provided, along with a p value, in order to fully assess the results. Table 6-2 demonstrates the levels of strength of association (Cohen, 2013). All analysis was conducted using data from the dominant leg for all variables, as participants demonstrated no significant differences between the two legs, as explained in the previous chapter. The correlation test was performed for each biomechanical variable and MV on the same variable. In addition, the analysis was conducted for each task independently due to the significant differences between the activities, as reported in the previous chapter.

*Table 6-2: Levels of strength of association*

| Correlation coefficient value (r) | Strength of association |
|---|---|
| $0.1 < \| r \| < 0.3$ | Small correlation (weak) |
| $0.3 < \| r \| < 0.5$ | Medium correlation (moderate) |
| $\| r \| > 0.5$ | Large correlation (strong) |

(Cohen, 2013)

## 6.4 Results

### 6.4.1 Overground running

A correlation test was conducted for each knee biomechanical variable and MV for the same variable during the running task. The results of all correlation tests are shown in Table 6-3. In general, this task demonstrated one significant correlation. A significantly strong negative correlation occurred between the peak knee valgus moment and MV for the peak knee valgus moment ($r_s = -0.50$, $p < 0.0005$, Es = 0.25) where an increase in MV may lead to a decrease in the peak knee valgus moment. In addition, the scatter plots for all significant correlations are illustrated in Figure 6-1.

*Table 6-3: Correlations between the knee biomechanics (peaks) and MV during the running task.*

| **Variables** | **Correlation coefficient (r) / ($r_s$)** | **P value** | **Effect size** |
|---|---|---|---|
| Knee flexion angle & MV for knee flexion angle | -0.38 | 0.02 | 0.15 |
| Knee extension moment & MV for knee extension moment | -0.29 | 0.05 | 0.09 |
| Knee valgus angle & MV for knee valgus angle | -0.06 | 0.70 | 0.003 |
| Knee valgus moment & MV for knee valgus moment | -0.50 | < 0.0005*† | 0.25 |
| V-GRF & MV for V-GRF | 0.23 | 0.14 | 0.05 |

** Significant difference following Bonferroni correction ($p \leq 0.01$). Negative (-) indicates there is a negative relationship. † (< 0.0005) indicates that the SPSS reading was 0.000.*

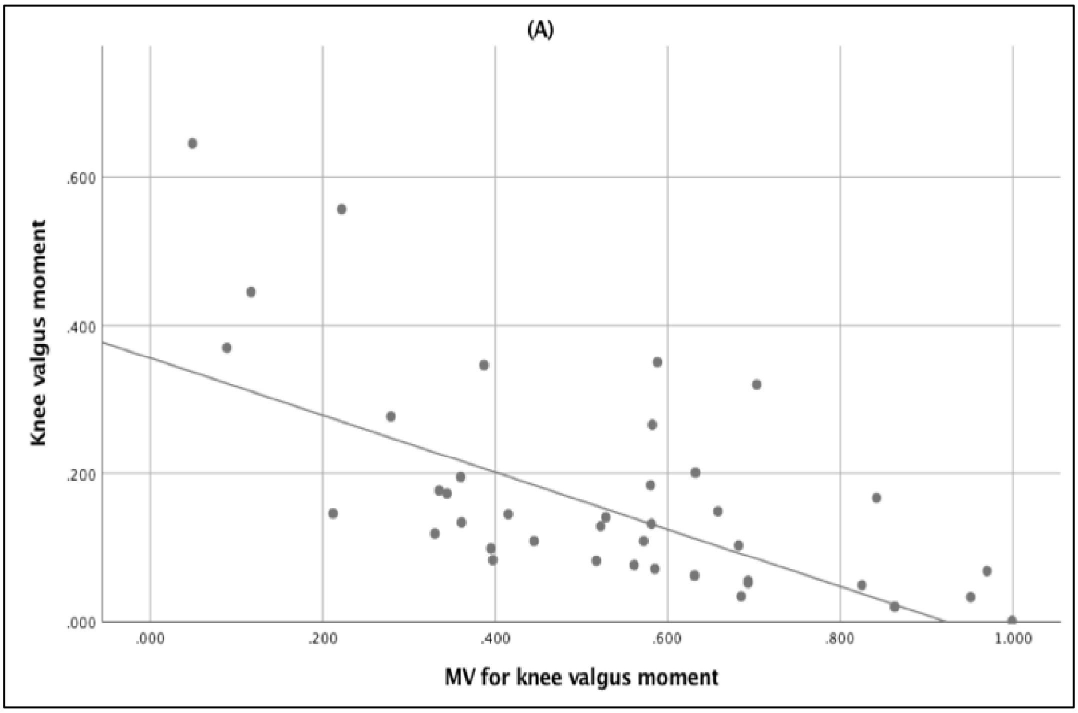



*Figure 6-1: Scatter plot for the significant correlation between the peak knee valgus moment and MV for the peak knee valgus moment during the running task..*

### 6.4.2 Cutting to 90 degrees

An association test was undertaken for each knee biomechanical variable and MV for the same variable during the cutting to 90 degrees task. The results of all correlation tests are outlined in Table 6-4. Generally, three significant correlations were found to occur during this task. Firstly, there was a significantly strong negative correlation between the peak knee valgus angle and MV for the peak knee valgus angle ($r_s$ = -0.71, p < 0.0005, Es = 0.50) meaning that an increase in MV can lead to a decrease in the peak knee valgus angle. Secondly, there was a significant correlation between the peak knee valgus moment and MV for the peak knee valgus moment, and it was a moderately negative relationship ($r_s$ = -0.43, p = 0.003, Es = 0.18) meaning that an increase in MV can lead to a decrease in the peak knee valgus moment. The third significant correlation was found to occur between the peak knee extension moment and MV for the peak knee extension moment, and it was a strong negative association ($r_s$ = -0.65, p < 0.0005, Es =

0.42), where an increase in MV can lead to a decrease in the peak knee extension moment. Furthermore, the scatter plots for all significant correlations are illustrated as follows: Figure 6-2 for the *correlation between the peak knee valgus angle and MV for the peak knee valgus angle*, Figure 6-3 for the *correlation between the peak knee valgus moment and MV for the peak knee valgus moment and* Figure 6-4 for the *correlation between the peak knee extension moment and MV for the peak knee extension moment.*

*Table 6-4: Correlations between the knee biomechanics (peaks) and MV during the cutting to 90 degrees task.*

| **Variables** | **Correlation coefficient (r) / ($r_s$)** | **P value** | **Effect size** |
|---|---|---|---|
| Knee flexion angle & MV for knee flexion angle | -0.30 | 0.05 | 0.09 |
| Knee extension moment & MV for knee extension moment | -0.65 | < 0.0005*† | 0.42 |
| Knee valgus angle & MV for knee valgus angle | -0.71 | < 0.0005* | 0.50 |
| Knee valgus moment & MV for knee valgus moment | -0.43 | 0.003* | 0.18 |
| V-GRF & MV for V-GRF | 0.13 | 0.41 | 0.02 |

** Significant difference following Bonferroni correction ($p \leq 0.01$). Negative (-) indicates there is a negative relationship. † (< 0.0005) indicates that the SPSS reading was 0.000.*

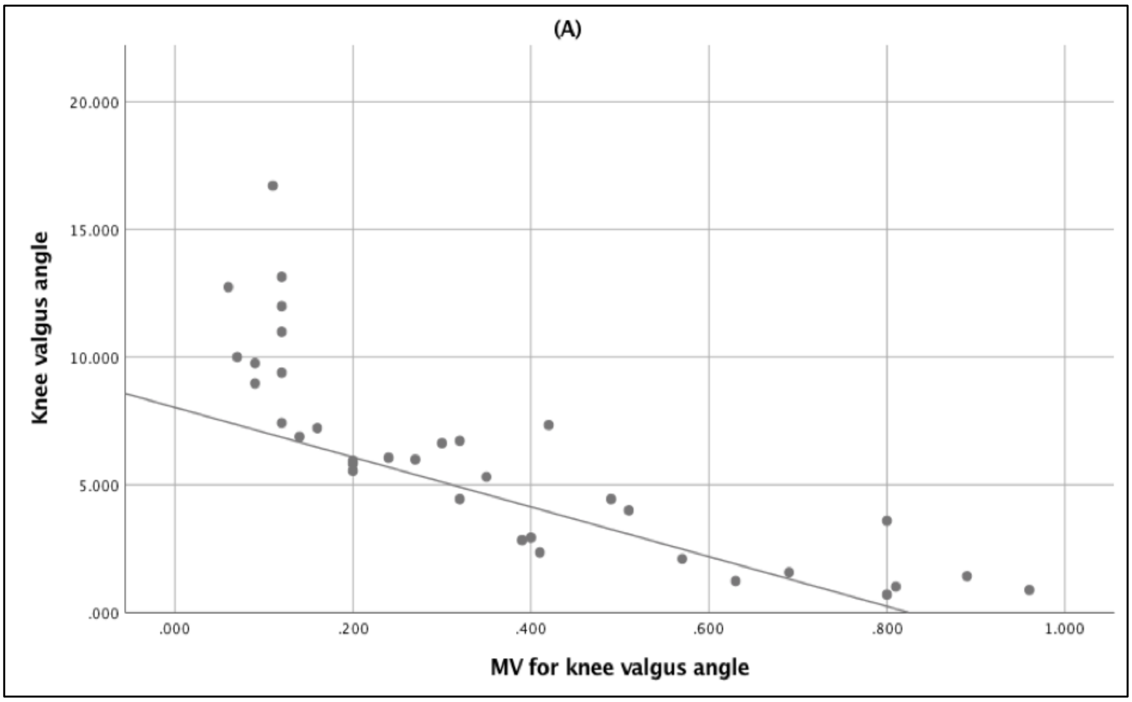



*Figure 6-2: Scatter plot for the significant correlation between the peak knee valgus angle and MV for the peak knee valgus angle during the cutting to 90 degrees task.*

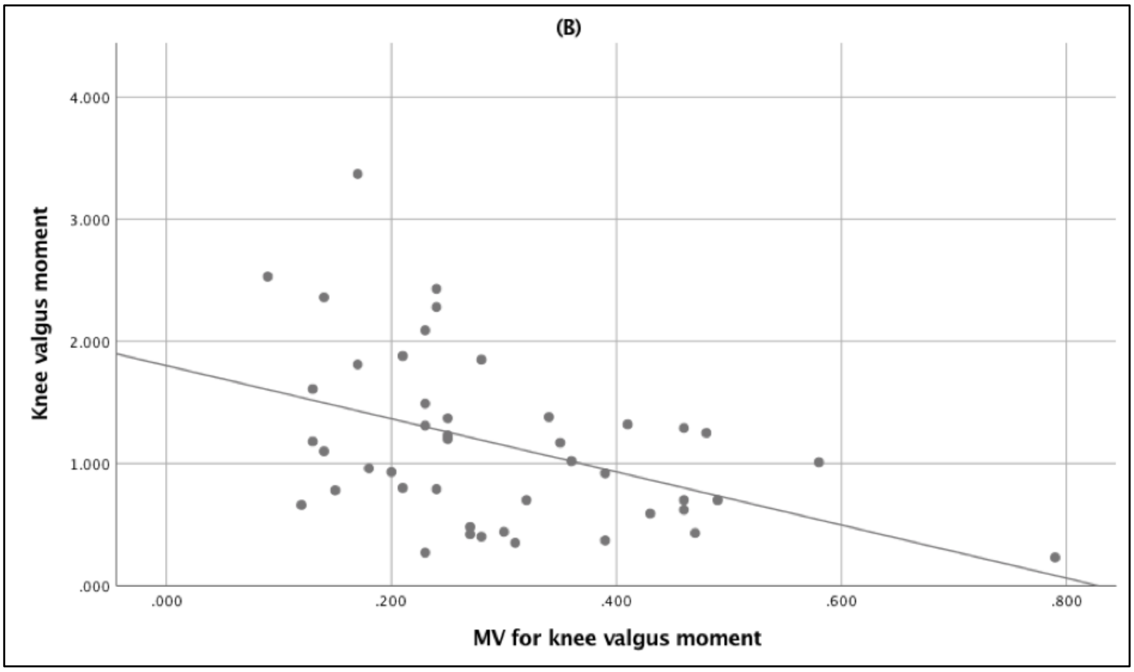



*Figure 6-3: Scatter plot for the significant correlation between the peak knee valgus moment and MV for the peak knee valgus moment during the cutting to 90 degrees task.*

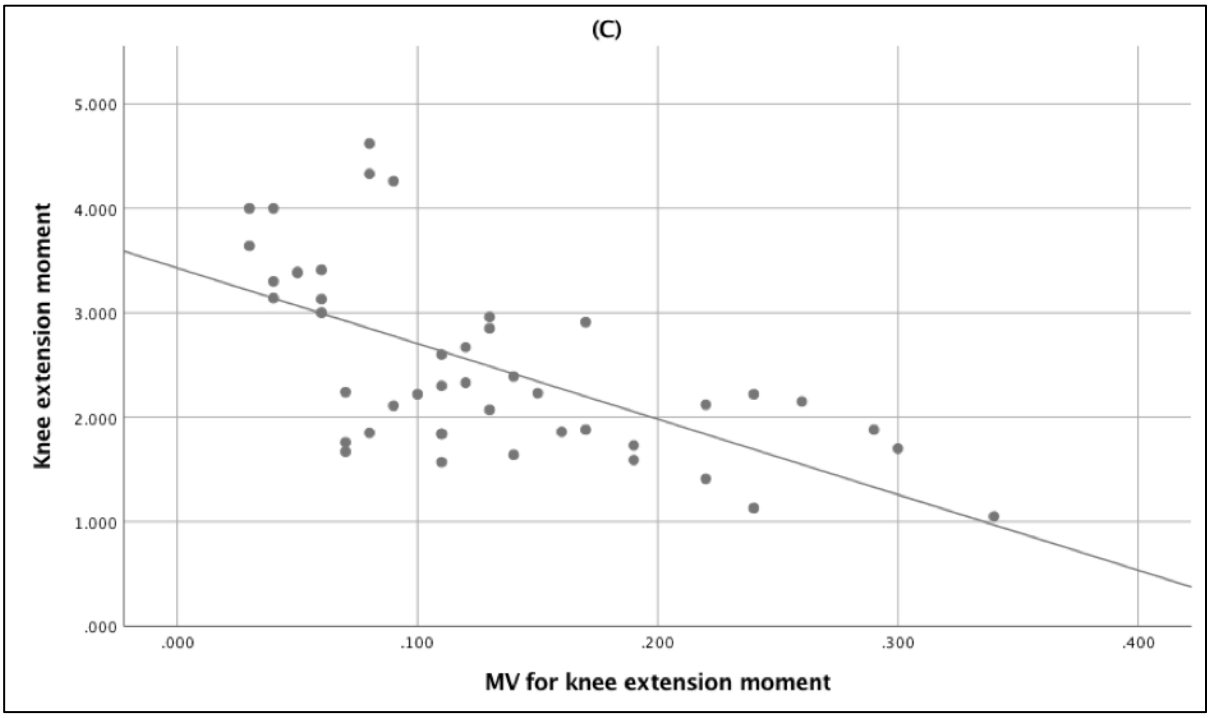



*Figure 6-4: Scatter plot for the significant correlation between the peak knee extension moment and MV for the peak knee extension moment during the cutting to 90 degrees task.*

### 6.4.3 SLL

A correlation test was applied for each knee biomechanical variable and MV for the same variable during SLL. The results of all correlation tests are presented in Table 6-5. Overall, similar to the cutting task, two significant associations were reported during SLL. Firstly, there was a significantly strong negative correlation between the peak knee valgus moment and MV for the peak knee valgus moment ($r_s$ = -0.69, p < 0.0005, Es = 0.48) meaning that an increase in MV can lead to a decrease in the peak knee valgus moment. The second significant correlation during this task was found to occur between the peak GRF and MV for the peak GRF, and it was a strong positive correlation ($r_s$ = 0.79, p < 0.0005, Es = 0.63) meaning that an increase in MV can lead to an increase in the peak knee valgus moment. In addition, the scatter plots for these two significant correlations are illustrated in two figures, Figure 6-5 for the correlation between the peak knee valgus moment and MV for the peak knee valgus moment and Figure 6-6 for the correlation between the peak vertical GRF and MV for the peak vertical GRF.

*Table 6-5: Correlations between the knee biomechanics (peaks) and MV during SLL.*

| **Variables** | **Correlation coefficient (r) / ($r_s$)** | **P value** | **Effect size** |
|---|---|---|---|
| Knee flexion angle & MV for knee flexion angle | -0.06 | 0.71 | 0.003 |
| Knee extension moment & MV for knee extension moment | -0.15 | 0.34 | 0.02 |
| Knee valgus angle & MV for knee valgus angle | -0.02 | 0.91 | 0.003 |
| Knee valgus moment & MV for knee valgus Moment | -0.69 | < 0.0005* | 0.48 |
| V-GRF & MV for V-GRF | 0.79 | < 0.0005* | 0.63 |

** Significant difference following Bonferroni correction ($p \leq 0.01$). Negative (-) indicates there is a negative relationship. † (< 0.0005) indicates that the SPSS reading was 0.000.*

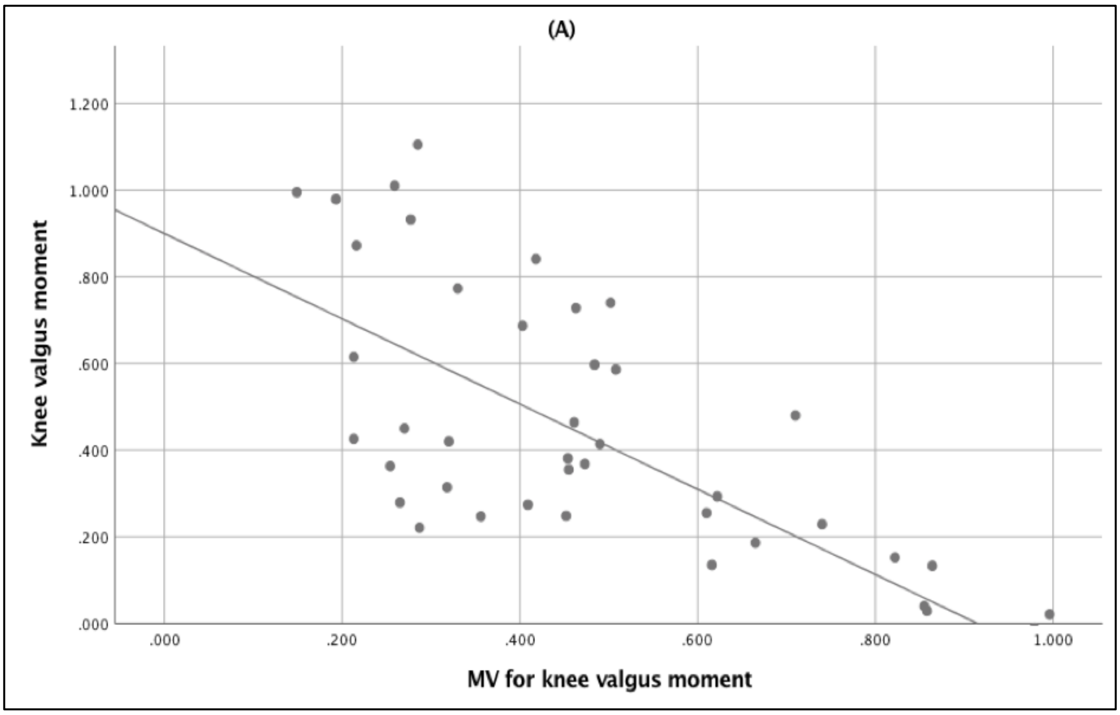



*Figure 6-5: Scatter plot for the significant correlation between the peak knee valgus moment and MV for the peak knee valgus moment during SLL task.*

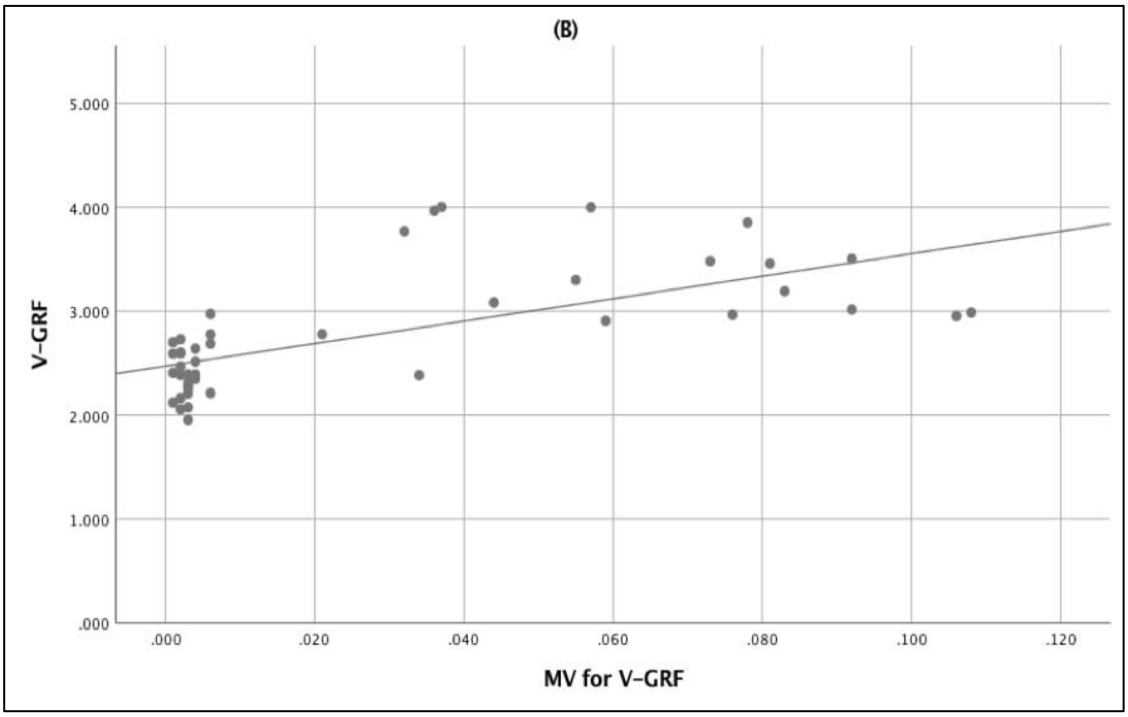



*Figure 6-6: Scatter plot for the significant correlation between the peak vertical GRF and MV for the peak vertical GRF during the SLL task.*

### 6.4.4 SLS

A correlation test was conducted for each knee biomechanical variable and MV for the same variable during SLS. The results of all correlation tests are shown in Table 6-6. In line with cutting to 90 degrees and SLL, two significant correlations were found to occur during the SLS task. Firstly, there was a significantly strong negative correlation between the peak knee valgus moment and MV for the peak knee valgus moment ($r_s$ = -0.76, p < 0.0005, Es = 0.57). The second significant correlation occurred between the peak GRF and MV for the peak GRF, and it was a strong positive correlation ($r_s$ = 0.64, p < 0.0005, Es = 0.41). Furthermore, the scatter plots for the two significant correlations are presented as follows; Figure 6-7 for the correlation between the peak knee valgus moment and MV for the peak knee valgus moment and Figure 6-8 for the correlation between the peak vertical GRF and MV for the peak vertical GRF.

*Table6-6: Correlations between the knee biomechanics (peaks) and MV during SLS.*

| Variables | Correlation coefficient (r) / ($r_s$) | P value | Effect size |
|---|---|---|---|
| Knee flexion angle & MV for knee flexion angle | -0.28 | 0.06 | 0.08 |
| Knee extension moment & MV for knee extension moment | -0.24 | 0.11 | 0.06 |
| Knee valgus angle & MV for knee valgus angle | 0.22 | 0.18 | 0.11 |
| Knee valgus moment & MV for knee valgus Moment | -0.76 | < 0.0005* | 0.57 |
| V-GRF & MV for V-GRF | 0.64 | < 0.0005* | 0.41 |

** Significant difference following Bonferroni correction (p ≤ 0.01). Negative (-) indicates there is a negative relationship. † (< 0.0005) indicates that the SPSS reading was 0.000.*

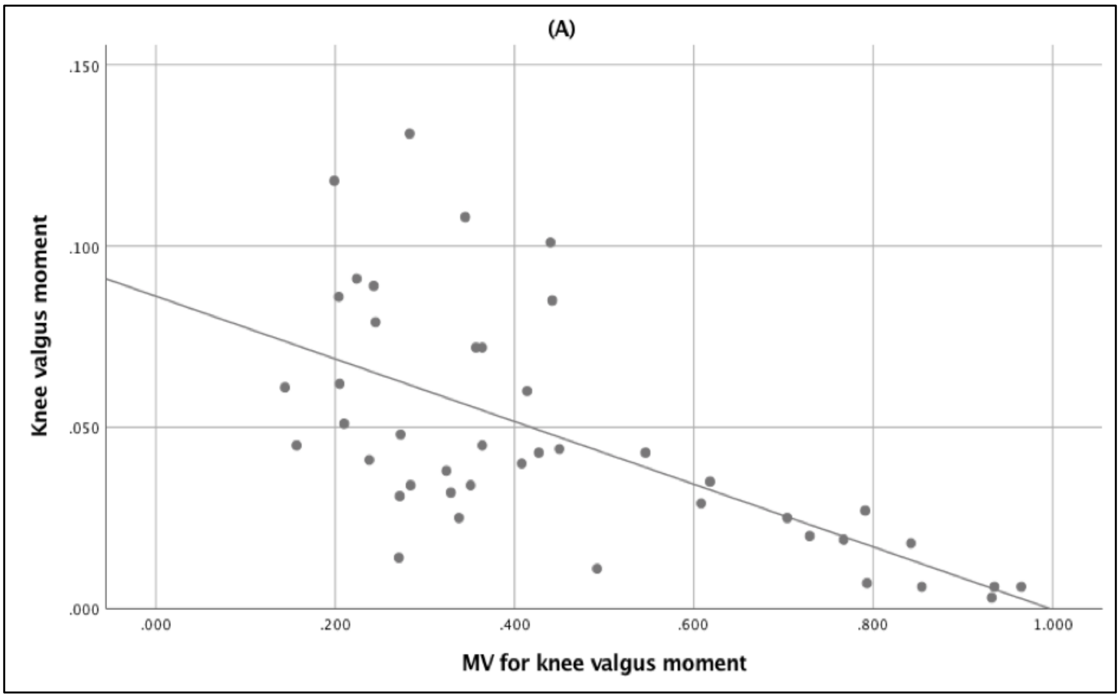



*Figure 6-7: Scatter plot for the significant correlation between the peak knee valgus moment and MV for the peak knee valgus moment during SLS task.*

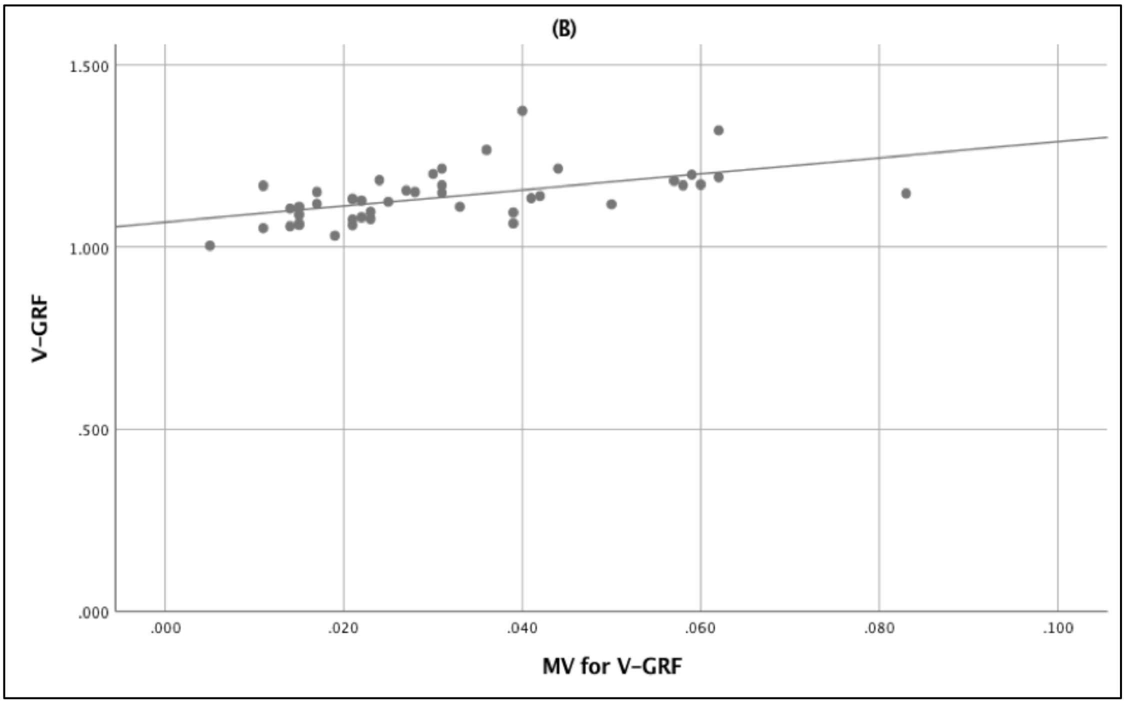



*Figure 6-8: Scatter plot for the significant correlation between the peak vertical GRF and MV for the peak vertical GRF during the SLS task.*

## 6.5 Discussion

The aim of this study was to investigate whether any relationship exists between MV and knee biomechanical risk factors for ACL injury. It has been suggested that MV may be an element that could contribute to developing and perpetuating injuries (Brown et al., 2009; Konradsen, 2002; Konradsen & Voigt, 2002). It has also been proposed that MV could be associated with the risk of developing musculoskeletal injuries, including ACL injuries (McLean et al., 1999). The findings of this study have shown that MV is significantly correlated with knee biomechanical risk factors for ACL injury. In general, the results demonstrated that there was a significant negative correlation between the peak knee valgus moment and MV for the peak knee valgus moment during all tasks. This type of correlation implies that if MV increases, then the knee valgus moment decreases. This suggests that MV could play a role in ACL injury. The results also showed that cutting to 90 degrees is the only task in which a significant negative correlation was reported in both the knee valgus angle and the knee valgus moment, which are the most common biomechanical risk factors for ACL injury (Hewett et al., 2005; Myer et al., 2015; Numata et al., 2018; McLean, Lipfert, & Van Den Bogert, 2004). Such results, in cases where MV is associated with more than one risk factor, indicates that the former could be a very important element in contributing to this injury during cutting to 90 degrees task. In particular, of all four tasks, cutting to 90 degrees is rated most risky, as reported in the previous chapter.

The strength of the correlation is very important, along with the effect size, as collectively they can provide a very clear insight into the nature of the relationship. All correlations reported in the frontal plane (mentioned above) were strong, accompanied by a large effect size, with one exception. A moderate correlation was found to exist between the peak knee valgus moment and MV for the peak knee valgus moment, during the cutting task, with a small effect size ($r_s$

= -0.43, p = 0.003, Es = 0.18). However, during the same task, the correlation was strong between the peak knee valgus angle and MV for the peak knee valgus angle, with a large effect size ($r_s$ = -0.71, p < 0.0005, Es = 0.50).

In terms of the sagittal plane, the results show that there were significant correlations between MV and the sagittal plane kinetics in cutting to 90 degrees task. In particular, it was a strong negative association. This suggests that an increase in MV can lead to a decrease in the peak knee extension moment. The sagittal plane variables have recently been identified as risk factors for ACL injury (Leppänen et al., 2017a). The findings in the sagittal plane are in line with those of the frontal plane, where it showed that a negative relationship occurred between MV and the biomechanical risk factors for ACL injury. Furthermore, However, the other three tasks did not report significant correlations between the peak knee extension moment and MV for the peak knee extension moment.

With regard to GRF, the results reported that there was a significant correlation between MV and the peak V-GRF during the two tasks. Firstly, there was a significant positive correlation between the peak GRF and MV for the peak GRF during SLL. This was also found to occur during SLS, where a strong positive correlation occurred ($r_s$ = 0.79, p < 0.0005, Es = 0.63). However, there was no significant correlation between the peak GRF and MV for the peak GRF during cutting and running tasks. When comparing correlations in the GRF with those in the frontal and sagittal planes, they differed in that it was positive between MV and the GRF, whereas it was negative in the frontal and sagittal planes. Another point of note is that a significant correlation emerged between the GRF and MV during only two tasks; however, this correlation was found to occur in the sagittal and frontal planes during all tasks.

In the literature, a number of studies in the field of biomechanics and motor control have demonstrated that MV has a functional role (Hamill et al., 2012). In addition, it has been shown that MV is crucial for coordinative changes in gait and in bimanual coordination (Scholz, Kelso, & Schöner, 1987; Seay et al., 2006; Emmerik & Wagenaar, 1996). Lipsitz, (2002) advanced the hypothesis known as 'loss of complexity hypothesis', suggesting that a lack of variability is one of the characteristics associated with disease, dysfunction or weakness. In terms of injuries, Hamill et al. (2012) emphasised the functional role of variability and linked it to overuse injury, which was derived from a dynamical systems perspective (Hamill et al., 1999), as explained in Chapter 2 (see Section 2.11.4.3). They stated that 'the higher variability state is the healthy state while the lower variability state is the unhealthy or pathological state'. Based on our findings, this indicates that increased MV leads to a decrease in ACL biomechanical risk factors. This supports the notion that higher variability is associated with a healthy state as it leads to lower risk, while lower variability correlates with an unhealthy state, thus resulting in higher risk. However, as highlighted by Hamill et al. (2012), very high variability levels could be harmful and may cause an injury. Therefore, it is important to have a window of 'higher variability' in which healthy individuals can function. This clearly highlights the importance of obtaining MV reference values. Furthermore, these reference values, along with their standard deviations, are crucial in gaining an understanding of the upper and lower parameters of MV in healthy individuals, which fortunately were obtained in the previous chapter.

In general, and aside from the above mentioned studies, some researchers are of the belief that MV requires minimisation or removal due to it creating a negative side-product of random noise on the central nervous system (Faisal et al., 2008; Harris & Wolpert, 1998; Van Beers et al., 2004). Other researchers, however, have advanced an alternate view, namely, rather than being a product of undesired noise or error in the movement system, variability in motor output is actually beneficial to this system, as it endows it with greater flexibility and adaptability (Newell & James, 2008; Riley & Turvey, 2002; Slifkin & Newell, 1998). The findings of the current study align with the second premise, where increased MV may lead to lower risk of sustaining an ACL injury.

In fact, despite the results emerging from this study, there is a lack of knowledge as to whether decreased MV is a risk factor for non-contact ACL injury. Unfortunately, as no study to date has investigated the relationship between MV and ACL injury, comparisons cannot be drawn between our findings and others. Hence, this is the first study to investigate this issue. In addition, there is very limited evidence currently available regarding the role MV plays in sport injuries in general.

### 6.6 Clinical implications

A number of clinical implications have been identified in this study. Firstly, clinicians and researchers should be cognisant of the relationship between MV and knee biomechanical risk factors for ACL injury. Most importantly, they should be aware that a negative relationship exists between both factors, whereby if MV increases then the biomechanical risk factors decrease. This reinforces the point that increased MV could be a primary goal when developing ACL injury risk mitigation programmes and returning to sport following ACL reconstruction and rehabilitation. However, this study proposes that MV should remain within the average reference values recommended for healthy individuals in the previous chapter.

## 6.7 Study novelty

The findings highlighted in the discussion section of this chapter underline that this study is the first of its kind to establish that a relationship exists between MV and knee biomechanical risk factors for ACL injury during common sporting tasks.

## 6.8 Limitations of this study and future work

The current study has several limitations. Firstly, the investigation was confined to a male-only cohort. Furthermore, it solely focused on a healthy population group. Future studies need to be conducted on females in order to investigate whether similar results will emerge. New research is also needed to establish whether clinical assessments would help clinicians to assess MV within clinical practice, as it is not possible to conduct 3D motion analysis in every clinical assessment, due to both the cost and time constraints associated with usage of this method.

## 6.9 Conclusion

The current study has established that a relationship exists between MV and knee kinematics and kinetics, which is associated with the risk of ACL injury during common sporting tasks. A significant negative relationship was found to occur between MV and peak knee valgus moment during all four functional sporting tasks. In addition, a significant negative relationship emerged between MV and the peak knee valgus angle during a cutting to 90 degrees task. In terms of the sagittal plane, a significant difference between MV and the peak knee extension moment was reported during cutting to 90 degrees. All these relationships are negative, which is indicative that an increase in MV could lead to a decrease in these variables. MV should be considered when developing ACL injury risk mitigation and ACL reconstruction rehabilitation RTS programmes.

Having completed our investigations, we have gained a full understanding of the relationship between MV and knee biomechanics. However, as highlighted in the previous two chapters, other research questions are warranted in order to examine MV and knee biomechanics in other populations in relation to ACL injury (ACL-RC/ACL/DF). The latter will be also investigated in the next chapter.

# Chapter 7

## 7 Movement Variability and Knee Biomechanics in ACL Deficient Individuals During Four Common Sporting Tasks

The two previous chapters answered very important research questions for this thesis, as they investigated MV and knee biomechanics in a healthy population, while also providing reference values and undertaking comparisons between the two legs, as well as between tasks. Furthermore, the relationship between MV and knee biomechanics was established. This chapter now seeks to conduct further investigations on MV and knee biomechanics on another ACL injury-affected population, namely ACL deficient individuals.

### 7.1 Introduction

No consensus currently exists within published research to identify the optimal way in which to manage ACL rupture, as a result of injury (Smith et al., 2014). Some researchers, such as Kwok, Harrison and Servant (2012), recommended early ACL reconstruction in order to restore normal tibiofemoral joint kinematics and decrease the risk of joint instability. Thus, they have proposed that this approach reduces the likelihood of secondary joint damage occurring, as well as susceptibility to osteoarthritis. In addition, Kwok et al. (2012) have argued that surgical intervention is the sole method by which to ensure that individuals will gain sufficient stability in order to actively engage in pivoting sporting tasks. Conversely, other researchers, such as Toby, Smith, Davies and Hing (2010), have contended that stringent neuromusculoskeletal rehabilitation programmes aid recovery within this specific cohort and would not heighten the risk of degenerative change in the longer term. In addition, surgical interventions are associated with a number of adverse outcomes, most notably, arthrofibrosis, infection, graft failure, pain, donor site morbidity and additional costs, as opposed to the adoption of other non-surgical forms of treatment.

Having the capacity to return to their preinjury level of activities, including sporting activities, is a very important issue for ACL-injured individuals. A recent study by Keays, Newcombe and Keays (2018) investigated return to sport (RTS) levels in ACL-DF individuals who underwent conservative treatment, which included a 12-year follow-up. They found that 89% of ACL-DF patients who underwent conservative treatment were able to engage in an active sporting life. One-third of the ACL-DF cohort had returned to pivoting sports. In the literature, the majority of biomechanical studies have investigated biomechanics in ACL-DF individuals during walking (Ismail et al., 2015; Ferber et al., 2002; Hart, Ko, Konold, & Pietrosimione, 2010; Lewek et al., 2002; Roberts et al., 1999; Rudolph et al., 2001; Chmielewski et al., 2001) and running (Waite et al., 2005; Lewek et al., 2002; Rudolph et al., 2001; Chmielewski et al., 2001). However, few studies are available in relation to biomechanics in individuals with ACL-DF during functional sporting tasks, including cutting (Waite et al., 2005) and SLS tasks (Yamazaki et al., 2010). As a large number of ACL patients choose not to undertake ACL-RC and return to sport following conservative treatment, it is important to carry out biomechanical investigations on them, similar to those conducted on healthy individuals. In fact, such investigations will help to identify the biomechanical characteristics associated with this cohort. Accordingly, it will help to develop the most appropriate rehabilitation programme for patients who are considering returning to sport. Therefore, this study aims to investigate MV and knee biomechanics in ACL-DF individuals. In addition, it seeks to investigate whether significant differences exist in MV and knee biomechanics between the ACL-DF knee and the uninjured knee during four common sporting tasks.

## 7.2 Null hypothesis

1- $H0_1$: There are no significant differences in knee kinematics and kinetics between the ACL-DF knee and the uninjured knee during four common sporting tasks.

2- $H0_2$: There are no significant differences in MV between the ACL-DF knee and the uninjured knee during these four tasks.

## 7.3 Methods

### 7.3.1 Participants

Power analysis was performed using G*Power (Version 3.1.9.3) based on the peak knee flexion angle obtained in a previous study by Waite et al. (2005) who investigated knee biomechanics in ACL-DF individuals during cutting task (Es = 0.66). The rationale for selecting this variable is because it is the main variable affected by this injury, as evidenced in most of biomechanical studies conducted with this population cohort, during various tasks and will be discussed later in this chapter. Furthermore, Waite et al. (2005) is the only study that has included the cutting task with an ACL-DF population. The effect size of 0.66 was used to calculate the sample size, and it was calculated electronically using G*Power software. A minimum of 16 participants was required in order to attain 80% statistical power with a type 1 error alpha level of 5% ($\alpha$ = 0.05) for such investigations (Faul et al., 2007). (See appendix 12.6). A total of 20 ACL-DF individuals participated in this study which accounted for any potential drop-out and/or loss of data. The participants' demographic profile is presented in Table 7-1. They all completed a consent form prior to engaging in the testing process.

Participants on the ACL-DF group were included on the basis that they had a unilateral, total and isolated rupture of the ACL, which had been diagnosed by a clinician through the use of magnetic resonance imaging (MRI) techniques and clinical examinations. Additionally, this group have undergone and completed a supported and structured programme of rehabilitation under supervision. They also have demonstrated a mean time from the initial ACL rupture of greater than 6.0 months and returning levels of fitness to within their normal pre-injury levels with no notable changes to their exercise programme. Furthermore, participants in this group have reported <=1.0 incidents of 'giving way' following their initial injury. They have not reported any secondary complications as a result of their injury, including episodes of recurring bouts of effusion during exercise, any pain or worrying clinical signs and symptoms related to knee joint OA at the time of the initial injury or when signing up for the tests. Finally, anterior drawer and pivot shift tests were performed on every participant to examine the function of ACL.

*Table 7-1: Participants' demographic profiles*

| | Number (Gender) | Mean | SD |
|---|---|---|---|
| Age (years) | 20 (male) | 25.8 | 5.0 |
| Height (m) | 20 (male) | 1.72 | 0.06 |
| Body mass (kg) | 20 (male) | 71.6 | 11.4 |

### 7.3.2 Preparation and tasks undertaken

Each participant was required to do some general warm-up exercises of ten minutes' duration, followed by a further five-minute warm-up on an exercise bike, prior to beginning the test. The preparation process was explained in detail in Chapter 4 (Section 4.5). In terms of the tasks undertaken, all participants were invited to complete all four functional sporting tasks: SLS, running, cutting to 90 degrees and SLL with their two legs (both ACL-DF and uninjured). Details of the requirements involved in performing these tasks were outlined in Chapter 4 (Section 4.5.1 for running, Section 4.5.2 for SLS, Section 4.5.3 for SLL and Section 4.5.4 for cutting to 90 degrees).

### 7.3.3 Procedure

The biomechanical data were collected for this study using 3D motion analysis at the 3D motion analysis laboratory at Imam Abdulrahman Bin Faisal University which was described in details in Chapter 4 (Section 4.2.2).

### 7.3.4 Outcome measures

This chapter includes all outcome measures, as explained in Chapter 4 (Section 4.9).

### 7.3.5 Statistical analysis

All statistical analysis in this study was carried out using the Statistical Package for the Social Sciences (SPSS) (Version 24). Descriptive analysis (means and standard deviations) was performed. Prior to conducting the data analysis, its normality distribution was observed visually following application of a Shapiro-Wilk test. A variable is normally distributed if the p-value is greater than 0.05 on the latter test; however, it is not normally distributed if the p-value is equal or less than 0.05. The differences between injured and non-injured legs were tested using a paired t-test for parametric variables (normally distributed) and a Wilcoxon Rank test for nonparametric (not normally distributed). The significance level was set at 5% (significant level when the p-value $< 0.05$, non-significant level when the p-value $> 0.05$).

In addition, in order to decrease the risk of type 1 errors occurring, Bonferroni correction was applied to the alpha levels used ($\alpha = 0.05$) for all correlation tests carried out, as there were multiple comparisons conducted in this study (corrected $\alpha = 0.01$). The effect size for each comparison was also provided, along with p-values, in order to be able to fully interpret the results. The effect size was calculated using Cohen's d Index (Cohen, 2013). Table 7-2 demonstrates the levels and classifications of effect size according to Cohen (2013). In terms of MV values, a zero value signals that no variability has occurred while, conversely, a score of one is indicative of the highest levels of variability.

*Table 7-2: Levels and classification of effect size*

| **Effect size (Es) value** | **Classification** |
|---|---|
| Es < 0.2 | Weak |
| $0.2 \leq \| Es \| < 0.5$ | Small |
| $0.5 \leq \| Es \| < 0.8$ | Moderate |
| $\| Es \| \geq 0.8$ | Large |

(Cohen, 2013)

## 7.4 Results

### 7.4.1 Overground running

All participants (n=20) performed the overground running task on the force plate with their two legs (ACL-DF/non-inured). Knee kinematics and kinetics are shown in Table 7-3. MV values for each variable are also presented in Table 7-3. In general, there was only one statistically significant difference between the injured and the non-injured limbs reported during this task. In the knee extension moment ($p = 0.01$; Es = 0.93), the non-injured limb demonstrated greater knee extension moment than the injured one (2.86 Nm/Kg and 2.29 Nm/Kg respectively). In addition, there was a difference (Es = 0.62) in MV in terms of the knee valgus angle between the two limbs, as the injured limb reported less MV for the knee valgus angle than the non-injured one. However, no significant difference was found to occur following Bonferroni correction (corrected $\alpha = 0.01$).

In addition, four figures were provided to illustrate the knee motion and moment curves, which includes the 95% confidence interval for both legs in the sagittal and frontal planes throughout the entire (100%) stance period of the running task. They are as follows: Figure 7-1 displays the knee sagittal plane motion; Figure 7-2 displays the knee sagittal plane moment; Figure 7-3 displays the knee frontal plane motion; and Figure 7-4 displays the knee frontal plane moment. The V-GRF is also shown in Figure 7-5. All graphs presented in these figures represent the average scores obtained for all participants (n=20).

*Table 7-3: Knee kinematics, kinetics (peaks) and MV in each variable during running.*

| Variable | Injured leg | | Uninjured leg | | P-value | Effect size |
|---|---|---|---|---|---|---|
| | Mean | SD | Mean | SD | | |
| Knee flexion angle (°) | 43.17 | 6.69 | 46.20 | 7.24 | 0.14 | 0.43 |
| MV for knee flexion angle | 0.05 | 0.02 | 0.05 | 0.02 | 0.11 | 0.00 |
| Knee extension moment (Nm/Kg) | 2.29 | 0.69 | 2.86 | 0.52 | 0.01* | 0.93 |
| MV for knee extension moment | 0.11 | 0.06 | 0.09 | 0.05 | 0.31 | 0.36 |
| Knee valgus angle (°) | -3.61 | 4.14 | -2.18 | 4.12 | 0.09 | 0.34 |
| MV for knee valgus angle | 0.19 | 0.15 | 0.32 | 0.23 | 0.03 | 0.62 |
| Knee valgus moment (Nm/Kg) | 0.19 | 0.25 | 0.12 | 0.16 | 0.03 | 0.38 |
| MV for knee valgus moment | 0.54 | 0.29 | 0.64 | 0.25 | 0.17 | 0.37 |
| V-GRF (*Body weight) | 2.00 | 0.25 | 2.05 | 0.23 | 0.10 | 0.21 |
| MV for V-GRF | 0.03 | 0.03 | 0.02 | 0.03 | 0.41 | 0.33 |

*Negative (-) represents knee valgus angle and knee varus moment.*

**Significant difference between injured and uninjured legs following Bonferroni correction ($p \leq 0.01$).*

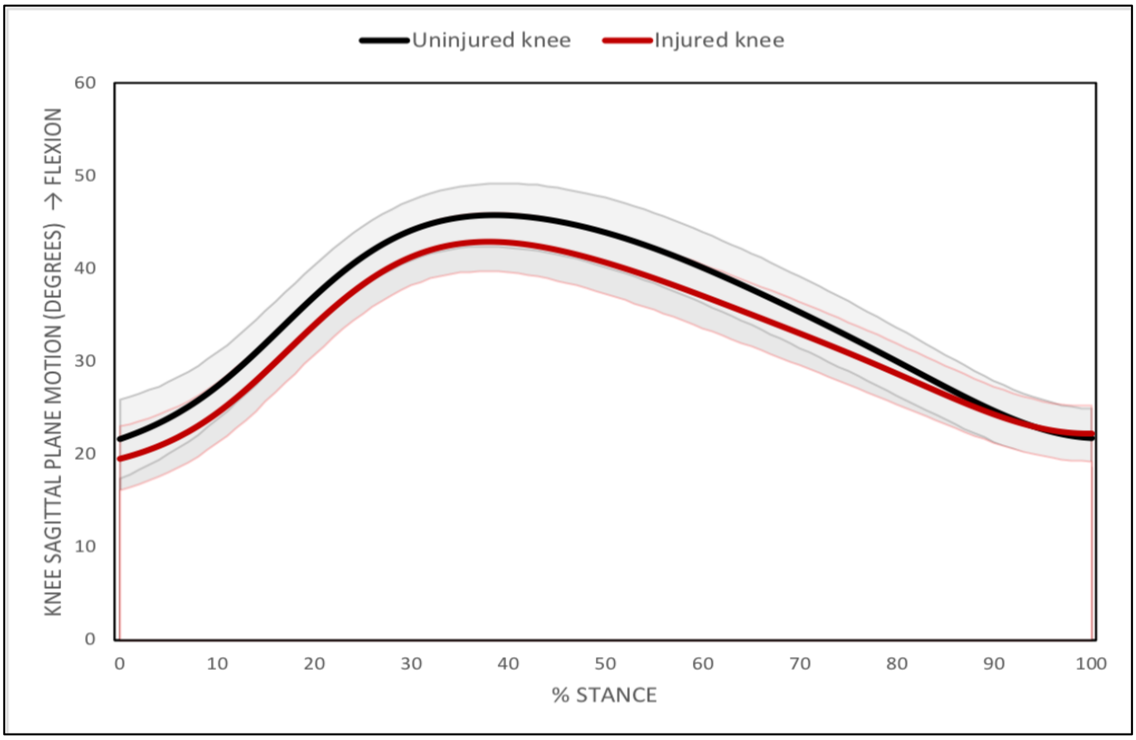



*Figure 7-1: Knee sagittal plane motion plus 95% confidence interval (shaded areas) for both legs during running task.*

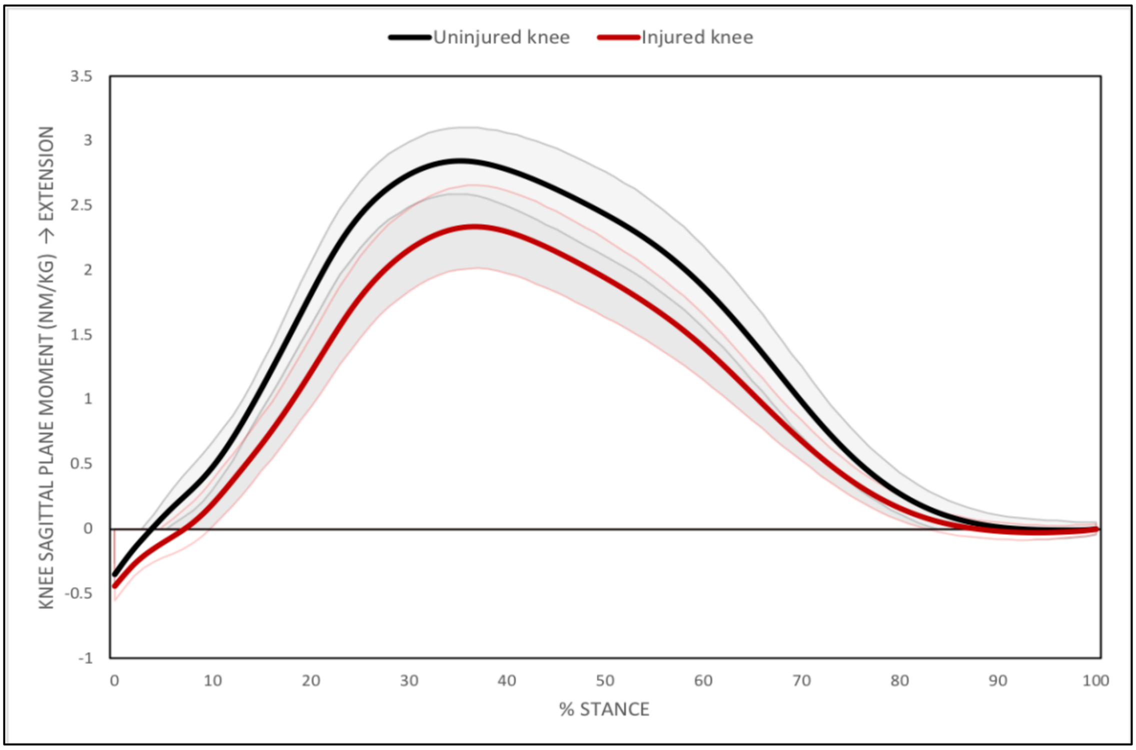



*Figure 7-2: Knee sagittal plane moment plus 95% confidence interval (shaded areas) for both legs during running task.*

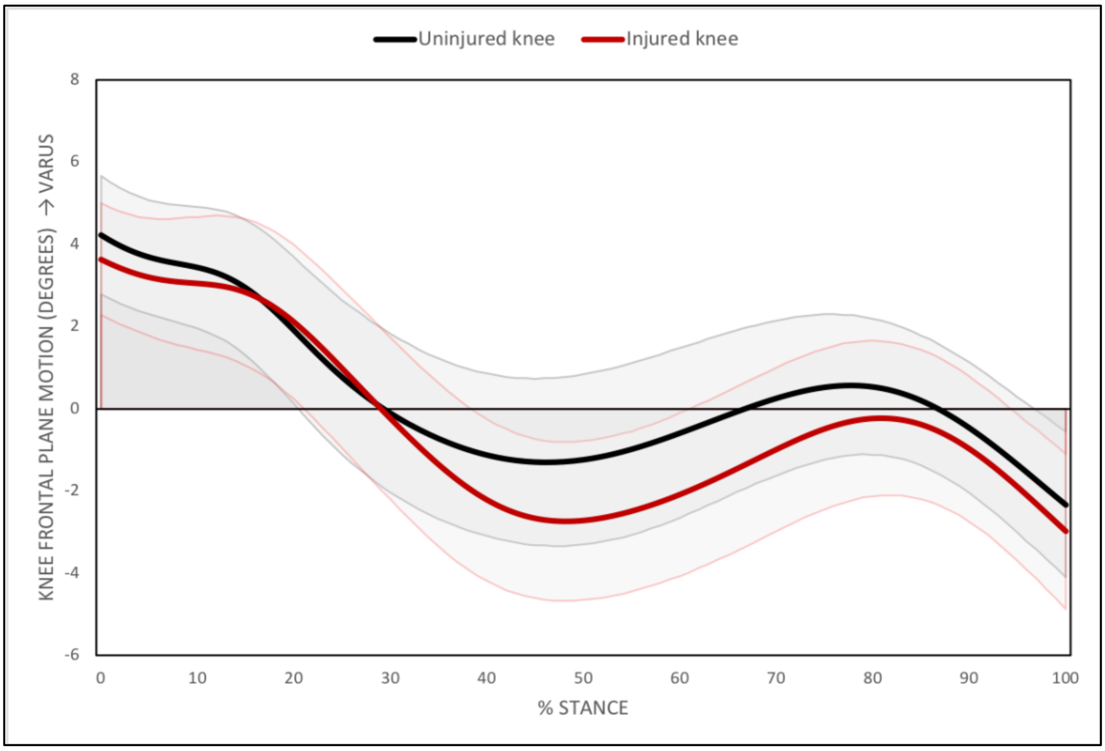



*Figure 7-3: Knee frontal plane motion plus 95% confidence interval (shaded areas) for both legs during running task.*

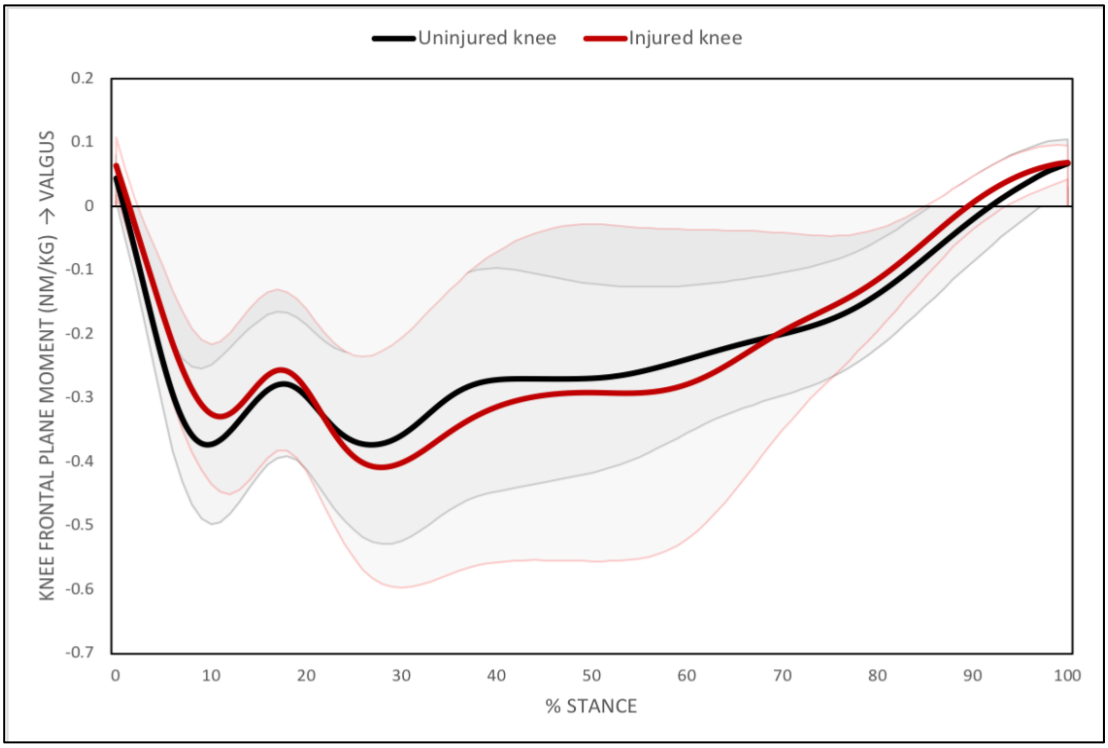



*Figure 7-4: Knee frontal plane moment plus 95% confidence interval (shaded areas) for both legs during running task.*

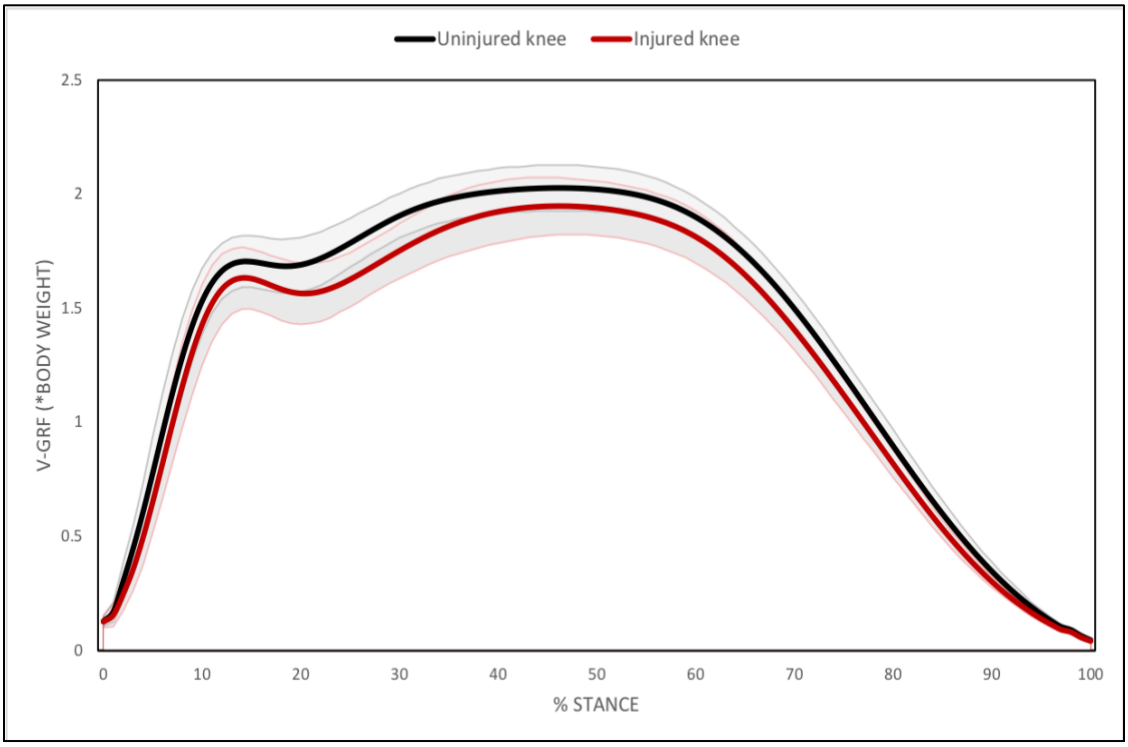



*Figure 7-5: Vertical ground reaction force plus 95% confidence interval (shaded areas) for both legs during running task.*

### 7.4.2 Cutting to 90 degrees

Each participant in this study (n=20) performed cutting to 90 degrees on the force plate with their two legs (ACL-DF/non-injured). Knee kinematics and kinetics results for this task are shown in Table 7-4. MV values for each variable are also presented in Table 7-4. Overall, of all the variables, there were four reported significant differences between injured and non-injured legs. Firstly, a significant difference occurred in MV for the peak knee flexion angle between the two limbs ($p = 0.01$; Es = 0.66). Another significant difference was found in the peak knee extension moment, where the uninjured limb demonstrated greater peak knee extension moment than the injured one ($p = 0.005$; Es = 0.75). There was also a significant difference in peak GRF between the two limbs, as the uninjured leg showed greater peak GRF than the injured one ($p = 0.008$; Es = 0.75). The last significant difference during this task was found to occur in MV for the peak GRF between the two limbs, as the injured leg showed greater MV for peak GRF than the uninjured one ($p = 0.01$; Es = 0.88). In addition, there was a difference with moderate effect size (Es = 0.55) in the peak knee flexion angle, where the injured leg showed a lower peak knee flexion angle than the non-injured one. Similarly, there was a difference with moderate effect size (Es = 0.54) in the peak knee valgus moment, where the injured leg showed less of a peak knee valgus moment than the uninjured one. However, these differences were not significant following Bonferroni correction (corrected $\alpha = 0.01$).

Furthermore, four figures were provided to illustrate the knee motion and moment curves, which includes the 95% confidence interval for both legs in the sagittal and frontal planes throughout the entire (100%) stance period of the cutting to 90 degrees task. They are as follows: Figure 7-6 displays the knee sagittal plane motion; Figure 7-7 displays the knee sagittal plane moment; Figure 7-8 displays the knee frontal plane motion; and Figure 7-9 displays the knee frontal plane moment. The V-GRF is also illustrated in Figure 7-10. All graphs presented in these figures represent the average scores obtained for all participants (n=20).

*Table 7-4: Knee kinematics, kinetics (peaks) and MV in each variable during cutting to 90 degrees task.*

| Variable | Injured leg | | Uninjured leg | | P-value | Effect size |
|---|---|---|---|---|---|---|
| | Mean | SD | Mean | SD | | |
| Knee flexion angle (°) | 56.04 | 9.47 | 61.85 | 11.51 | 0.04 | 0.55 |
| MV for knee flexion angle | 0.08 | 0.05 | 0.05 | 0.04 | 0.01* | 0.66 |
| Knee extension moment (Nm/Kg) | 1.42 | 0.61 | 1.82 | 0.45 | 0.005* | 0.75 |
| MV for knee extension moment | 0.17 | 0.10 | 0.14 | 0.08 | 0.27 | 0.33 |
| Knee valgus angle (°) | -4.18 | 4.73 | -4.58 | 4.42 | 0.76 | 0.09 |
| MV for knee valgus angle | 0.43 | 0.28 | 0.35 | 0.27 | 0.28 | 0.29 |
| Knee valgus moment (Nm/Kg) | 0.73 | 0.53 | 1.08 | 0.74 | 0.05 | 0.54 |
| MV for knee valgus moment | 0.34 | 0.18 | 0.32 | 0.20 | 0.68 | 0.10 |
| V-GRF (*Body weight) | 1.76 | 0.34 | 1.99 | 0.27 | 0.008* | 0.75 |
| MV for V-GRF | 0.07 | 0.05 | 0.03 | 0.04 | 0.01* | 0.88 |

*Negative (-) represents knee valgus angle and varus moment.*

*Significant difference between injured and uninjured legs following Bonferroni correction ($p \leq 0.01$).

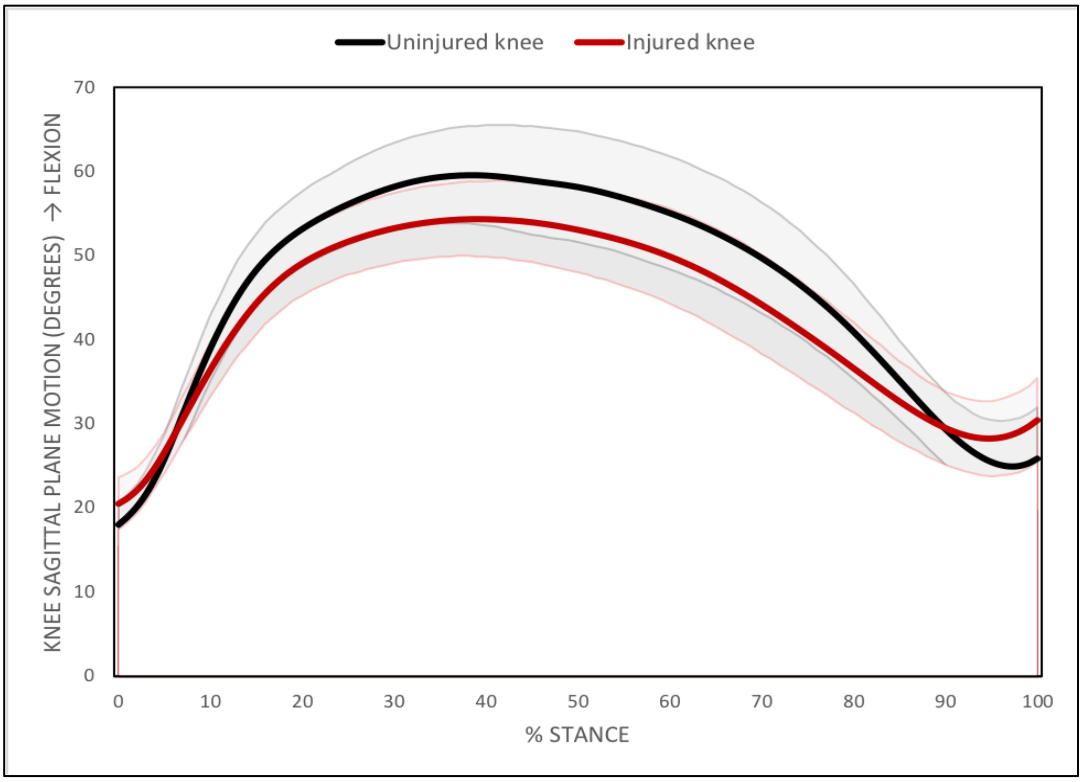



*Figure 7-6: Knee sagittal plane motion plus 95% confidence interval (shaded areas) for both legs during cutting to 90 degrees task.*

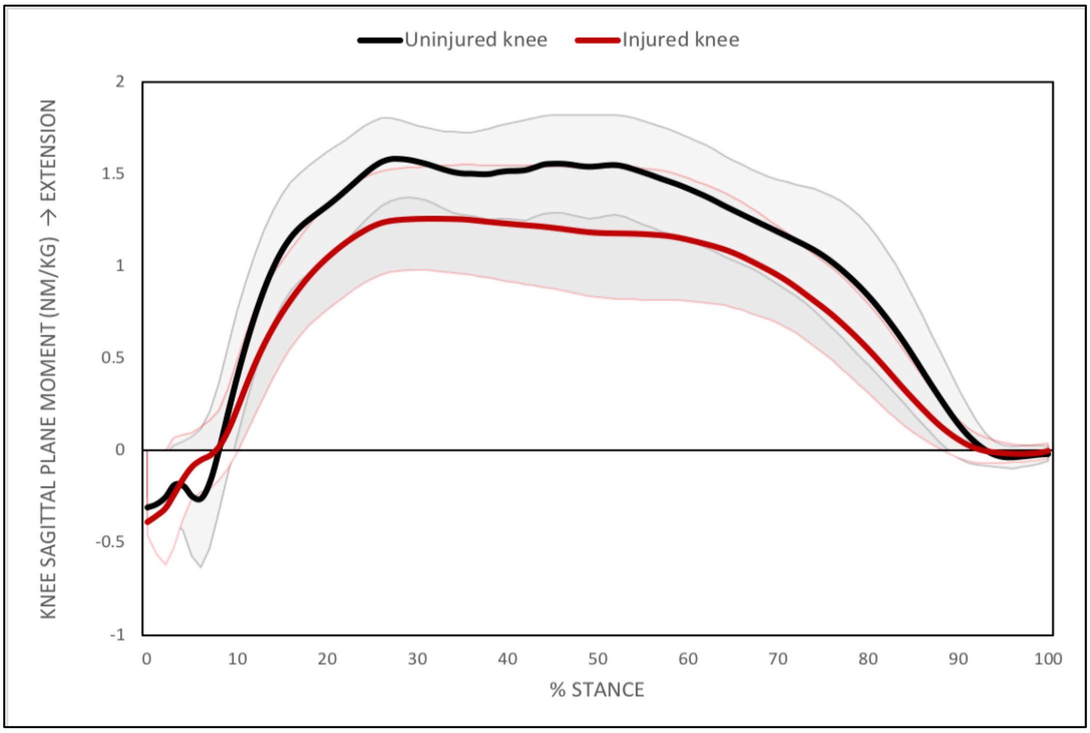



*Figure 7-7: Knee sagittal plane moment plus 95% confidence interval (shaded areas) for both legs during cutting to 90 degrees task.*

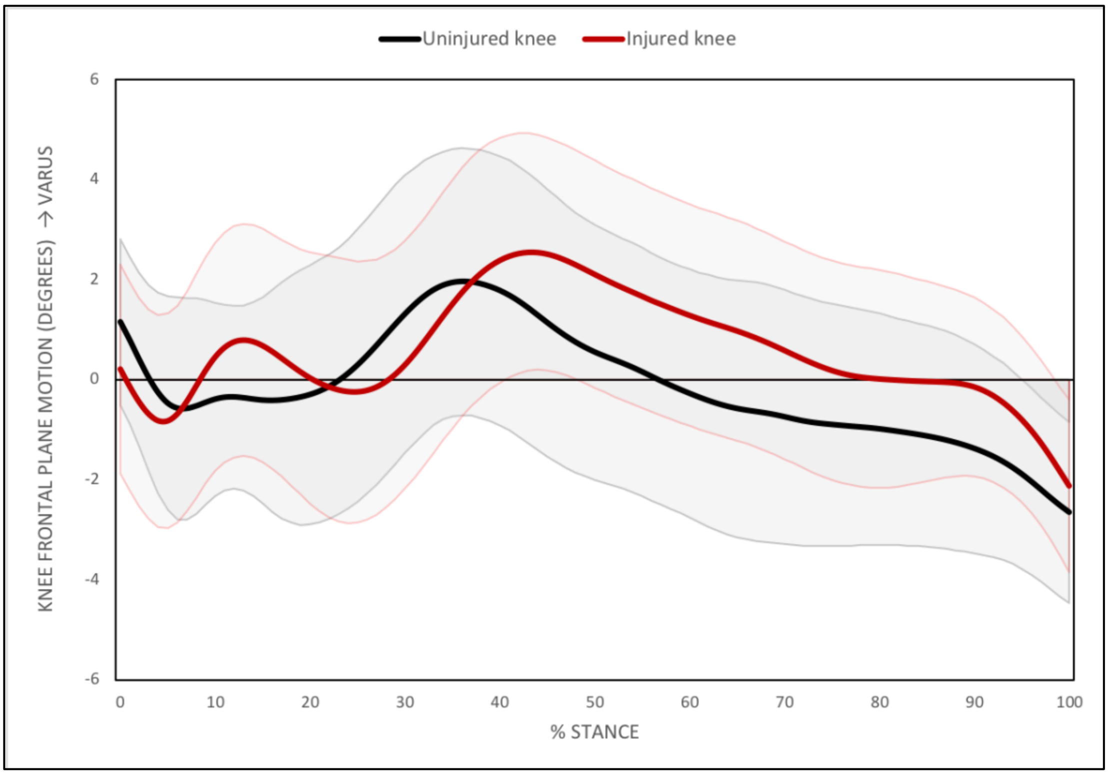



*Figure 7-8: Knee frontal plane motion plus 95% confidence interval (shaded areas) for both legs during cutting to 90 degrees task.*

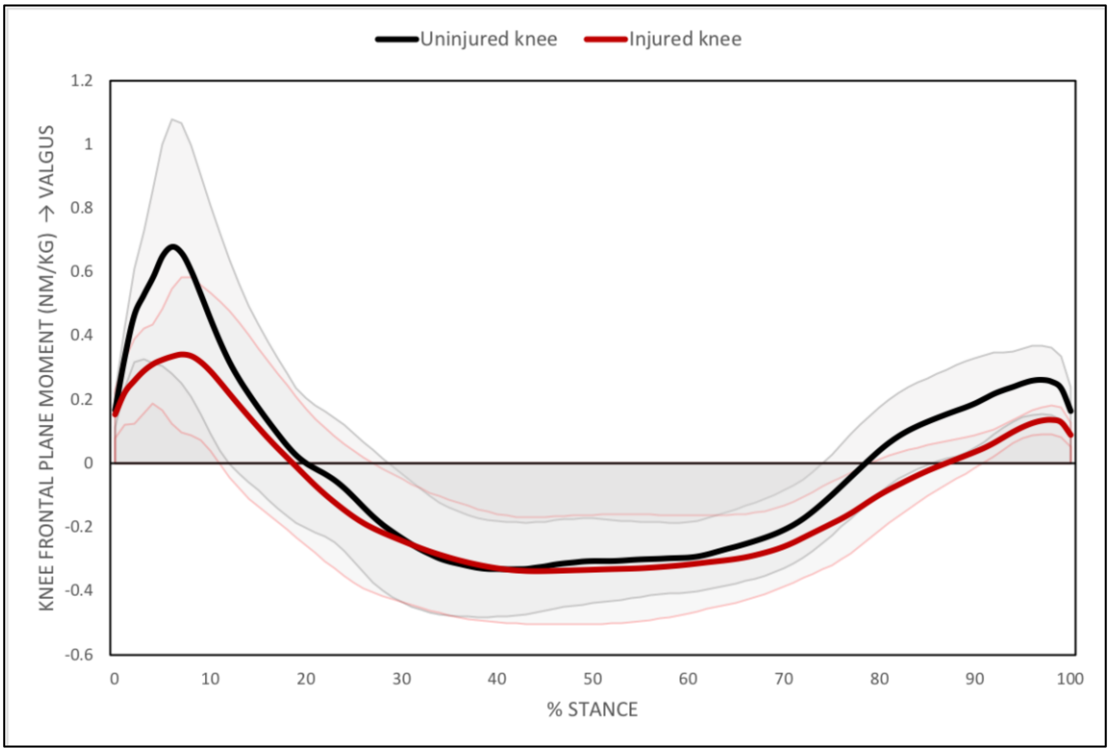



*Figure 7-9: Knee frontal plane moment plus 95% confidence interval (shaded areas) for both legs during cutting to 90 degrees task.*

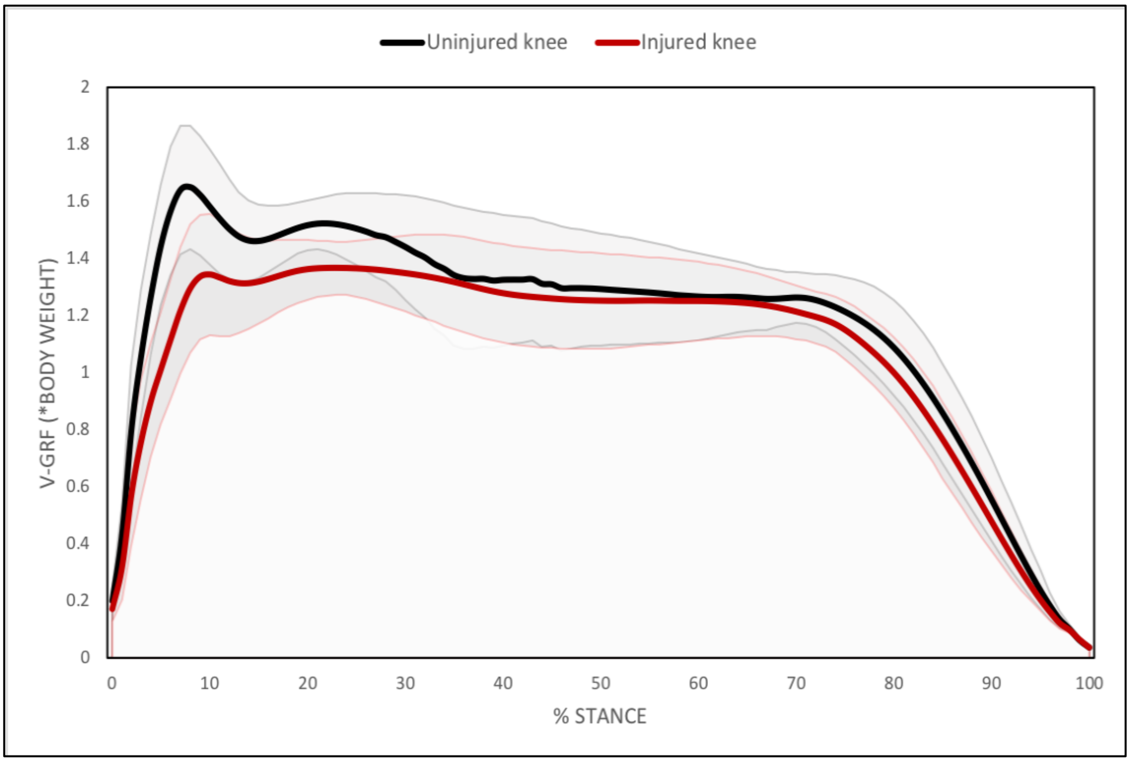



*Figure 7-10: Vertical ground reaction force plus 95% confidence interval (shaded areas) for both legs during cutting to 90 degrees task.*

### 7.4.3 SLL

All participants (n=20) performed SLL on the force plate with their two legs (ACL-DF/uninjured). Knee kinematics and kinetics are outlined in Table 7-5. MV values for each biomechanical variable are also presented in Table 7-5. In general, there was only one significant difference between the injured and the uninjured limbs reported during this task. This occurred in the knee extension moment ($p = 0.001$; Es = 1.50), where the injured limb demonstrated less knee extension moment than the uninjured one (2.39 Nm/Kg and 3.22 Nm/Kg respectively). Furthermore, there was a difference with moderate effect size (Es = 0.55) in the peak knee flexion angle, where the injured leg demonstrated a lower peak knee flexion angle than the uninjured one. There was also a difference with small effect size (Es = 0.26) in the peak GRF, where the injured leg showed less peak GRF than the uninjured one. However, these differences were not significant following Bonferroni correction (corrected $\alpha = 0.01$).

In addition, four figures were provided to illustrate the knee motion and moment curves, which includes the 95% confidence interval for both legs in the sagittal and frontal planes throughout the entire (100%) defined phase of the SLL task. They are as follows: Figure 7-11 displays the knee sagittal plane motion; Figure 7-12 displays the knee sagittal plane moment; Figure 7-13 displays the knee frontal plane motion; and Figure 7-14 displays the knee frontal plane moment. The V-GRF is also illustrated in Figure 7-15. All graphs presented in these figures represent the average scores obtained for all participants (n=20)

*Table 7-5: Knee kinematics, kinetics (peaks) and MV in each variable during SLL.*

| Variable | Injured leg | | Uninjured leg | | P-value | Effect size |
|---|---|---|---|---|---|---|
| | Mean | SD | Mean | SD | | |
| Knee flexion angle (°) | 54.26 | 11.04 | 60.16 | 10.59 | 0.04 | 0.55 |
| MV for knee flexion angle | 0.07 | 0.02 | 0.06 | 0.03 | 0.83 | 0.39 |
| Knee extension moment (Nm/Kg) | 2.39 | 0.57 | 3.22 | 0.56 | 0.001* | 1.50 |
| MV for knee extension moment | 0.09 | 0.04 | 0.09 | 0.06 | 0.41 | 0.00 |
| Knee valgus angle (°) | -2.52 | 3.72 | -2.28 | 3.83 | 0.79 | 0.06 |
| MV for knee valgus angle | 0.47 | 0.31 | 0.37 | 0.25 | 0.57 | 0.36 |
| Knee valgus moment (Nm/Kg) | 0.34 | 0.36 | 0.51 | 0.42 | 0.18 | 0.43 |
| MV for knee valgus moment | 0.50 | 0.22 | 0.51 | 0.20 | 0.79 | 0.05 |
| V-GRF (*Body weight) | 2.12 | 0.28 | 2.24 | 0.59 | 0.03 | 0.26 |
| MV for V-GRF | 0.02 | 0.03 | 0.02 | 0.04 | 0.41 | 0.00 |

*Negative (-) represents knee valgus angle and varus moment.*

**Significant difference between injured and uninjured legs following Bonferroni correction ($p \leq 0.01$).*

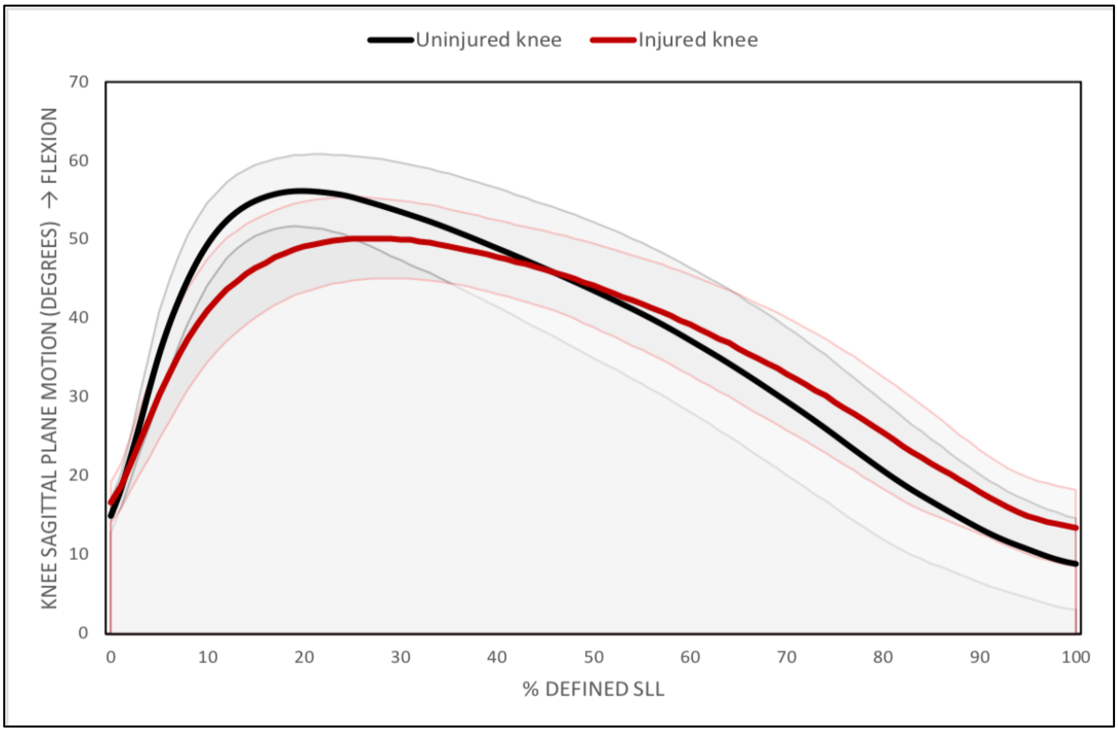



*Figure 7-11: Knee sagittal plane motion plus 95% confidence interval (shaded areas) for both legs during SLL task.*

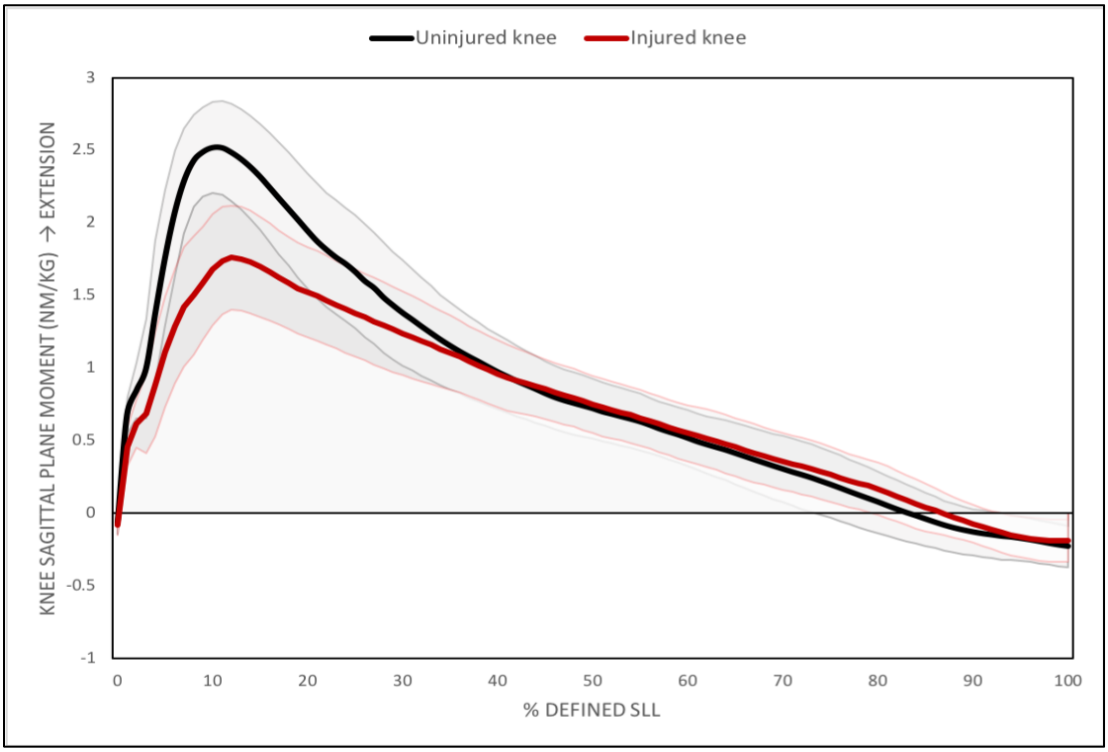



*Figure 7-12: Knee sagittal plane moment plus 95% confidence interval (shaded areas) for both legs during SLL task.*

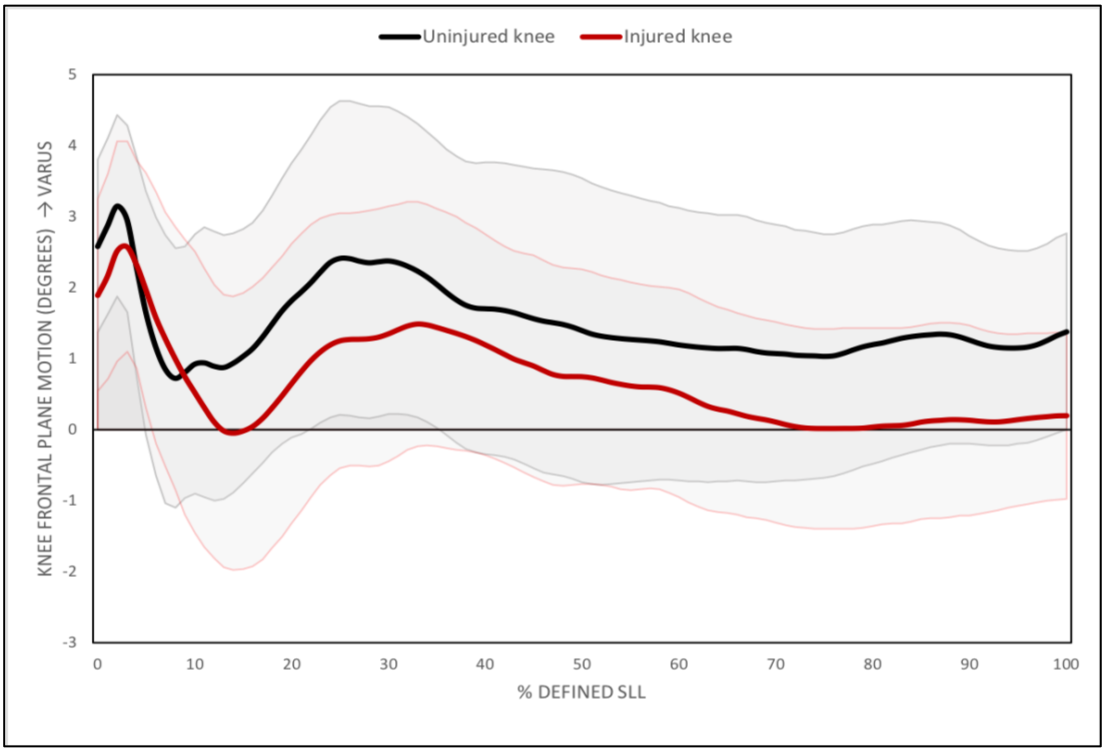



*Figure 7-13: Knee frontal plane motion plus 95% confidence interval (shaded areas) for both legs during SLL task.*

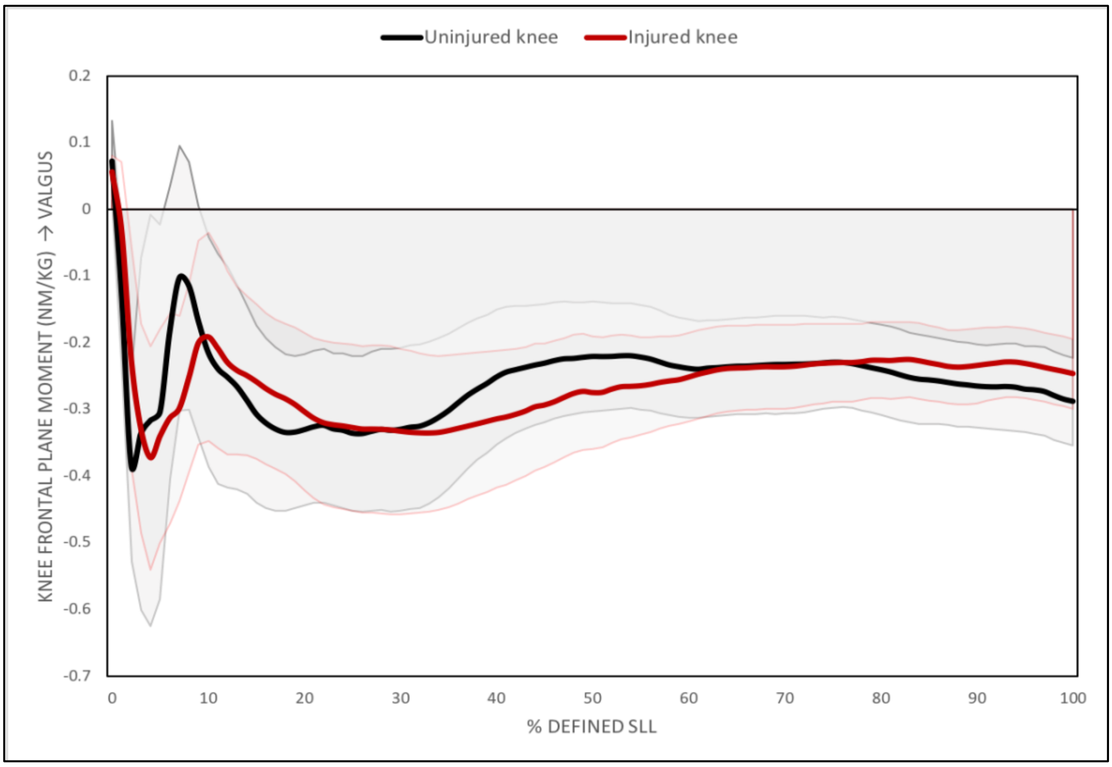



*Figure 7-14: Knee frontal plane moment plus 95% confidence interval (shaded areas) for both legs during SLL task.*

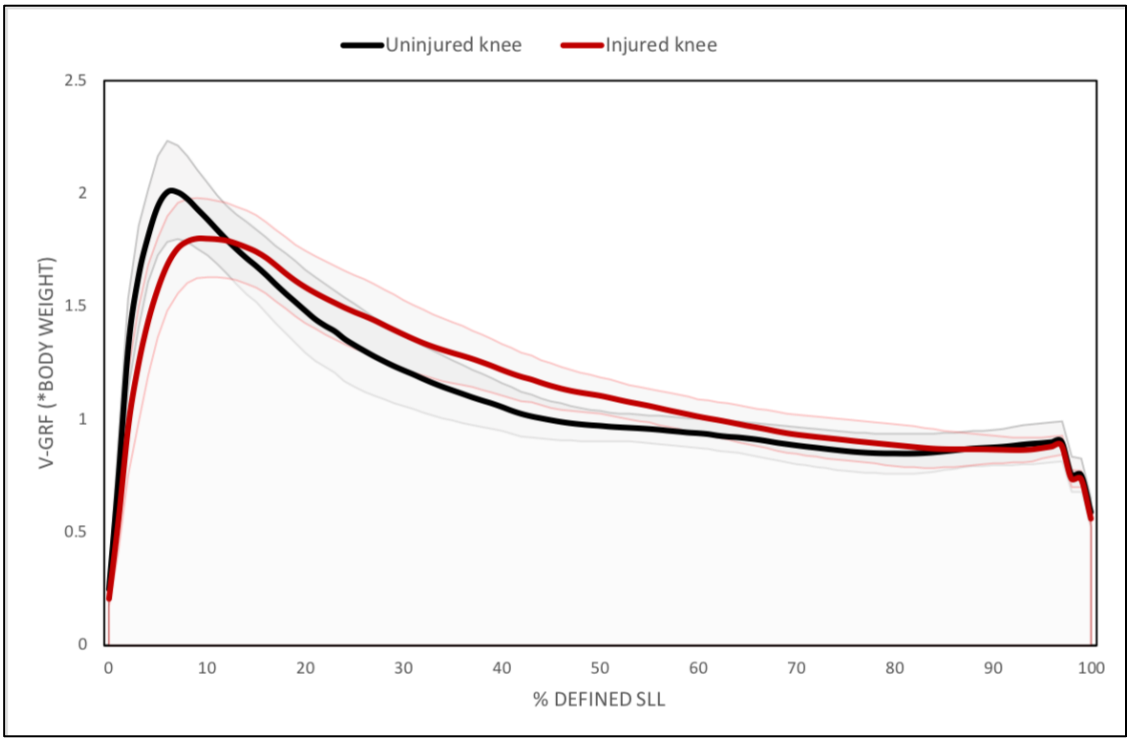



*Figure 7-15: Vertical ground reaction force plus 95% confidence interval (shaded areas) for both legs during SLL task.*

### 7.4.4 SLS

Each participant in this study (n=20) performed SLS on the force plate with their two legs (ACL-DF/uninjured). Knee kinematics and kinetics for this task are shown in Table 7-6. MV values for each variable are also presented in Table 7-6. Overall, of all the variables, only one variable reported a significant difference between the injured and uninjured legs. This occurred in the peak knee flexion angle (p = 0.01; Es = 0.50), where the injured limb demonstrated a lower peak knee flexion angle than the uninjured one (78.25 Nm/Kg and 83.22 Nm/Kg respectively). Furthermore, there was a difference with moderate effect size (Es = 0.50) in MV for the peak knee flexion angle, where the injured leg demonstrated greater MV for the peak knee flexion angle than the uninjured one. Similarly, there was a difference with moderate effect size (Es = 0.55) in MV for the peak knee valgus angle, where the injured leg showed greater MV for the peak knee valgus angle than the uninjured one. However, these differences were not significant following Bonferroni correction (corrected $\alpha = 0.01$).

Furthermore, four figures were drawn to illustrate the knee motion and moment curves, which includes the 95% confidence interval for both legs in the sagittal and frontal planes throughout the entire (100%) defined phase of the SLS task. They are as follows: Figure 7-16 displays the knee sagittal plane motion; Figure 7-17 displays the knee sagittal plane moment; Figure 7-18 displays the knee frontal plane motion; and Figure 7-19 displays the knee frontal plane moment. The V-GRF is also illustrated in Figure 7-20. All graphs presented in these figures represent the average scores obtained for all participants (n=20).

*Table 7-6: Knee kinematics, kinetics (peaks) and MV in each variable during SLS.*

| Variable | Injured leg | | Non-injured leg | | P-value | Effect size |
|---|---|---|---|---|---|---|
| | Mean | SD | Mean | SD | | |
| Knee flexion angle (°) | 78.25 | 11.76 | 83.22 | 7.74 | 0.01* | 0.50 |
| MV for knee flexion angle | 0.06 | 0.05 | 0.04 | 0.03 | 0.05 | 0.50 |
| Knee extension moment (Nm/Kg) | 1.12 | 0.44 | 1.15 | 0.53 | 0.79 | 0.06 |
| MV for knee extension moment | 0.20 | 0.14 | 0.22 | 0.13 | 0.52 | 0.15 |
| Knee valgus angle (°) | -0.35 | 2.45 | 0.76 | 3.08 | 0.16 | 0.40 |
| MV for knee valgus angle | 0.57 | 0.27 | 0.41 | 0.31 | 0.18 | 0.55 |
| Knee valgus moment (Nm/Kg) | 0.04 | 0.03 | 0.05 | 0.07 | 0.46 | 0.19 |
| MV for knee valgus moment | 0.63 | 0.23 | 0.62 | 0.22 | 0.86 | 0.04 |
| V-GRF (*Body weight) | 1.13 | 0.08 | 1.15 | 0.07 | 0.04 | 0.27 |
| MV for V-GRF | 0.03 | 0.02 | 0.04 | 0.02 | 0.18 | 0.50 |

*Negative (-) represents knee valgus angle and varus moment.*

**Significant difference between injured and non-injured legs following Bonferroni correction ($p \leq 0.01$).*

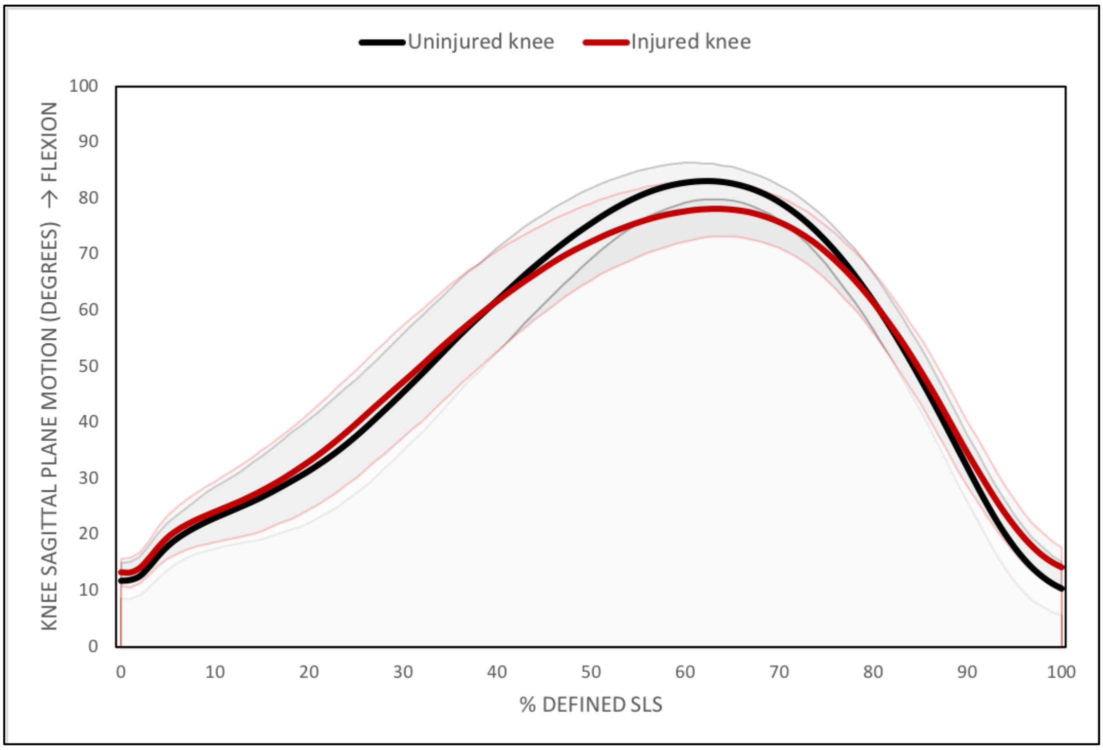



*Figure 7-16: Knee sagittal plane motion plus 95% confidence interval (shaded areas) for both legs during SLS task.*

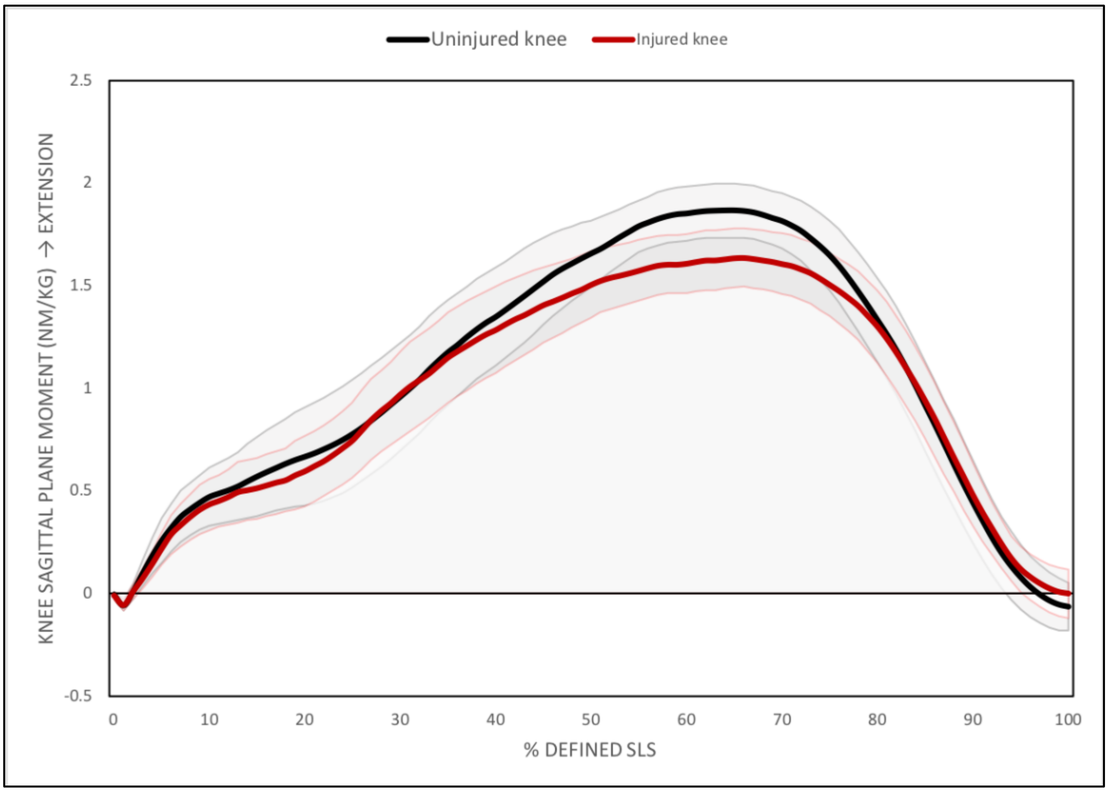



*Figure 7-17: Knee sagittal plane moment plus 95% confidence interval (shaded areas) for both legs during SLS task.*

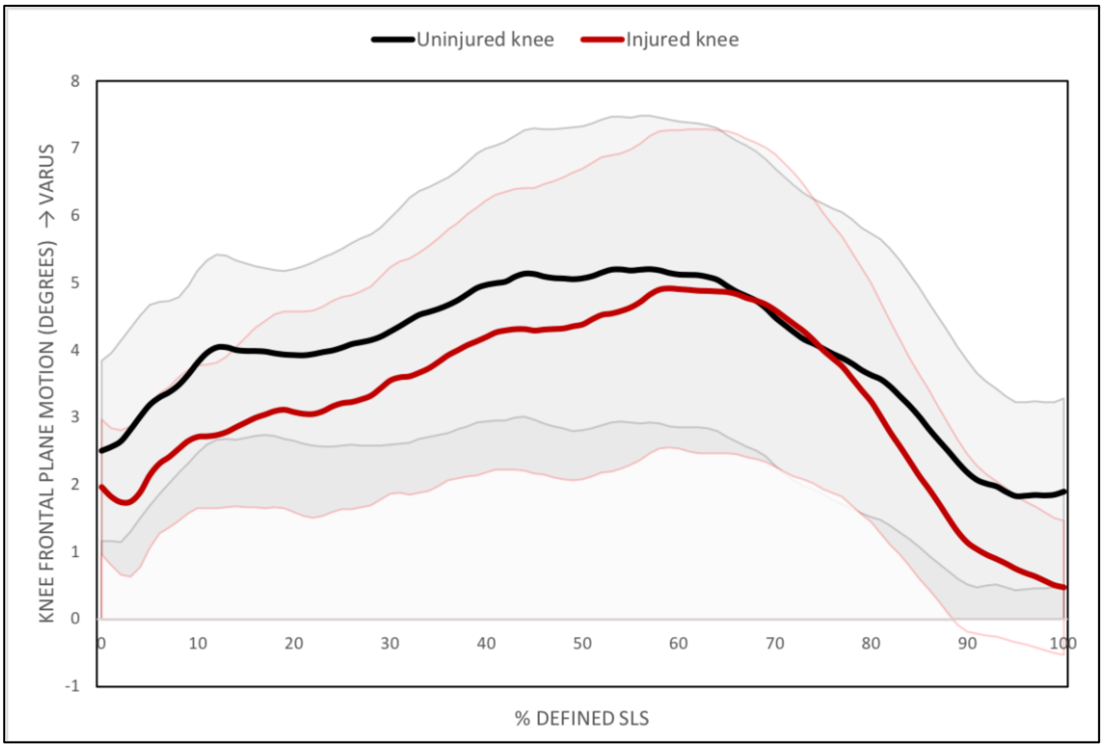



*Figure 7-18: Knee frontal plane motion plus 95% confidence interval (shaded areas) for both legs during SLS task.*

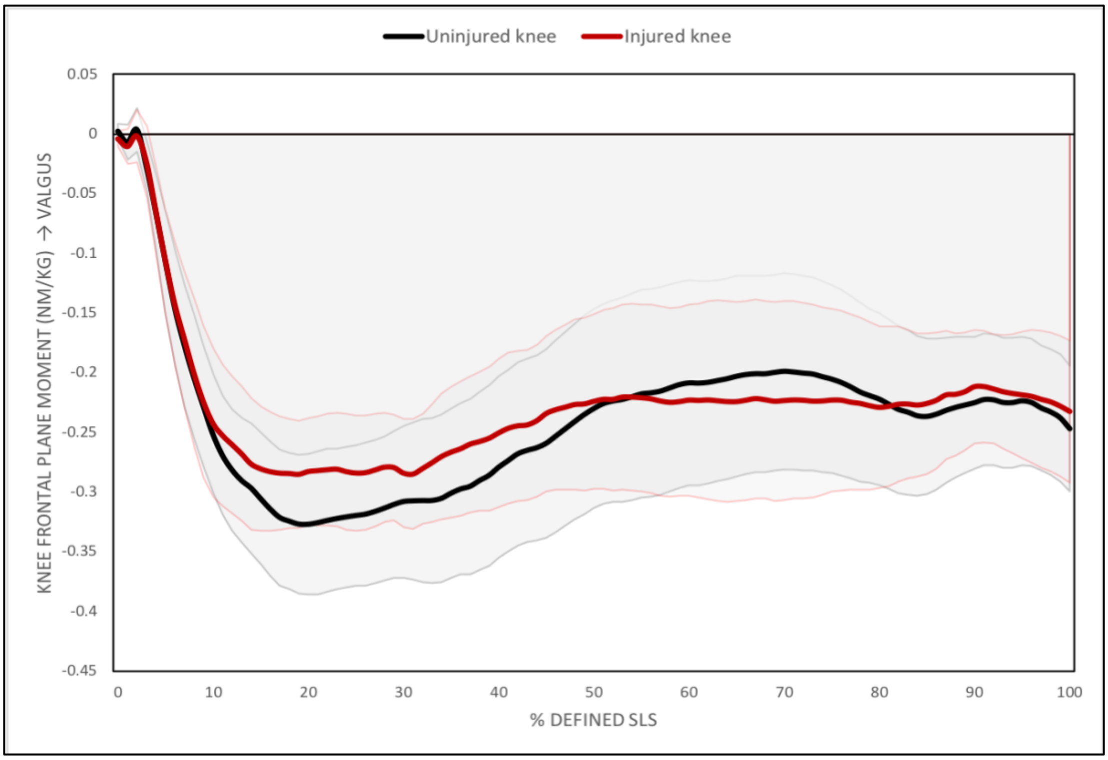



*Figure 7-19: Knee frontal plane moment plus 95% confidence interval (shaded areas) for both legs during SLS task.*

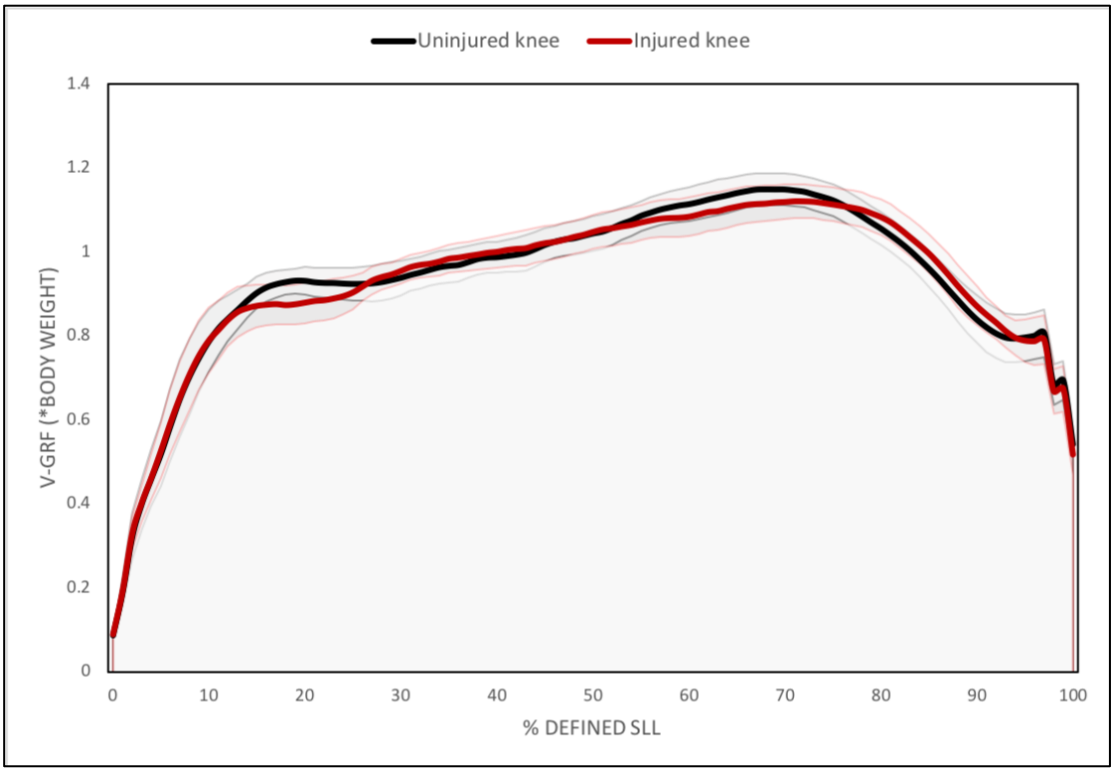



*Figure 7-20: Vertical ground reaction force plus 95% confidence interval (shaded areas) for both legs during SLS task.*

## 7.5 Discussion

The main aims of this study were to investigate MV and knee biomechanics in an ACL-DF population during four common sporting tasks. It also aimed to compare MV and knee biomechanics between injured and uninjured legs during these four tasks. In general, the results showed that there was a significant difference in at least one variable between injured and uninjured legs in all four tasks. Cutting to 90 degrees showed the highest number (4) of variables reporting significant differences between the two legs, as compared to the other tasks.

In terms of the biomechanical variables, the injured leg demonstrated a significantly lower peak knee extension moment during three tasks: running, cutting to 90 degrees and SLL ($p \leq 0.01$), than the uninjured one. Similar results were reported by Kawahara et al. (2012) and Lewek, Rudolph, Axe, and Snyder-Mackler (2002), who noted a decrease in extension moment in the ACL-DF knee, as compared to healthy controls during walking and jogging. Unfortunately, no studies to date have investigated the extension moment during functional sporting tasks. One explanation for the decrease in extension moment in the ACL-DF side could be attributable to the weakness of the quadriceps muscles caused by the injury, as a significant difference in quadriceps muscle strength between the injured and uninjured legs was reported in a previous study (H.-M. Lee, Cheng, & Liau, 2009). In addition, Ingersoll, Grindstaff, Pietrosimone and Hart (2008) reported that persistent quadriceps weakness and inhibition are common in both ACL-DF and ACL-RC populations. Unfortunately, this may lead to early-onset degeneration at the knee joint (Roos, 2005). However, the SLS task did not report a significant difference in the peak knee extension moment, as this task was less complex and less stressful as compared to the other task.

In addition, the findings of this study also showed that the injured knee demonstrated a lower peak knee flexion angle than the uninjured leg, in all tasks. Similar results were reported in previous research during: running and cutting tasks (Waite et al., 2005); jogging (Lewek et al., 2002; Rudolph, Axe, Buchanan, Scholz, & Snyder-Mackler, 2001; Chmielewski et al., 2001); SLS (Yamazaki et al., 2010); and walking (Kawahara et al., 2012; Berchuck, Andriacchi, Bach, & Reider, 1990; Rudolph, Axe, Buchanan, Scholz, & Snyder-Mackler, 2001; Chmielewski et al., 2001). In contrast, some studies have reported conflicting results and demonstrated that the injured knee showed higher knee flexion during walking (Shabani et al., 2015; Hurd & Snyder-Mackler, 2007; Kadaba et al., 1990); however, no study reported this during functional sporting tasks. In fact, large differences have emerged between the four sporting tasks and walking, and, therefore, it is inaccurate to compare them. Unfortunately, there is very limited evidence available regarding ACL-DF biomechanics during functional tasks. One explanation for demonstrating less knee flexion during sporting tasks could be because some ACL-DF patients develop a quadriceps avoidance phenomenon (Berchuck et al., 1990). The current study reported significant differences in knee extension moment between the injured and uninjured legs during three sporting tasks, thus increasing the likelihood of this phenomenon occurring. However, debate surrounds whether this pattern is more or less prevalent among individuals with an ACL-DF, as some studies did not support this finding (Roberts et al., 1999; Waite et al., 2005). They were of the opinion, based on their findings, that ACL-DF individuals included in their studies did not demonstrate an internal knee flexion moment, a decreased internal knee extension moment or a decreased duration of quadriceps electromyographic (EMG) activity during the stance phase.

Furthermore, the injured leg showed lower peak V-GRF in all tasks, with the exception of running, and it was statistically significant during cutting to 90 degrees ($p = 0.008$; Es = 0.75). As this task (cutting to 90 degrees) is considered to be the most complex of all four tasks, it requires greater coordination and more muscle balance. However, all these are affected by both injury and the absence of ACL. Accordingly, the DF side did hit the force plate with less force than the uninjured side, leading to a lower GRF. In terms of the running task, the findings from this study are consistent with previous research conducted by Chmielewski et al. (2001) and Rudolph et al. (2001) who did not report significant differences in peak GRF between ACL-DF and contralateral legs during jogging. However, the only study (Waite et al., 2005) investigating a cutting task in individuals with ACL-DF did not include kinetic values. These changes in biomechanical patterns may develop as a result of muscle adaptations and neuromuscular reprogramming, possibly in response to pain or instability, in order to stabilise the knee and to prevent reinjury (Reed Ferber et al., 2002).

Several points need to be taken into consideration regarding these studies. Firstly, all available research investigating biomechanics on individuals with ACL-DF have predominantly focused on walking gait only. Few studies have included jogging to gait (Lewek et al., 2002; Rudolph, Axe, Buchanan, Scholz, & Snyder-Mackler, 2001; Chmielewski et al., 2001), and only two study investigated functional sporting tasks (cutting and running) (Waite et al., 2005) and SLS (Yamazaki et al., 2010). However, these studies neither included kinetics in their investigations nor performed power calculations to justify their sample size, which was 15. Research undertaken by Waite et al. (2005) comprised 11 males and 4 females, and the data were not segregated. Conversely, while Yamazaki et al. (2010) did segregate male and female data, they did not perform power calculations. However, they did include larger sample sizes (32 males and 31 females). Speed is another important factor to take into consideration in cutting and running tasks. While this study determined the speed for running and cutting tasks, as explained in Chapter 3, unfortunately, all previously cited studies did not control speed and they also defined it as self-selected in cutting and running tasks (Waite et al., 2005). Similarly, other studies investigated jogging (Lewek et al., 2002; Rudolph, Axe, Buchanan, Scholz, & Snyder-Mackler, 2001; Chmielewski et al., 2001). Running speed has been identified as an element that can affect lower limb mechanical loading (Vanrenterghem, Venables, Pataky, & Robinson, 2012). Consequently, with low speed no meaningful loading on the knee occurs in terms of the risk of ACL injury. Finally, participant gender is one further important consideration, as gender differences were found to occur during these sporting tasks, as discussed in Chapter 4. While this study included only males, all aforementioned studies included both males and females. However, one study by Yamazaki et al. (2010) segregated male and female data. Hence, the current study is the first of its kind to investigate knee kinematics and kinetics in individuals with ACL-DF during SLL, as well as knee kinetics during SLS and cutting tasks.

In terms of MV, cutting to 90 degrees is the only task where significant differences were reported, between the ACL-DF side and the uninjured limb. These differences were noted in MV for the peak knee flexion angle ($p = 0.01$; Es = 0.66) and also for the peak GRF ($p = 0.01$; Es = 0.88). In both variables, the ACL-DF side demonstrated greater MV than the uninjured one. In addition, there were differences with moderate effect size in MV for the peak knee valgus angle between the two sides during SLS and running (Es = 0.62 and 0.55 respectively). Another difference with a moderate effect size was seen in MV for the peak knee flexion angle during SLS (Es = 0.50). In all these variables, the ACL-DF side demonstrated greater MV than the uninjured one, with one exception. MV for the peak knee valgus angle during running was the only variable in which the injured limb showed less MV than the uninjured one. That could be due to this task being less complex than the other tasks, while the mechanical load on the knee is also reduced during this activity. However, these differences were not significant following Bonferroni correction (corrected $\alpha = 0.01$).

MV studies in different injured populations have attracted the attention of some researchers. A very recent systematic review was undertaken by Baida et al. (2018), which investigated MV in lower limbs among different injured populations. However, they included studies with different lower limb injuries: ACL-RC, chronic ankle instability, patellar tendinopathy, running-related injuries, iliotibial band syndrome, patellofemoral pain, athletic groin pain and total hip arthroplasty. With regard to ACL injury, four studies were included, all of which investigated MV among ACL-RC during: a cutting task (Pollard et al., 2015); single leg hopping (Van Uden et al., 2003); instep kicks (Cordeiro, Cortes, Fernandes, Diniz, & Pezarat-Correia, 2015); and walking and jogging (Gribbin et al., 2016). In general, the results showed that there was an overall trend toward greater MV in the injured group, which is consistent with the findings of the present study. Unfortunately, no study was available to investigate MV in ACL-DF during functional tasks, in order to facilitate a comparative analysis. Hence, the current study is the first of its kind to investigate MV in individuals with ACL-DF during functional sporting tasks.

One explanation as to why greater MV was demonstrated in the ACL-DF limb could be that the absence of an ACL leads to a decrease in neuromuscular control, which gives rise to poorly controlled movement (Schmidt, 2003). In addition, the ACL injury can result in muscle weakness and low proprioception, all of which can affect knee biomechanics and MV. Another possible explanation could be that greater MV has been adopted as a compensatory movement in order to minimise the loading on painful tissues (Hodges & Tucker, 2011). This greater variability could create a risk of further damage or injury if it is sustained over time and if it remains untreated (Hamill, Palmer, & Van Emmerik, 2012; Stergiou, Harbourne, & Cavanaugh, 2006). The best exercise option in this case would be to target neuromuscular control.

### 7.6 Clinical implications

Different clinical implications were identified during this study. Firstly, clinicians and researchers should be aware of the significant differences in some knee biomechanics between the ACL-DF side and the uninjured one. Most importantly, they should also know that knee extension moment is significantly lower in the injured limb, which means that a weakness occurs in the quadriceps muscle and this should be targeted when implementing a rehabilitation programme. The SLS task did not show a significant difference in knee extension moment between the ACL-DF and the uninjured side, which implies that this task could be the best option to select to reduce the imbalance in the extension moment between the two sides. In addition, the ACL-DF side showed a lower peak knee flexion angle than the uninjured one.

Improving knee flexion during functional tasks should be targeted to avoid further damage or injury, which may occur due to altered knee flexion, as decreased knee injury is an indication of risk of further damage. In terms of MV, clinicians should be aware that the ACL-DF side demonstrated greater MV than the uninjured one, which could be attributable to a reduction in neuromuscular control. Hence, practitioners should include exercises which can improve this function, such as strengthening and movement control exercises.

## 7.7 Study novelty

As highlighted throughout this chapter, this novel study is the first of its kind to investigate MV for knee kinematics and kinetics among individuals with ACL-DF during four common sporting tasks. It also investigates knee kinematics and kinetics among individuals with ACL-DF during SLL. In addition, it investigates knee kinetics among individuals with ACL-DF during SLS and cutting tasks. Another novel aspect of this study is that it compares MV between an ACL-DF limb and the contralateral side during the four tasks, as well as to compare knee kinematics and kinetics between an ACL-DF limb and the contralateral side during SLL.

## 7.8 Limitations of this study and future work

The current study has several limitations. Firstly, the investigations in this study were confined to males only, as well as solely to an ACL-DF population. Future studies are needed to carry out such investigations in an ACL-DF female cohort, where it is highly anticipated that similar results would be reported. New research is also needed to develop a rehabilitation programme for ACL-DF patients that will help to reduce all these biomechanical impairments and facilitate their safe return to sport. Finally, there is a need for new research to develop other methods to identify MV in a non-3D world. This would enable the assessment of MV in clinical practice to be carried out at a low cost, as 3D motion analysis is not available in every clinic.

## 7.9 Conclusion

The ACL-DF limb demonstrated significant differences as compared to the uninjured one in both knee biomechanics and MV. However, this finding was not reported in all variables. The ACL-DF limb showed significantly lower peak knee extension moment as compared to the uninjured side in most of the tasks. It also showed both a lower peak knee flexion angle and peak GRF. However, the ACL-DF limb demonstrated greater MV than the uninjured one. These findings should be considered during the development of rehabilitation programmes and return to sport for ACL-DF patients in order to determine the most appropriate treatment.

Having completed these investigations, we have gained a full understanding of MV and knee biomechanics among an ACL-DF population. In addition, differences were identified in both MV and knee biomechanics between the injured and uninjured limbs. However, as a significant number of ACL-DF individuals choose to undertake surgery, as highlighted in the previous chapters, other research questions still need to investigate MV and knee biomechanics in the third ACL injury population, namely ACL-RC. This area will also be investigated in the next chapter.

# Chapter 8

## 8 Movement Variability and Knee Biomechanics in ACL-Reconstructed Individuals During Four Common Sporting Tasks

In the last chapter, we demonstrated that individuals who decided not to undergo ACL reconstruction (ACL-RC) experienced significant impairment in MV and knee biomechanics. This chapter seeks to understand MV and knee biomechanics in those who have undergone surgery.

### 8.1 Introduction

Individuals with an ACL rupture typically undergo surgery known as ACL reconstruction (ACL-RC). However, such an intervention is not always applied consistently throughout the globe (Filbay & Grindem, 2019). Despite the considerable debate that exists within the literature in terms of the most effective ACL rupture management strategies, two types of individuals have been identified as generally more likely to undergo ACL reconstruction. These include those with side-to-side laxity or who are very physically active (Delincé & Ghafil, 2012; Kostogiannis et al., 2007; Moksnes & Risberg, 2009). A number of researchers, for example, Kwok, Harrison and Servant (2012) contend that early ACL reconstruction is effective in restoring normal tibiofemoral joint kinematics, as well as to minimise the risk of joint instability. Furthermore, they claim that other associated benefits include reducing the likelihood of developing osteoarthritis or for secondary joint damage to occur.

However, a very recent study by Culvenor, Eckstein, Wirth, Lohmander and Frobell (2019) investigated the patellofemoral cartilage thickness over a five-year post-ACL injury period, while the impact of the treatment type using the KANON randomised controlled trial showed discouraging results in terms of ACL-RC surgery. They found that early-onset ACL-RC was associated with greater patellofemoral cartilage thickness loss over five years, as compared to optional delayed ACL-RC. In addition, the incidence of a second ACL injury either in the ipsilateral knee or in the contralateral one has been found to be greatest among younger athletes (Dekker et al., 2017; Morgan et al., 2016; Paterno et al., 2014; Webster & Hewett, 2019), for whom incidence rates of approximately 35% have been reported (Webster & Feller, 2016).

In order to understand why the reinjury rate is high, it is important to investigate lower limb biomechanics, as they can provide us with good insight into movement patterns among this specific cohort. A number of studies have investigated lower limb biomechanics in individuals with ACL-RC during various different sporting tasks. Findings have shown that biomechanical differences exist between the ACL-RC side and the uninjured side during different functional tasks, for example: landing (King et al., 2018; Lepley & Kuenze, 2018; Schmitt et al., 2015); cutting (King et al., 2018); running (Pairot-de-Fontenay et al., 2019; Pamukoff et al., 2017; Pratt & Sigward, 2017; Herrington, Alarifi, & Jones, 2017); and squats (Sanford, Williams, Zucker-Levin, & Mihalko, 2016; Neitzel, Kernozek, & Davies, 2002).

Unlike the ACL-DF population, the ACL-RC group has attracted multiple researchers to become involved in the investigation of biomechanics in this cohort during different functional sporting tasks: cutting (King et al., 2018; Pollard et al., 2015a; Saxby et al., 2016; Stearns & Pollard, 2013); running (Bowersock, Willy, DeVita, & Willson, 2017; Gribbin et al., 2016b; Herrington et al., 2017; Lee, Chow, & Tillman, 2014; Noehren, Abraham, Curry, Johnson, & Ireland, 2014; Pamukoff et al., 2017; Derek Pamukoff et al., 2018; Pratt & Sigward, 2017); SLL (Antolič et al., 1999; Gokeler et al., 2010; King, Richter, Franklyn-Miller, Daniels, Wadey, Moran et al., 2018; Pratt & Sigward, 2017; Webster & Feller, 2012b; Webster, Gonzalez-Adrio, & Feller, 2004; Webster, Santamaria, McClelland, & Feller, 2012) and SLS (Neitzel et al., 2002; Salem, Salinas, & Harding, 2003; Sanford et al., 2016).

In terms of MV, a significant correlation was found to occur between MV and biomechanical risk factors for ACL injury, as outlined in Chapter 5. Therefore, it is important to investigate the amount of MV ACL-RC individuals display, as it may play a role in this injury. In the literature, few studies have investigated MV in an ACL-RC population during different tasks: running and walking (Gribbin et al., 2016b); cutting to 45 degrees (Pollard et al., 2015b); and single leg hopping (Van Uden et al., 2003). They all reported that the reconstructed limb showed greater MV than the uninjured limb, in the case of most of the variables. However, Pollard et al. (2015b) included only females in their study and the cutting angle was at 45 degrees. In addition, to date no study has investigated MV in an ACL-RC cohort during SLL and SLS tasks. Therefore, this study aims to investigate MV and knee biomechanics in male ACL-RC individuals during four common sporting tasks. In addition, it aims to investigate whether significant differences exist in MV and knee biomechanics between the ACL-RC knee and the uninjured knee during these four common sporting tasks.

## 8.2 Null hypothesis

1- $H0_1$: There are no significant differences in knee kinematics and kinetics between the ACL-RC knee and the uninjured knee during four common sporting tasks.

2- $H0_2$: There are no significant differences in MV between the ACL-RC knee and the uninjured knee during these four tasks.

## 8.3 Methods

### 8.3.1 Participants

Power analysis was performed using G*Power (Version 3.1.9.4) based on the knee valgus moment obtained in a previous study by King et al. (2018) who investigated knee biomechanics in ACL-RC individuals during cutting to 90 degrees task (Es = 0.72). The rationale for this decision was that the peak knee valgus moment is a primary variable and it has been shown to be a risk factor in non-contact ACL injury, as previously outlined in Chapter 3. The reason for choosing the study by King et al. (2018) is because it has included similar tasks to the tasks included in this thesis and included similar population (ACL-RC). The effect size of 0.72 was used to calculate the sample size and was calculated electronically using G*Power software. A minimum of 14 participants was required in order to attain 80% statistical power with a type 1 error alpha level of 5% ($\alpha = 0.05$) for such investigations (Faul et al., 2007) (See appendix 12.7). A total of 20 ACL-RC individuals participated in this study which accounted for any potential drop-out and/or loss of data. The participants' demographic profile is presented in Table 8-1. All participants completed a consent form prior to engaging in the testing process.

Participants assigned to the ACL-RC group were included if they had a unilateral reconstructed ACL and had already returned to sport 6–12 months post-surgery (average of 9 ± 3 months). On testing at six months, participants had to have been cleared by their surgeon to return to sport and also have completed rehabilitation. All participant ACL-RC surgeries were performed by different surgeons operating in different orthopaedic practices, who had undertaken either a bone-patellar tendon-bone or a hamstring tendon autograft. In total, there were 18 participants with hamstring tendon autografts and two with bone-patellar tendon-bone autografts. All participants were Saudi Arabian nationals, drawn from the eastern region of the Kingdom, and they were either students or in employment. They all demonstrated good levels of fitness and health, while engaging in moderate levels of physical activity. This was defined as performing any sport or exercise for at least half an hour, three times a week, in the previous six months (Munro & Herrington, 2011). Potential participants were excluded if they had a history of multiple ACL injuries or if a total knee dislocation occurred at the time of injury; however, meniscectomy or concurrent meniscus repair may have been performed at the time of ACL-RC. The study also excluded participants who had undergone failed meniscal repair/graft surgery or multiple ligament reconstruction, or experienced surgical complications or significant chondral failure.

*Table 8-1: Participants' demographic profiles*

| | Number (Gender) | Mean | SD |
|---|---|---|---|
| Age (years) | 20 (male) | 26.15 | 3.76 |
| Height (m) | 20 (male) | 1.72 | 0.07 |
| Body mass (kg) | 20 (male) | 73.3 | 10.33 |

### 8.3.2 Preparation and tasks undertaken

Each participant was required to do some general warm-up exercises of ten minutes' duration, followed by a further five-minute warm-up on an exercise bike, prior to beginning the test. The preparation process was explained in detail in Chapter 4 (Section 4.5). In terms of the tasks undertaken, all participants were invited to complete all four functional sporting tasks: SLS, running, cutting to 90 degrees and SLL with their two legs (both ACL-RC and uninjured). Details of the requirements involved in performing these tasks were outlined in Chapter 4 (Section 4.5.1 for running, Section 4.5.2 for SLS, Section 4.5.3 for SLL and Section 4.5.4 for cutting to 90 degrees).

### 8.3.3 Procedure

The biomechanical data were collected for this study using 3D motion analysis at the 3D motion analysis laboratory at Imam Abdulrahman Bin Faisal University which was described in details in Chapter 4 (Section 4.2.2).

### 8.3.4 Outcome measures

This chapter includes all outcome measures, as explained in Chapter 4 (Section 4.9).

### 8.3.5 Statistical analysis

All statistical analysis in this study was carried out using the Statistical Package for the Social Sciences (SPSS) (Version 24). Descriptive analysis (means and standard deviations) was performed. Prior to conducting the data analysis, its normality distribution was observed visually following application of a Shapiro-Wilk test. A variable is normally distributed if the p-value is greater than 0.05 on the latter test; however, it is not normally distributed if the p-value is equal or less than 0.05. The differences between injured and non-injured legs were tested using a paired t-test for parametric variables (normally distributed) and a Wilcoxon Rank test for nonparametric (not normally distributed). The significance level was set at 5% (significant level when the p-value $< 0.05$, non-significant level when the p-value $> 0.05$).

In addition, in order to decrease the risk of type 1 errors occurring, the Bonferroni correction was applied to the alpha level used ($\alpha = 0.05$) for all comparisons carried out, as there were multiple comparisons conducted in this study (corrected $\alpha = 0.01$). The effect size for each comparison was also provided, along with p-values, in order to be able to fully interpret the results. The effect size was calculated using Cohen's d Index (Cohen, 2013). Table 8-2 demonstrates the levels and classifications of effect size according to Cohen (2013). In terms of MV values, a zero value signals that no variability has occurred while, conversely, a score of one is indicative of the highest levels of variability.

*Table 8-2: Levels and classification of effect size.*

| **Effect size (Es) value** | **Classification** |
|---|---|
| Es < 0.2 | Weak |
| $0.2 \leq \lvert Es \rvert < 0.5$ | Small |
| $0.5 \leq \lvert Es \rvert < 0.8$ | Moderate |
| $\lvert Es \rvert \geq 0.8$ | Large |

(Cohen, 2013)

## 8.4 Results

### 8.4.1 Overground running

All participants (n=20) performed the overground running task on the force plate with both legs (ACL-RC/uninjured). Knee kinematics and kinetics are presented in Table 8-3. MV values for each variable are also shown in Table 8-3. In general, there were two statistically significant differences between the injured and the uninjured limbs reported during this task. The first one occurred in the peak knee flexion angle, where the reconstructed leg showed a lower peak knee flexion angle than the uninjured one ($p = 0.008$; Es = 0.57). The second one was reported in the peak knee extension moment ($p = 0.002$; Es = 0.59), as the reconstructed limb demonstrated lower peak knee extension moment than the uninjured one (2.32 Nm/Kg and 2.76 Nm/Kg, respectively).

In addition, four figures were drawn to illustrate the knee motion and moment curves, which includes the 95% confidence interval for both legs in the sagittal and frontal planes throughout the entire (100%) stance period of the running task. They are as follows: Figure 8-1 displays the knee sagittal plane motion; Figure 8-2 displays the knee sagittal plane moment; Figure 8-3 displays the knee frontal plane motion; and Figure 8-4 displays the knee frontal plane moment. The V-GRF is also presented in Figure 8-5. All graphs illustrated in the figures represent the average scores obtained for all participants (n=20).

*Table 8-3: Knee kinematics, kinetics (peaks) and MV in each variable during running.*

| Variable | RC leg | | Uninjured leg | | P-value | Effect size |
|---|---|---|---|---|---|---|
| | Mean | SD | Mean | SD | | |
| Knee flexion angle (°) | 40.19 | 8.11 | 44.27 | 6.07 | 0.008* | 0.57 |
| MV for knee flexion angle | 0.06 | 0.05 | 0.05 | 0.03 | 0.51 | 0.24 |
| Knee extension moment (Nm/Kg) | 2.32 | 0.83 | 2.76 | 0.66 | 0.002* | 0.59 |
| MV for knee extension moment | 0.15 | 0.18 | 0.11 | 0.06 | 0.50 | 0.30 |
| Knee valgus angle (°) | -1.86 | 4.88 | -1.87 | 4.27 | 0.99 | 0.00 |
| MV for knee valgus angle | 0.36 | 0.32 | 0.30 | 0.23 | 0.85 | 0.22 |
| Knee valgus moment (Nm/Kg) | 0.23 | 0.35 | 0.19 | 0.23 | 0.77 | 0.14 |
| MV for knee valgus moment | 0.54 | 0.21 | 0.57 | 0.28 | 0.65 | 0.12 |
| V-GRF (*Body weight) | 1.91 | 0.49 | 2.00 | 0.37 | 0.79 | 0.21 |
| MV for V-GRF | 0.06 | 0.19 | 0.01 | 0.003 | 0.43 | 0.37 |

*Negative (-) represents knee valgus angle and knee varus moment. *Significant difference between reconstructed and uninjured legs following Bonferroni correction ($p \leq 0.01$).*

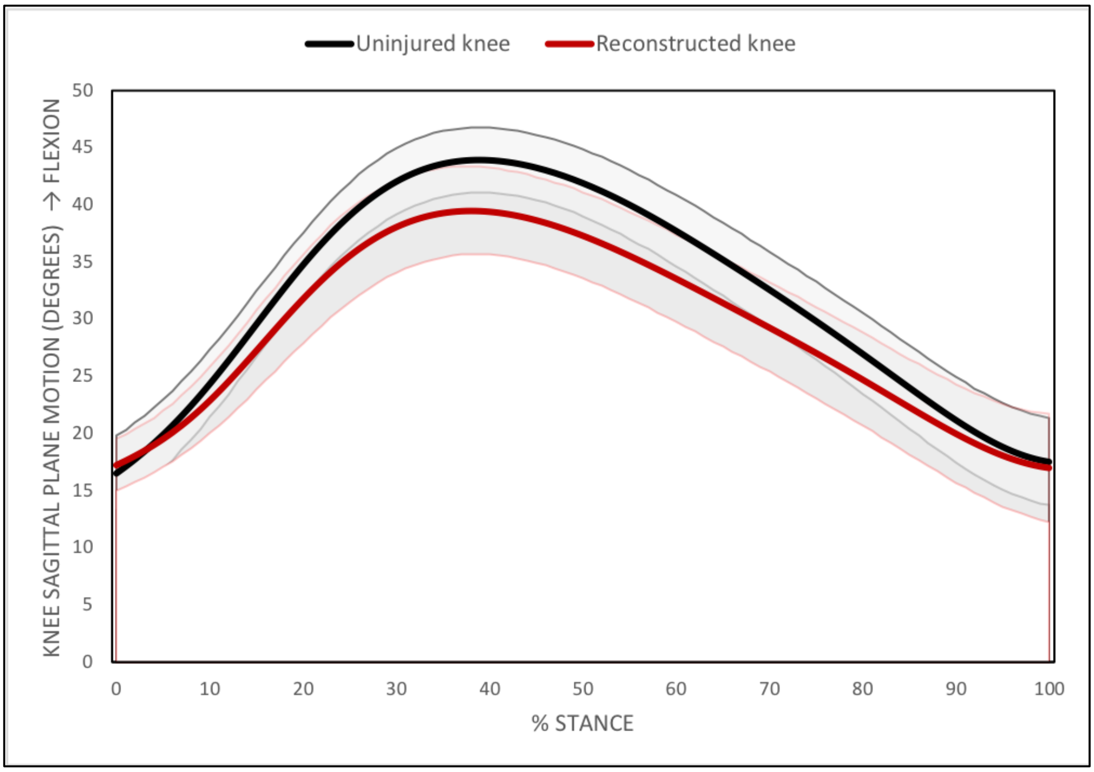



*Figure 8-1: Knee sagittal plane motion plus 95% confidence interval (shaded areas) for both legs during running task.*

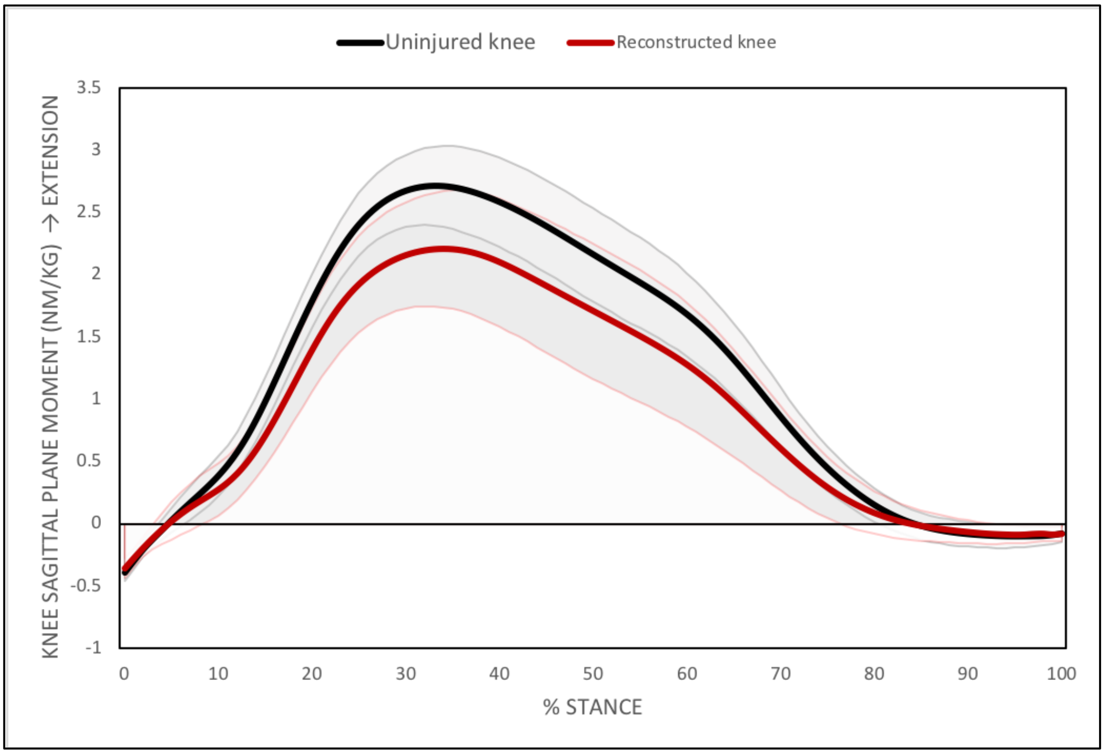



*Figure 8-2: Knee sagittal plane moment plus 95% confidence interval (shaded areas) for both legs during running task.*

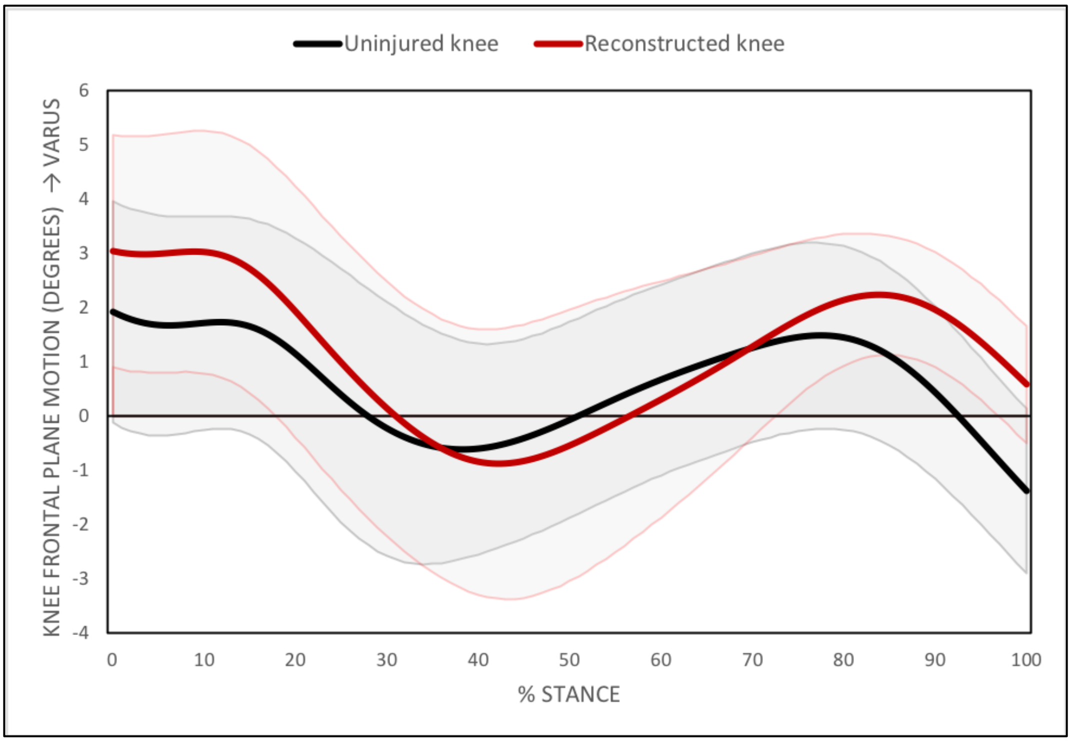



*Figure 8-3: Knee frontal plane motion plus 95% confidence interval (shaded areas) for both legs during running task.*

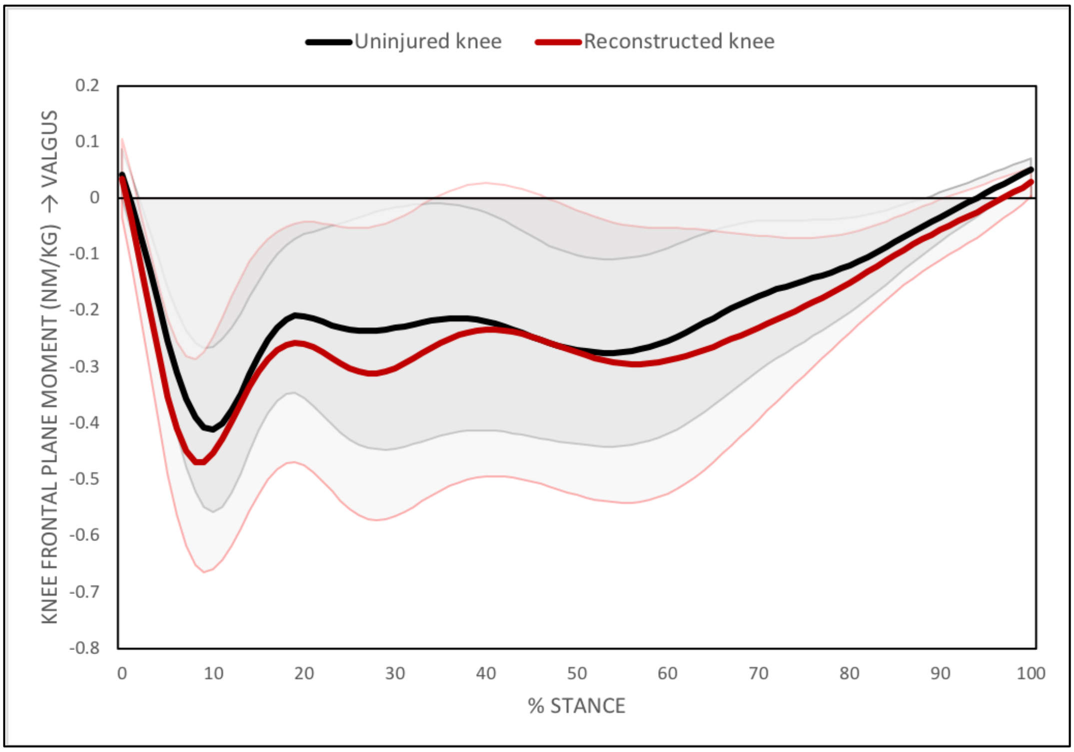



*Figure 8-4: Knee frontal plane moment plus 95% confidence interval (shaded areas) for both legs during running task.*

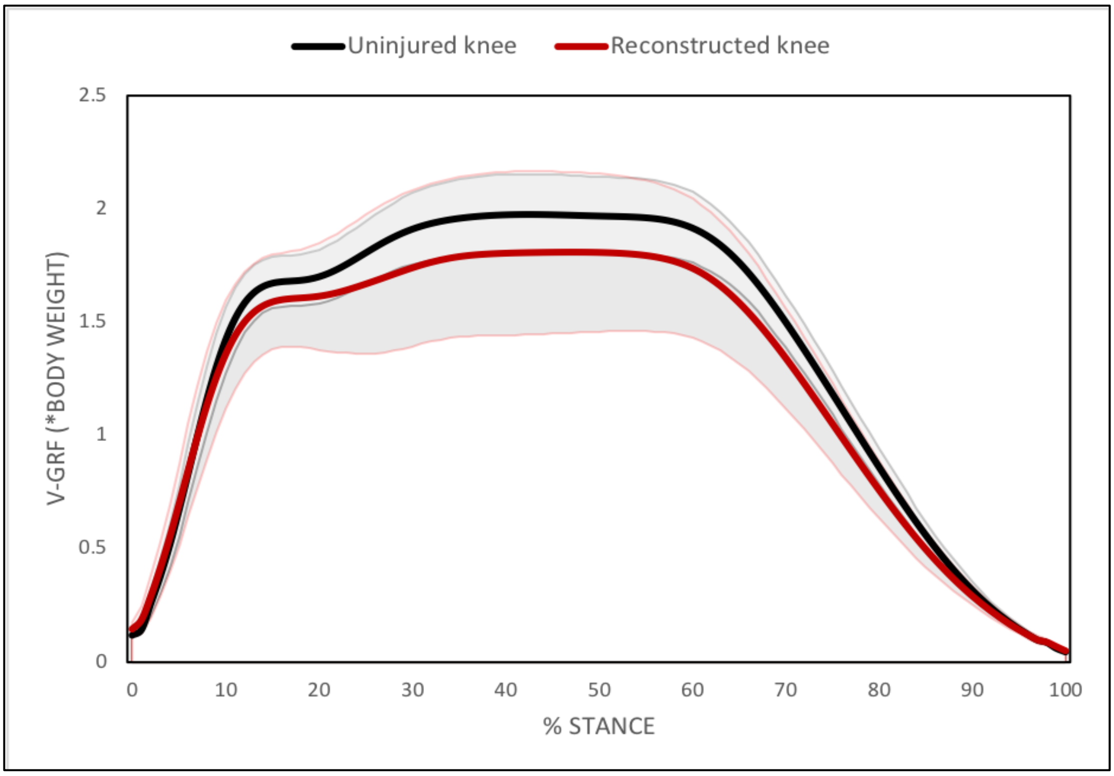



*Figure 8-5: Vertical ground reaction force plus 95% confidence interval (shaded areas) for both legs during running task.*

### 8.4.2 Cutting to 90 degrees

Each participant in this study (n=20) performed a cutting to 90 degrees task on the force plate with both legs (ACL-RC/uninjured). Knee kinematics and kinetics results for this task are shown in Table 8-4. MV values for each variable are also presented in Table 8-4. Overall, of all the variables, two significant differences occurred between the reconstructed and uninjured legs. Firstly, a significant difference occurred in the peak knee extension moment, where the reconstructed limb demonstrated a lower peak knee extension moment than the uninjured one ($p = 0.01$; Es = 0.43). The other significant difference was found in the peak knee flexion angle, where the reconstructed leg showed a lower peak knee flexion angle than the uninjured one ($p = 0.04$; Es = 0.38). However, the latter was not significant following Bonferroni correction (corrected $\alpha = 0.01$).

In addition, there was a difference with moderate effect size (Es = 0.44) in MV for the peak knee valgus moment, where the reconstructed leg showed a greater MV for the peak knee valgus moment than the uninjured one. However, this difference was not significant ($p = 0.23$).

Furthermore, four figures were drawn to illustrate the knee motion and moment curves, which includes the 95% confidence interval for both legs in the sagittal and frontal planes throughout the entire (100%) stance period of the cutting to 90 degrees task. They are as follows: Figure 8-6 displays the knee sagittal plane motion; Figure 8-7 displays the knee sagittal plane moment; Figure 8-8 displays the knee frontal plane motion; and Figure 8-9 displays the knee frontal plane moment. The V-GRF is also illustrated in Figure 8-10. All graphs presented in these figures represent the average scores obtained for all participants (n=20).

*Table 8-4: Knee kinematics, kinetics (peaks) and MV in each variable during cutting to 90 degrees task.*

| Variable | RC leg | | Uninjured leg | | P-value | Effect size |
|---|---|---|---|---|---|---|
| | Mean | SD | Mean | SD | | |
| Knee flexion angle (°) | 61.05 | 12.41 | 65.51 | 10.83 | 0.04 | 0.38 |
| MV for knee flexion angle | 0.08 | 0.09 | 0.07 | 0.07 | 0.26 | 0.13 |
| Knee extension moment (Nm/Kg) | 1.39 | 0.62 | 1.64 | 0.50 | 0.01* | 0.43 |
| MV for knee extension moment | 0.21 | 0.17 | 0.17 | 0.17 | 0.39 | 0.24 |
| Knee valgus angle (°) | -4.21 | 5.1 | -5.96 | 5.08 | 0.09 | 0.34 |
| MV for knee valgus angle | 0.24 | 0.14 | 0.32 | 0.28 | 0.20 | 0.36 |
| Knee valgus moment (Nm/Kg) | 1.25 | 0.59 | 1.31 | 0.74 | 0.62 | 0.09 |
| MV for knee valgus moment | 0.27 | 0.17 | 0.21 | 0.09 | 0.23 | 0.44 |
| V-GRF (*Body weight) | 1.89 | 0.26 | 1.90 | 0.23 | 0.77 | 0.04 |
| MV for V-GRF | 0.06 | 0.08 | 0.05 | 0.07 | 0.16 | 0.13 |

*Negative (-) represents knee valgus angle and knee varus moment.*

**Significant difference between reconstructed and uninjured legs following Bonferroni correction ($p \leq 0.01$).*

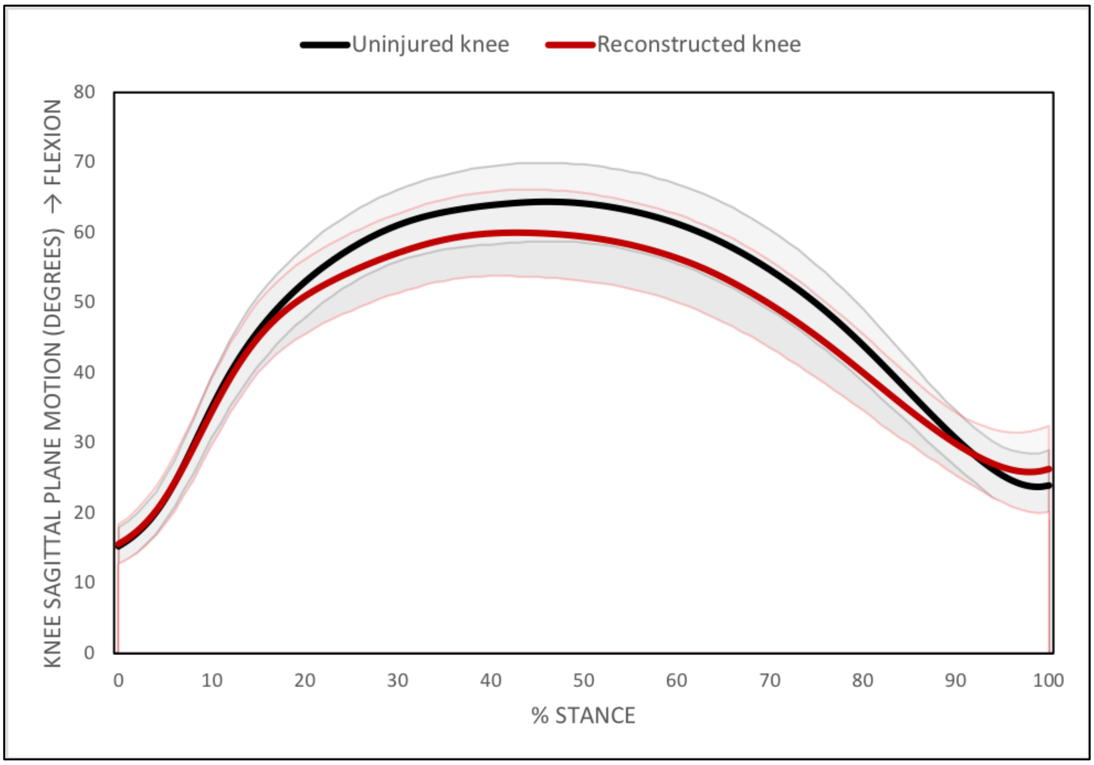



*Figure 8-6: Knee sagittal plane motion plus 95% confidence interval (shaded areas) for both legs during cutting to 90 degrees task.*

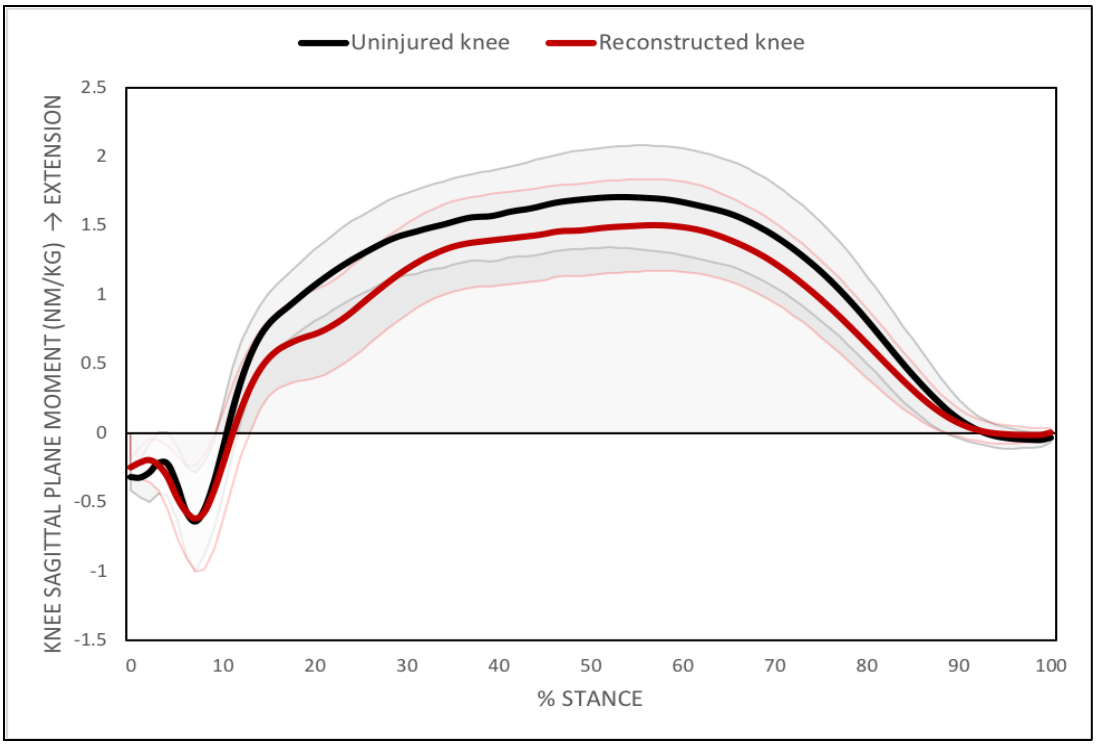



*Figure 8-7: Knee sagittal plane moment plus 95% confidence interval (shaded areas) for both legs during cutting to 90 degrees task.*

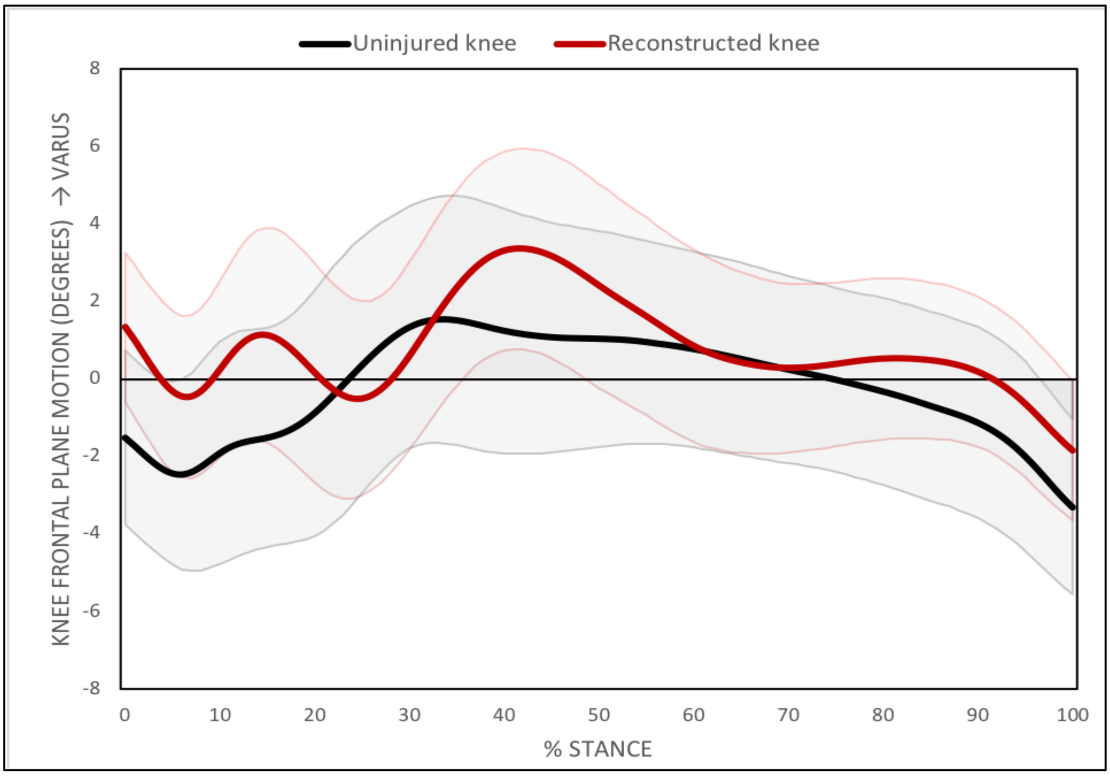



*Figure 8-8: Knee frontal plane motion plus 95% confidence interval (shaded areas) for both legs during cutting to 90 degrees task.*

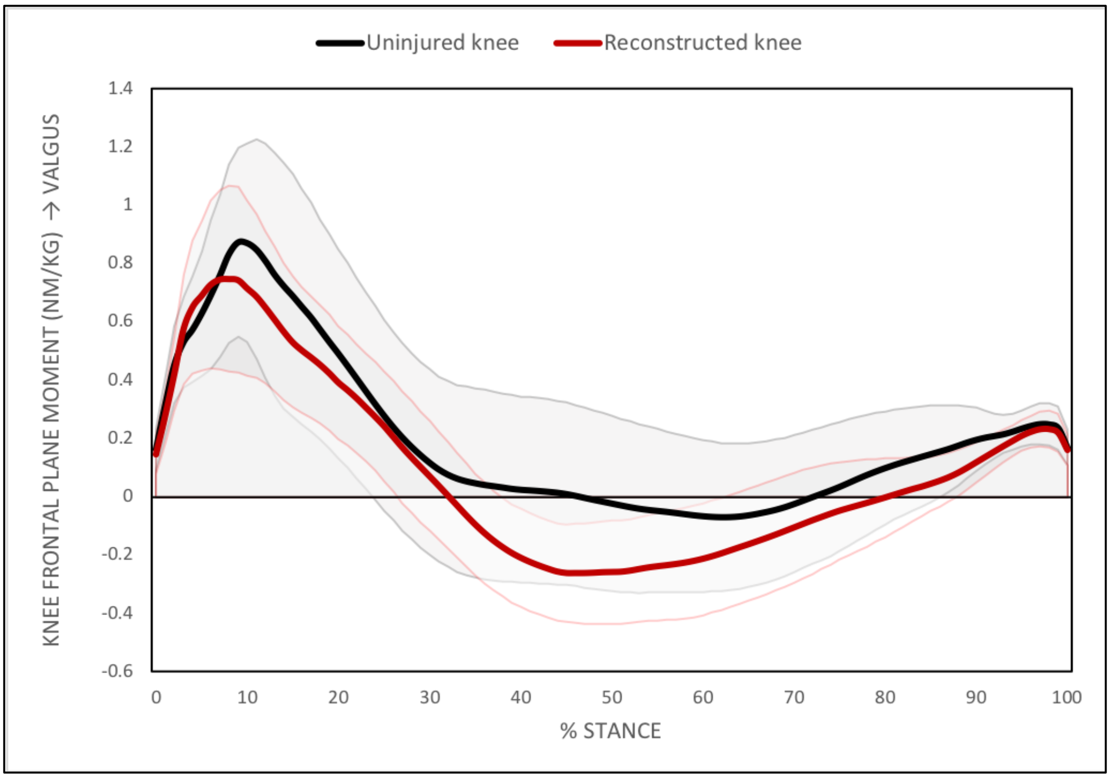



*Figure 8-9: Knee frontal plane moment plus 95% confidence interval (shaded areas) for both legs during cutting to 90 degrees task.*

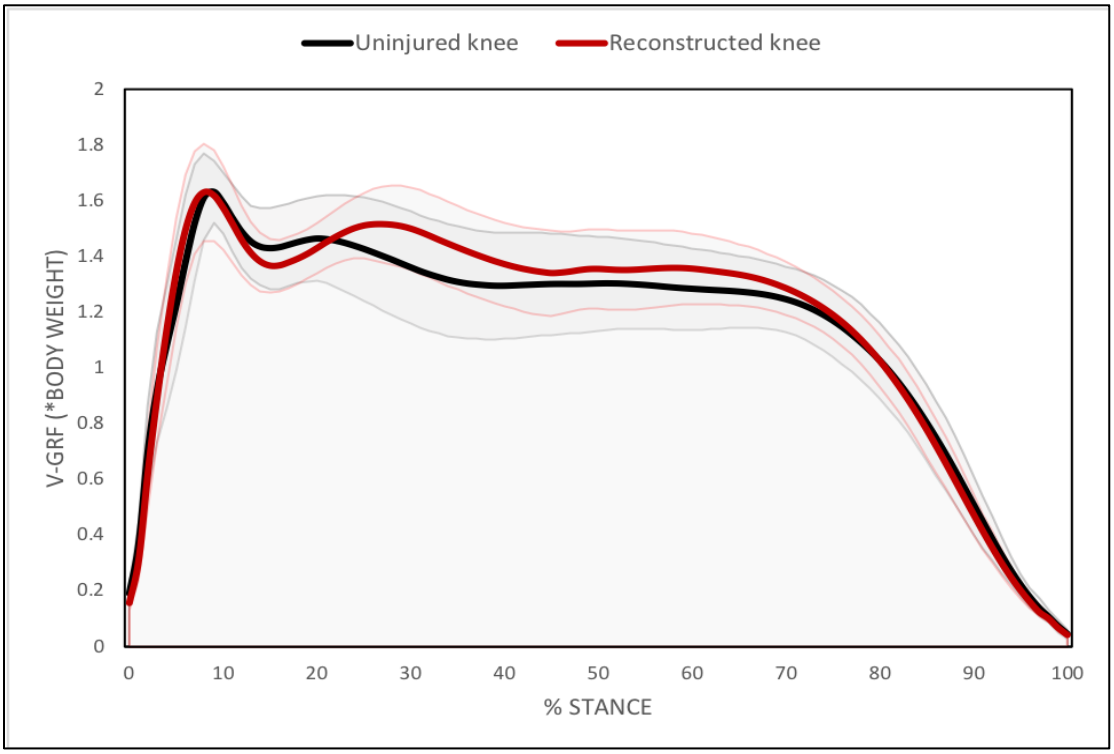



*Figure 8-10: Vertical ground reaction force plus 95% confidence interval (shaded areas) for both legs during cutting to 90 degrees task.*

### 8.4.3 SLL

All participants (n=20) performed SLL on the force plate with both legs (ACL-RC/uninjured). Knee kinematics and kinetics are outlined in Table 8-5. MV values for each biomechanical variable are also presented in Table 8-5. In general, there were four significant differences between the reconstructed and the uninjured limbs reported during this task. Firstly, a significant difference occurred in the peak knee extension moment, where the reconstructed limb demonstrated a lower peak knee extension moment than the uninjured one ($p = 0.01$; Es = 0.60). Another significant difference was reported in the peak knee valgus moment, where the reconstructed limb demonstrated a lower peak knee valgus moment than the uninjured one ($p = 0.001$; Es = 0.98). There was also a significant difference in MV for the peak knee valgus moment between the two limbs, as the reconstructed leg showed greater MV than the uninjured one ($p = 0.001$; Es = 1.10). The last significant difference during this task was found to occur in MV for the peak knee valgus angle, as the reconstructed leg showed greater peak MV than the uninjured one ($p = 0.05$; $Es = 0.60$). However, the latter was not significant following Bonferroni correction (corrected $\alpha = 0.01$). In addition, there was a difference with moderate effect size ($Es = 0.42$) in MV for the peak knee extension moment, where the reconstructed leg showed greater MV than the uninjured one. However, this difference was not significant ($p = 0.22$).

In addition, four figures were provided to illustrate the knee motion and moment curves, which includes the 95% confidence interval for both legs in the sagittal and frontal planes throughout the entire (100%) defined phase of the SLL task. They are as follows: Figure 8-11 displays the knee sagittal plane motion; Figure 8-12 displays the knee sagittal plane moment; Figure 8-13 displays the knee frontal plane motion; and Figure 8-14 displays the knee frontal plane moment. The V-GRF is also illustrated in Figure 8-15. All graphs presented in these figures represent the average scores obtained for all participants (n=20).

*Table 8-5: Knee kinematics, kinetics (peaks) and MV in each variable during SLL.*

| Variable | Injured leg | | Uninjured leg | | P-value | Effect size |
|---|---|---|---|---|---|---|
| | Mean | SD | Mean | SD | | |
| Knee flexion angle (°) | 66.06 | 12.53 | 66.33 | 13.89 | 0.93 | 0.02 |
| MV for knee flexion angle | 0.07 | 0.05 | 0.06 | 0.03 | 0.88 | 0.24 |
| Knee extension moment (Nm/Kg) | 2.62 | 0.63 | 2.96 | 0.50 | 0.01* | 0.60 |
| MV for knee extension moment | 0.09 | 0.06 | 0.07 | 0.03 | 0.22 | 0.42 |
| Knee valgus angle (°) | -0.27 | 3.92 | -1.65 | 3.97 | 0.11 | 0.35 |
| MV for knee valgus angle | 0.57 | 0.34 | 0.39 | 0.25 | 0.05 | 0.60 |
| Knee valgus moment (Nm/Kg) | 0.29 | 0.25 | 0.72 | 0.57 | 0.001* | 0.98 |
| MV for knee valgus moment | 0.59 | 0.17 | 0.38 | 0.21 | 0.001* | 1.10 |
| V-GRF (*Body weight) | 2.05 | 0.25 | 2.03 | 0.24 | 0.58 | 0.08 |
| MV for V-GRF | 0.01 | 0.03 | 0.01 | 0.02 | 0.43 | 0.00 |

*Negative (-) represents knee valgus angle and knee varus moment.*

*Significant difference between reconstructed and uninjured legs following Bonferroni correction ($p \leq 0.01$).

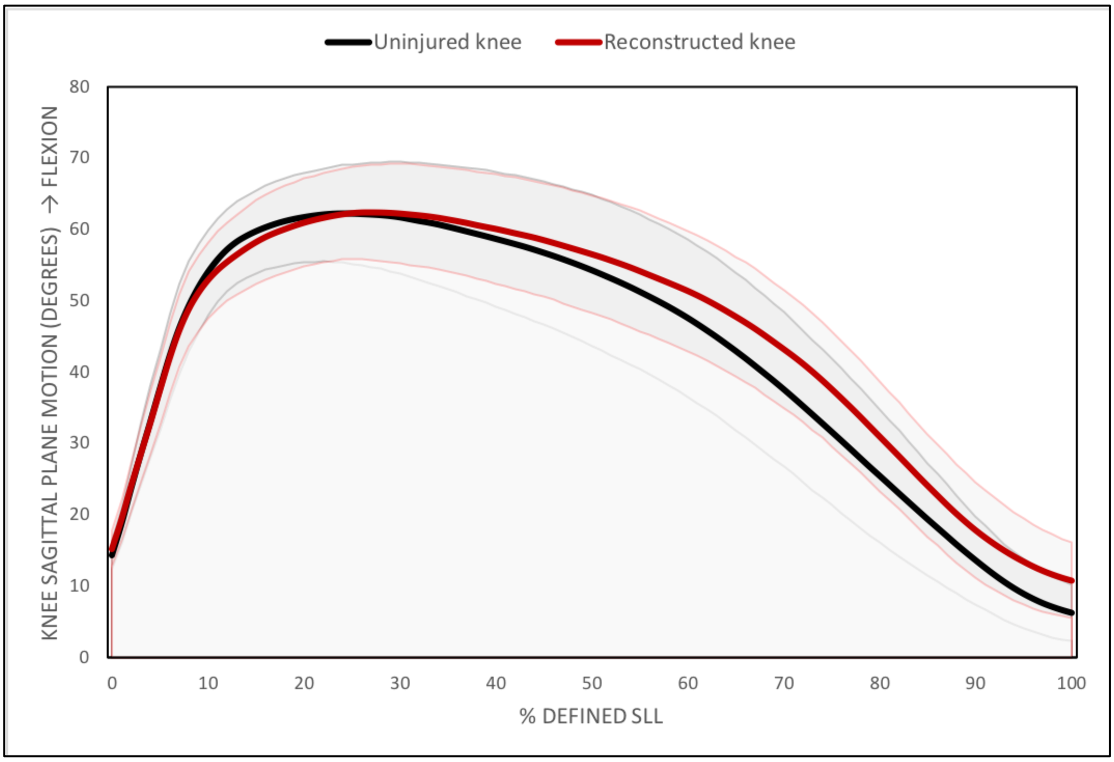



*Figure 8-11: Knee sagittal plane motion plus 95% confidence interval (shaded areas) for both legs during SLS task.*

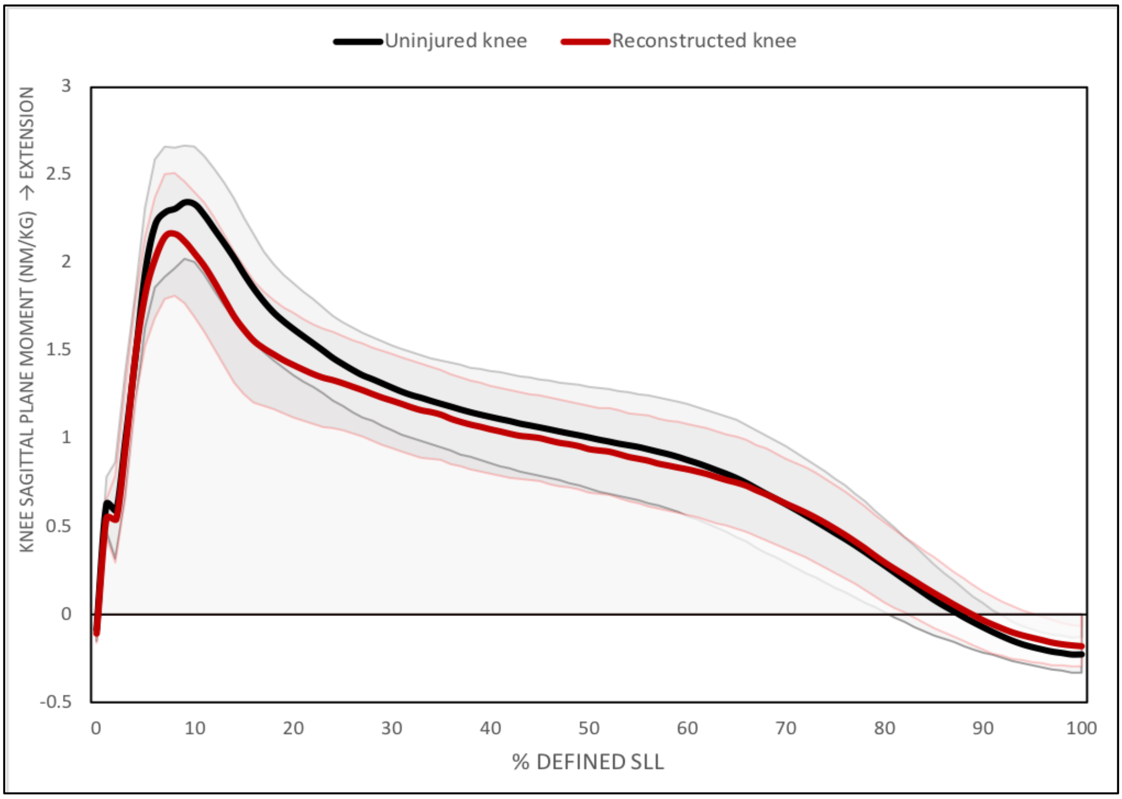



*Figure 8-12: Knee sagittal plane moment plus 95% confidence interval (shaded areas) for both legs during SLS task.*

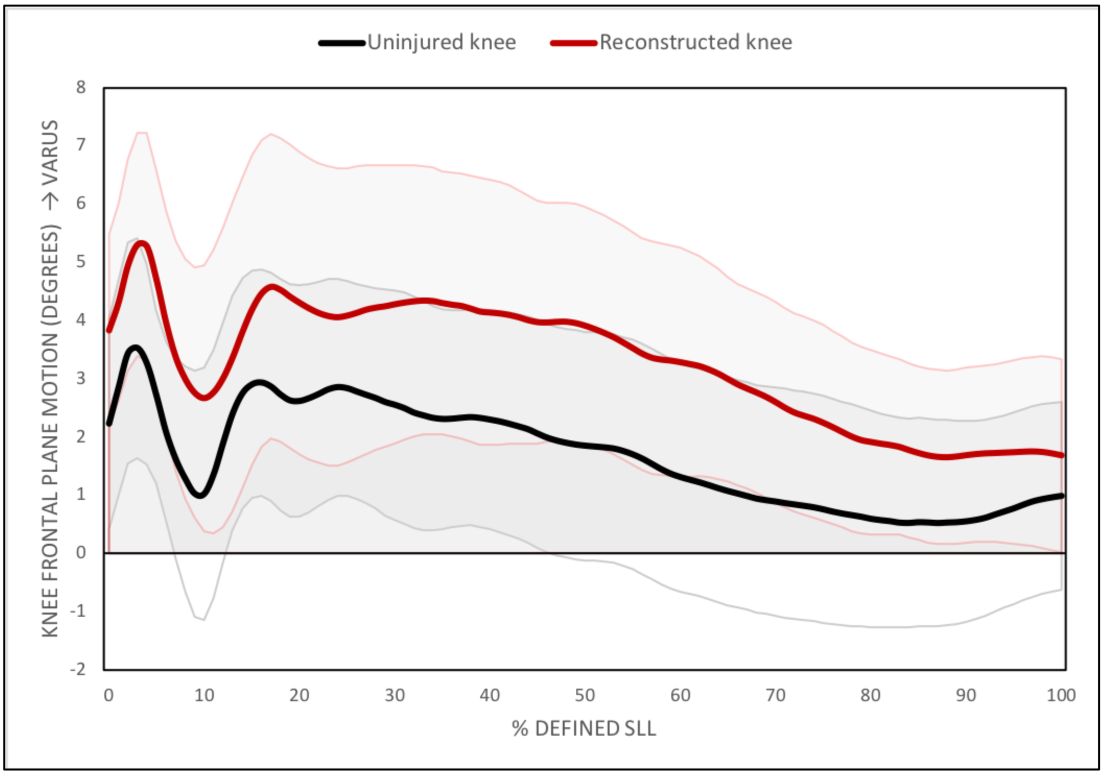



*Figure 8-13: Knee frontal plane motion plus 95% confidence interval (shaded areas) for both legs during SLS task.*

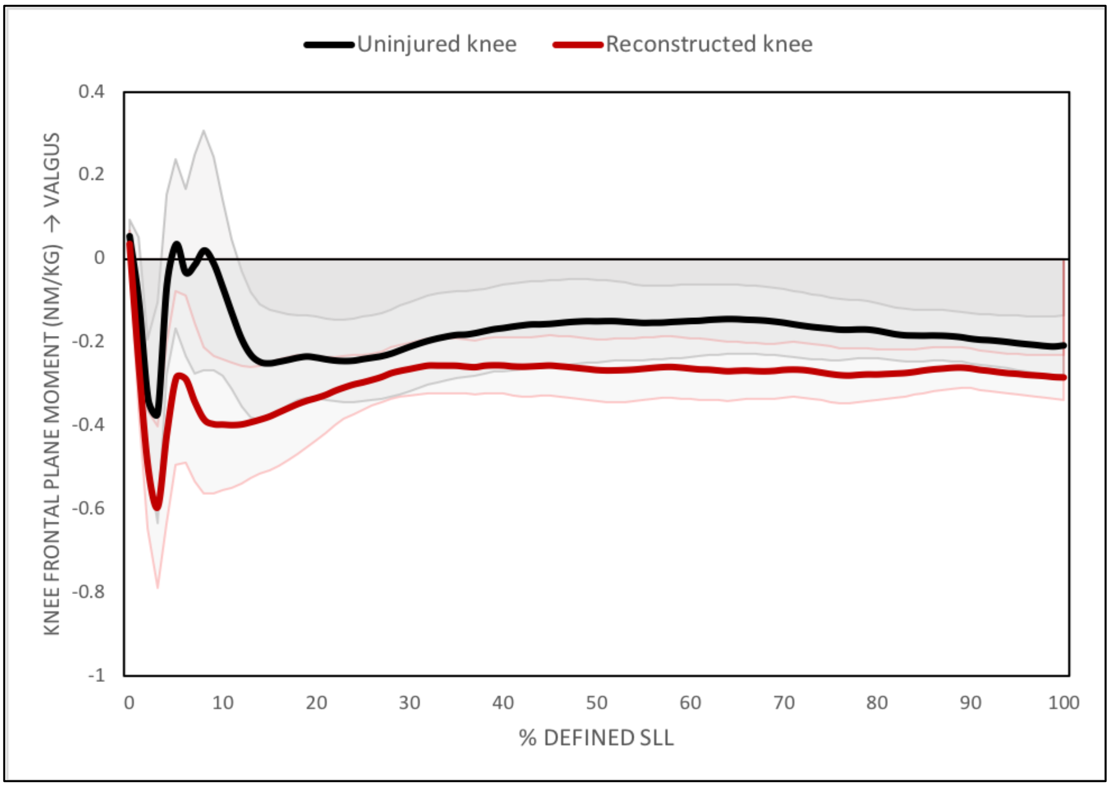



*Figure 8-14: Knee frontal plane moment plus 95% confidence interval (shaded areas) for both legs during SLS task.*

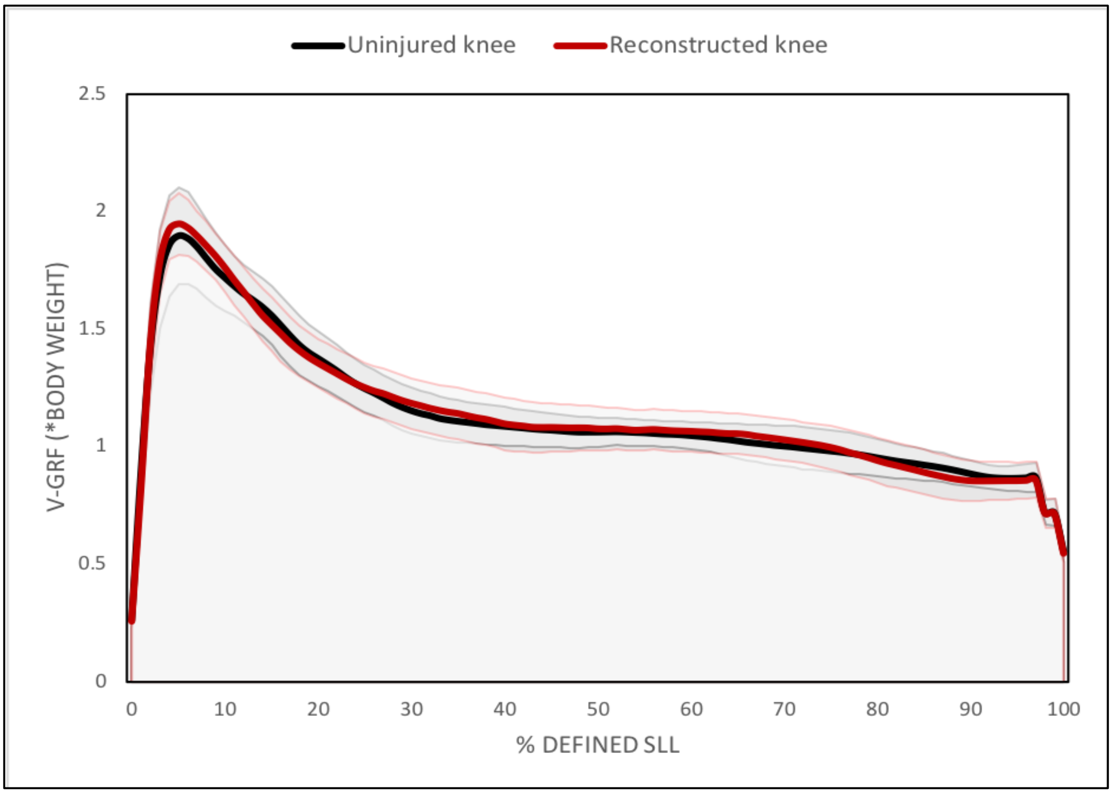



*Figure 8-15: Vertical ground reaction force plus 95% confidence interval (shaded areas) for both legs during SLS task.*

### 8.4.4 SLS

Each participant in this study (n=20) performed SLS on the force plate with both legs (ACL-RC/uninjured). Knee kinematics and kinetics for this task are shown in Table 8-6. MV values for each variable are also outlined in Table 8-6. Overall, of all the variables, two variables reported significant differences between the reconstructed and uninjured legs. Firstly, these occurred in the peak knee valgus moment ($p = 0.01$; Es = 0.79), where the reconstructed limb demonstrated a lower peak knee valgus moment than the uninjured one. The other significant difference occurred in the peak V-GRF, as the reconstructed limb demonstrated less peak V-GRF than the uninjured one ($p = 0.04$; Es = 0.27). However, the latter was not significant following Bonferroni correction (corrected $\alpha = 0.01$).

Furthermore, there was a difference with moderate effect size (Es = 0.63) in MV for the peak knee flexion angle, where the reconstructed leg demonstrated greater MV for the peak knee flexion angle than the uninjured one. Similarly, there was a difference with moderate effect size (Es = 0.43) in MV for the peak knee valgus moment, where the reconstructed leg showed greater MV for the peak knee valgus moment than the uninjured one. However, these differences were not significant ($p > 0.05$).

Furthermore, four figures were drawn to illustrate the knee motion and moment curves, which includes the 95% confidence interval for both legs in the sagittal and frontal planes throughout the entire (100%) defined phase of the SLS. They are as follows: Figure 8-16 displays the knee sagittal plane motion; Figure 8-17 displays the knee sagittal plane moment; Figure 8-18 displays the knee frontal plane motion; and Figure 8-19 displays the knee frontal plane moment. The V-GRF is also illustrated in Figure 8-20. All graphs presented in these figures represent the average scores obtained for all participants (n=20).

*Table 8-6: Knee kinematics, kinetics (peaks) and MV in each variable during SLS.*

| Variable | Injured leg | | Non-injured leg | | P-value | Effect size |
|---|---|---|---|---|---|---|
| | Mean | SD | Mean | SD | | |
| Knee flexion angle (°) | 82.42 | 12.85 | 84.73 | 8.65 | 0.32 | 0.21 |
| MV for knee flexion angle | 0.04 | 0.01 | 0.03 | 0.02 | 0.06 | 0.63 |
| Knee extension moment (Nm/Kg) | 1.15 | 0.43 | 1.26 | 0.46 | 0.12 | 0.25 |
| MV for knee extension moment | 0.18 | 0.16 | 0.21 | 0.14 | 0.26 | 0.20 |
| Knee valgus angle (°) | 0.67 | 2.87 | -0.64 | 3.08 | 0.09 | 0.44 |
| MV for knee valgus angle | 0.42 | 0.29 | 0.40 | 0.30 | 0.76 | 0.07 |
| Knee valgus moment (Nm/Kg) | 0.03 | 0.04 | 0.08 | 0.08 | 0.01* | 0.79 |
| MV for knee valgus moment | 0.61 | 0.23 | 0.50 | 0.28 | 0.11 | 0.43 |
| V-GRF (*Body weight) | 1.12 | 0.07 | 1.14 | 0.08 | 0.04 | 0.27 |
| MV for V-GRF | 0.03 | 0.02 | 0.03 | 0.01 | 0.70 | 0.00 |

*Negative (-) represents knee valgus angle and knee varus moment.*

**Significant difference between reconstructed and uninjured legs following Bonferroni correction ($p \leq 0.01$).*

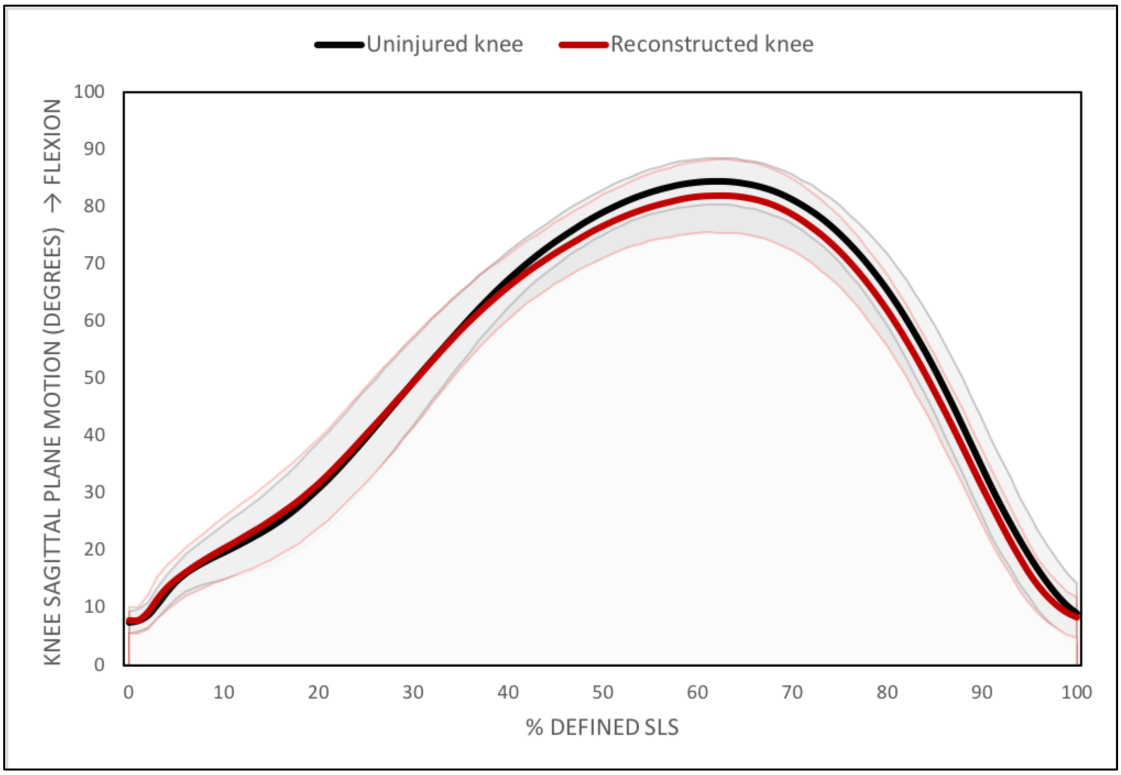



*Figure 8-16: Knee sagittal plane motion plus 95% confidence interval (shaded areas) for both legs during SLS task.*

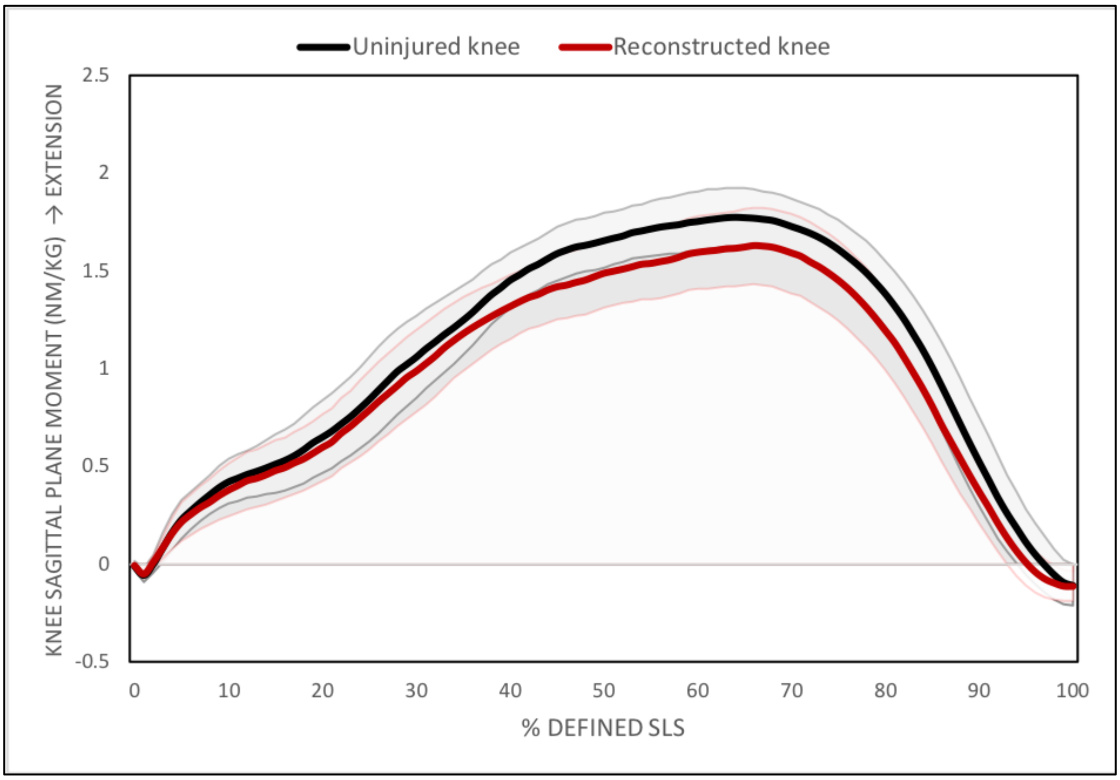



*Figure 8-17: Knee sagittal plane moment plus 95% confidence interval (shaded areas) for both legs during SLS task.*

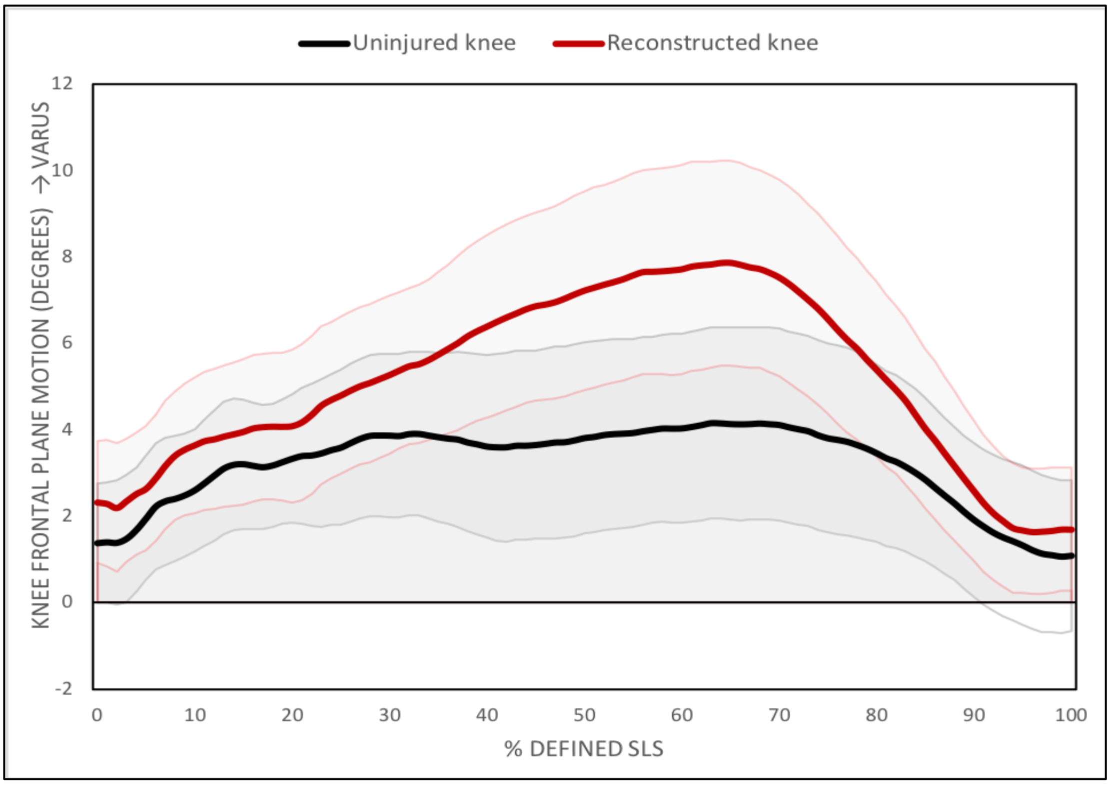



*Figure 8-18: Knee frontal plane motion plus 95% confidence interval (shaded areas) for both legs during SLS task.*

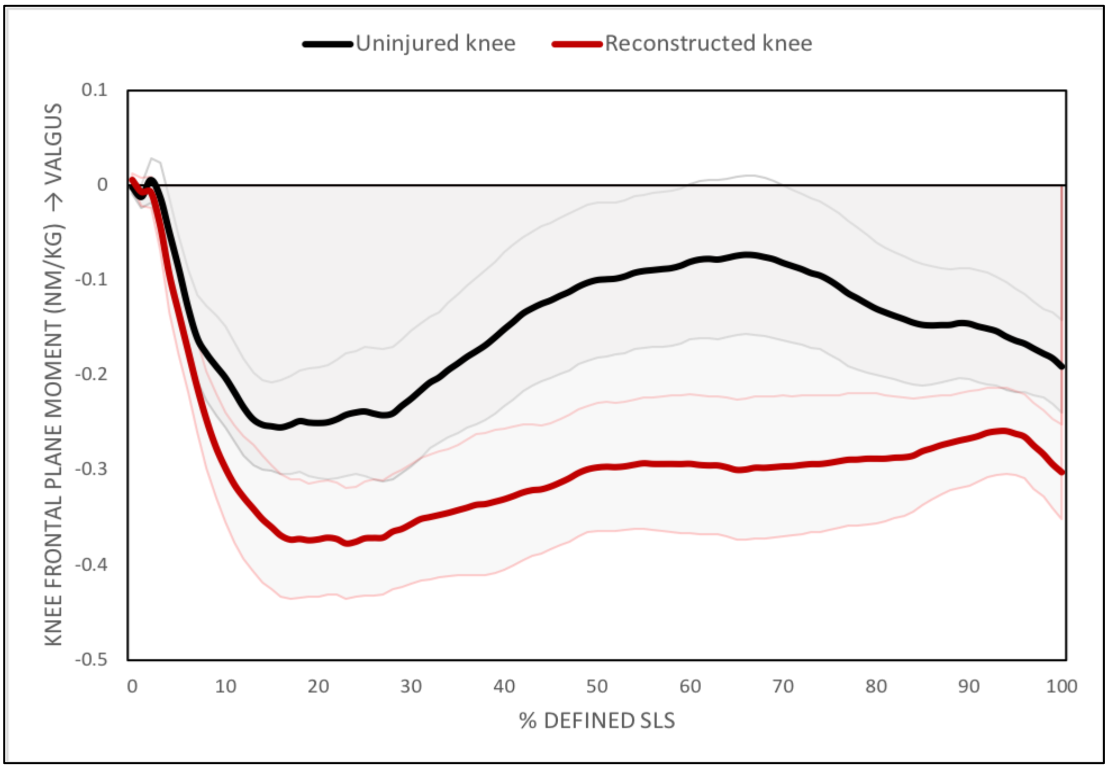



*Figure 8-19: Knee frontal plane moment plus 95% confidence interval (shaded areas) for both legs during SLS task.*

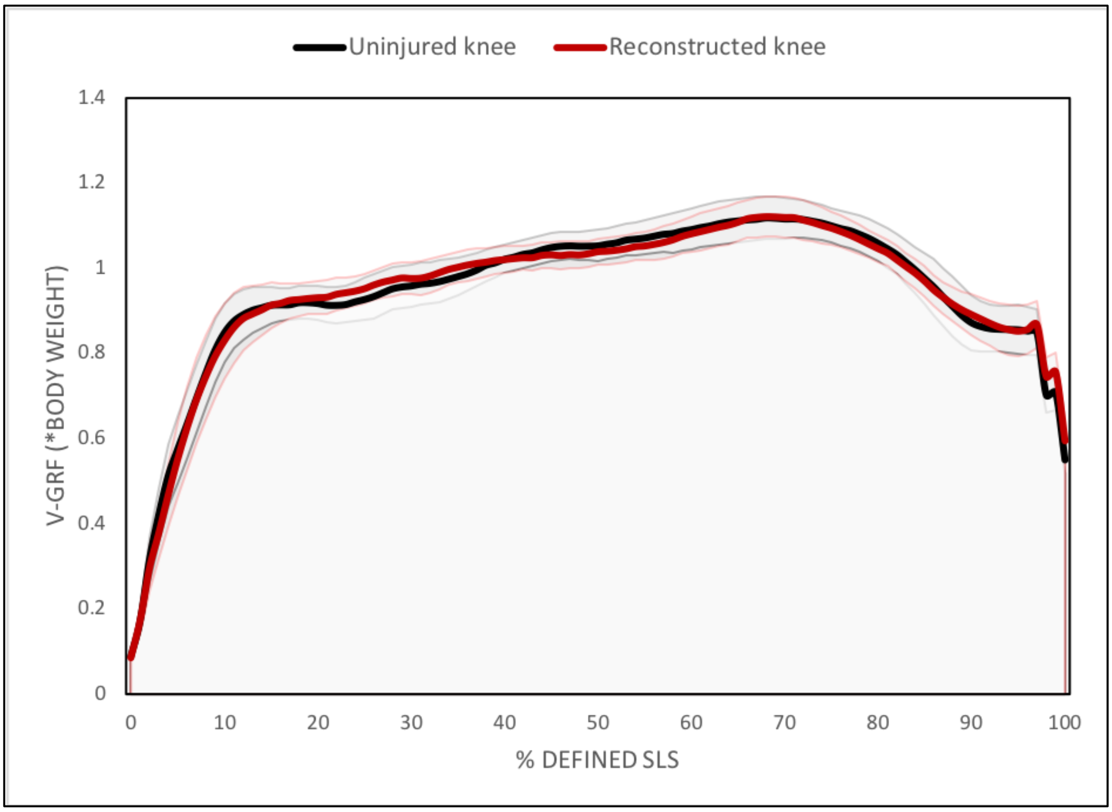



*Figure 8-20: Vertical ground reaction force plus 95% confidence interval (shaded areas) for both legs during SLS task.*

## 8.5 Discussion

The main aims of this study were to investigate MV and knee kinematics and kinetics in an ACL-RC population during four common sporting tasks. It also aimed to compare MV and knee kinematics and kinetics between reconstructed and uninjured legs during these four tasks. In general, the results showed that there was a significant difference between reconstructed and uninjured legs in at least one variable in each task. SLL showed the highest number (3) of variables reporting significant differences between the two legs, as compared to the other tasks. Beginning with the biomechanical variables, the reconstructed leg demonstrated a significantly lower peak knee extension moment during three tasks, namely running, cutting to 90 degrees and SLL ($p \leq 0.01$), than the uninjured one. This finding is consistent with a number of earlier studies which investigated knee extension moment during different tasks: running (Bowersock, Willy, DeVita, & Willson, 2017; Herrington et al., 2017; Noehren, Abraham, Curry, Johnson, & Ireland, 2014; Pamukoff et al., 2017; Pratt & Sigward, 2017); SLL (Pozzi et al., 2017; Gokeler et al., 2010; Pratt & Sigward, 2017; Webster, Gonzalez-Adrio, & Feller, 2004; Webster, Santamaria, McClelland, & Feller, 2012); and cutting to 90 degrees (King et al., 2018). In all cases, they reported a significant decrease in extension moment in the ACL-RC knee, as compared to the uninjured limb or a healthy control group.

The explanation as to why the ACL-RC limb demonstrates lower extension moment is more likely to be due to weakness of the quadriceps muscle, as these muscular contractions largely represent the knee extension moment (Shimokochi, Yong Lee, Shultz, & Schmitz, 2009). In addition, it has been reported that persistent quadriceps weakness and inhibition are common in both ACL-DF and ACL-RC populations (Ingersoll et al., 2008). Unfortunately, this may lead to early-onset degeneration at the knee joint if left untreated (Roos, 2005). In order to treat this deficit, there must be a specific emphasis placed on restoring quadriceps muscle strength during the rehabilitation programme. This will help to normalise moment at the knee joint. However, the SLS task did not report a significant difference in the peak knee extension moment, as this activity was less complex and less stressful as compared to the other tasks.

In addition, the findings of this study showed that the reconstructed knee demonstrated a lower peak knee flexion angle than the uninjured leg, in both running and cutting tasks ($p = 0.008$; $Es = 0.57$; $p = 0.04$; $Es = 0.38$, respectively). Similar results were reported in previous research during running (Pairot-de-Fontenay et al., 2019; Herrington et al., 2017; Pratt & Sigward, 2017; Saxby et al., 2016; Noehren et al., 2014); and cutting to 90 degrees (King et al., 2018). One explanation for the demonstration of less knee flexion during running and cutting tasks could be attributed to quadriceps deficits, as a significant decrease occurred in the knee extension moment. Unfortunately, a decreased peak knee flexion angle has been reported as a risk factor during ACL injury mechanisms (Leppänen et al., 2017). It has also been reported that it can lead to increased anterior tibial shear (Li, DeFrate, Rubash, & Gill, 2005; K. L. Markolf et al., 1995). Therefore, maintaining the reconstructed knee in a more extended position could increase the risk that the knee joint will develop another ACL injury on the ipsilateral side, especially during cutting tasks. Therefore, it is important to improve knee flexion when undertaking a rehabilitation programme.

In contrast, SLL and SLS did not show significant differences in the peak knee flexion angle between the reconstructed limb and the uninjured one. Previous research also reported no significant differences during SLL (Pozzi et al., 2017; Webster & Feller, 2012b; Webster et al., 2004, 2012; Ortiz et al., 2008) and SLS (Salem et al., 2003). However, one study did report significant differences in the knee flexion angle. Antolič et al. (1999) found that this occurred during SLL. One explanation as to why these different results emerged could be because Antolič et al. (1999) did not segregate male and female data and there were more females (n=14) taking part than males (n=11). Furthermore, gender differences in biomechanics during this task have been reported in several recent studies (Hughes, 2019; Weinhandl, Irmischer, Sievert, & Fontenot, 2017; Holden, Boreham, & Delahunt, 2016; Jenkins et al., 2017).

In addition, the reconstructed leg showed a lower peak V-GRF during the SLS task ($p = 0.04$; $Es = 0.27$). However, this difference was not significant following Bonferroni correction (corrected $\alpha = 0.01$). Similar results were reported by Sanford et al. (2016) who noted that there were significant differences in GRF between the reconstructed and the uninjured limb during SLS. One explanation is that ACL-RC subjects may adopt different weight-bearing strategies in order to try to reduce the stress across the knee joint. In contrast, a study by Salem et al.

(2003) did not report a significant difference between the two limbs. That is because they employed a different squatting technique, comprising a back squat exercise based on resistance training, using 35% body weight. However, no significant differences emerged in the other three tasks in terms of peak V-GRF. Similar results were reported in previous research during running (Bowersock et al., 2017) and SLL tasks (Pozzi et al., 2017) tasks.

With regard to the knee valgus, the results of this study showed that there was a significant difference in knee valgus moments between the two limbs during two tasks, namely SLL and SLS. In both tasks, the uninjured limb showed greater valgus moment than the reconstructed knee ($p = 0.001$; Es = 0.98; $p = 0.01$; Es = 0.79, respectively). One explanation for this could be that the impairments in the ACL-RC knee biomechanics may exert an influence on the contralateral knee by affecting motor control, thus resulting in altered biomechanics, such as increased valgus moment. Therefore, the findings of the current study are consistent with previous research which focused specifically on landing tasks (King et al., 2018). However, a study by Ortiz et al. (2008) reported no significant difference in knee valgus moment during SLL. One possible reason why different findings have emerged could be because Ortiz et al. (2008) conducted their study on females, whereas the current study selected a male cohort. In addition, it could be because of the small sample size used (n=13). With regard to the SLS task, unfortunately we could not compare our findings with previous research undertaken, as none of the studies examining SLS included a knee valgus angle/moment in their investigations.

Such findings whereby the uninjured limb shows a greater knee valgus moment than the reconstructed limb, highlights a very important point. In fact, increased knee valgus moment is a risk factor for ACL injury (Hewett et al., 2005; Myer et al., 2015; Numata et al., 2018), which implies that the uninjured limb is of greater risk of sustaining an ACL injury than the reconstructed limb. Several studies have reported a high incidence of ACL reinjury on the contralateral side (Webster & Hewett, 2019; Dekker et al., 2017; Morgan et al., 2016; Paterno et al., 2014; Wiggins et al., 2016). This is a potential cause of reinjury on the contralateral side. Therefore, clinicians and researchers should take this into consideration when developing and delivering ACL-RC rehabilitation programmes. Importantly, the contralateral limb should be specifically targeted during this process.

Several points need to be taken into consideration regarding the aforementioned studies which investigated ACL-RC individuals during the four sporting tasks. Firstly, few authors have investigated all knee kinematics and kinetics associated with the risk of ACL injury, with the exception of: King et al. (2018) and Webster et al. (2012), (SLL); Bowersock et al. (2017), Milandri et al. (2017), (running); and King et al. (2018), (cutting). In addition, only four studies, namely Bowersock et al. (2017), Perraton et al. (2018) and Pratt et al. (2017) (running) and Webster et al. (2012) (SLL) undertook power calculations to determine the sample size, while the remaining authors failed to do so. Adequate samples sizes are necessary as small samples bias the results and inflate type 1 errors, and they are also needed for multiple statistical tests of dependent variables (Knudson, 2017). Unfortunately, some authors included small sample sizes, without providing sufficient justification for this decision.

Gender is another important point that should be taken into consideration. The majority of the aforementioned studies included both males and females. However, only five studies investigated one gender only. Of these, three included males: SLL (King et al., 2018; Webster et al., 2012); cutting (King et al., 2018); and running (Milandri et al., 2017), while one included females (Noehren et al., 2014). It is important to segregate the data by gender in cohorts comprising both males and females, as significant gender differences in terms of biomechanical functioning have been reported during these tasks in several recently conducted studies, as discussed in Chapter 4. Unfortunately, all the aforementioned studies that involved both genders failed to do so. Notably, some of them had large differences in female-male participation ratios: 1:7 (Salem et al., 2003); 3:5 (Sanford et al., 2016); 2:18 (Webster et al., 2004); 32:3 (Webster et al., 2012a); and 10:24 (Herrington et al., 2017).

Furthermore, another important consideration is the need to specify the time period post-surgery in which the testing took place. In the majority of studies, the average time was between 6 to 12 months. However, four studies investigated ACL-RC individuals who were two years or more post-surgery (Bowersock et al., 2017; Milandri et al., 2017; D N Pamukoff et al., 2017; Sanford et al., 2016). In contrast, Pratt and Sigward (2017) studied ACL-individuals within 4.6 ± 1.4 months, which suggests that they might have tested some participants while they were still engaging in rehabilitation programmes. These post-surgery time differences can affect biomechanical functioning, as previous research has reported that the time frame following ACL reconstruction can result in differences in ACL-RC subjects' loading responses in both their uninjured and reconstructed limbs (Neitzel et al., 2002). Post-surgery time was also highlighted in a very recent systematic review by Webster and Hewett (2019), where they argued that if a player has made a successful return to sport and has performed for at least two full seasons, it may not be meaningful to link an injury that occurs after this time point back to the injury that was sustained several years earlier.

In terms of MV, the results have shown that there were significant differences between the reconstructed limb and the uninjured one. This was reported during the SLL task, where of the two limbs, the reconstructed limb demonstrated greater MV for the peak knee valgus moment ($p = 0.001$; $Es = 1.10$). Similar results were reported during the same task in terms of MV for the peak knee valgus angle, where the reconstructed leg also showed significantly greater peak MV than the uninjured one ($p = 0.05$; Es = 0.60). In addition, there was a difference with moderate effect size (Es = 0.42) in MV for the peak knee extension moment during SLL, where the reconstructed leg showed greater MV than the uninjured one. Similarly, there was a difference with moderate effect size (Es = 0.44) in MV for the peak knee valgus moment during the cutting to 90 degrees task, where the reconstructed leg showed a greater MV for the peak knee valgus moment than the uninjured one. However, these differences were not significant ($p > 0.05$).

Running and SLS also reported differences in MV between the reconstructed and uninjured legs. In particular, a difference with moderate effect size (Es = 0.63) in MV for the peak knee flexion angle, where the reconstructed leg demonstrated greater MV for the peak knee flexion angle than the uninjured one. Similarly, there was a difference with moderate effect size (Es = 0.43) in MV for the peak knee valgus moment during the same task, where the reconstructed leg showed greater MV for the peak knee valgus moment than the uninjured one. However, these differences were not significant ($p > 0.05$). Based on these findings, clearly an overall trend has emerged in terms of differences whereby the reconstructed limb demonstrated greater MV than the uninjured one.

In the literature, few studies have investigated MV in ACL-RC individuals while performing different tasks. The findings of the current study are consistent with previous studies which have investigated MV in ACL-RC individuals during running and walking tasks (Gribbin et al., 2016b); cutting to 45 degrees (Pollard et al., 2015b); and single leg hopping (Van Uden et al., 2003). However, they adopted different methods to measure MV, based on the dynamical systems theory, which was explained in Chapter 2 (see Section 2.11.4.3). They all reported that the reconstructed limb showed greater MV than healthy controls for most of the variables. However, Pollard et al. (2015b) included only females in their study and the cutting angle was 45 degrees. In addition, to date no study has investigated MV in ACL-RC individuals during SLL and SLS tasks. The reason why ACL-RC limbs demonstrate greater MV than uninjured limbs could be due to insufficient rehabilitation, resulting in inadequate neuromuscular control, and which, ultimately, leads to poorly controlled movement (Schmidt, 2003). A very recent systematic review conducted by Webster and Hewett (2019) revealed that only a low proportion of ACL-RC individuals (23%) had passed RTS test batteries. The paper also reported that while 43% had failed, nevertheless, they had subsequently returned to sport, which means that they may still exhibit poorly controlled movement and this could lead to impairment in MV in the reconstructed knee. Another supporting point for this is that athletes who pass a criterion at one time point may subsequently fail it at another juncture (Van Melick et al., 2016). Therefore, some ACL-RC individuals may still have poorly controlled movement, which will result in abnormal MV.

Another possible explanation, which is less convincing, is that the ACL reconstruction failed to improve the injured limb biomechanics, as some deficits in knee biomechanics still remain unchanged pre- and post-surgery, such as greater MV and decreased knee extension moment. Despite the functional stability that is provided by surgery, debate continues to surround the optimal treatment of this injury (Smith et al., 2014). In fact, there is a need to enhance the rehabilitation programmes on offer, as well as the RTS criteria for ACL-RC individuals. Improving these biomechanical deficits will lead to a decrease in reinjury rates and will help to ensure that ACL-RC patients can return to sport safely. Clinicians should place greater emphasis on neuromuscular control exercises, which can help to decrease biomechanical deficits and subsequently reduce the risk of reinjury. Another possible explanation could be that greater MV has been adopted as a compensatory movement in order to minimise the loading on painful tissues (Hodges & Tucker, 2011). This greater variability could create a risk of further damage or injury if it is sustained over time and if it remains untreated (Hamill, Palmer, & Van Emmerik, 2012; Stergiou, Harbourne, & Cavanaugh, 2006). The best exercise option in this case would be to target neuromuscular control.

## 8.6 Clinical implications

A number of different clinical implications have been identified in this study. Firstly, clinicians and researchers should be aware of the significant differences in some knee biomechanical functions between the ACL-RC side and the uninjured one. Most importantly, they should also know that the knee extension moment is significantly lower in the reconstructed limb, which indicates that a weakness can occur in the quadriceps muscle and this should be targeted when implementing a rehabilitation programme. In addition, the ACL-RC side showed a lower peak knee flexion angle than the uninjured one during running and cutting tasks. Improving knee flexion during functional tasks should be targeted so as to avoid further damage or injury, which may occur due to altered knee flexion. Clinicians should be aware that the ACL-RC side demonstrated greater MV than the uninjured one, which could be attributable to a reduction in neuromuscular control. Hence, practitioners should include exercises which can improve this deficit.

## 8.7 Study novelty

As highlighted throughout this chapter, this novel study is the first of its kind to investigate MV among individuals with ACL-RC during SLL and SLS tasks, as well as among a male ACL-RC cohort during a cutting task. In addition, it investigated knee frontal plane biomechanics among individuals with ACL-RC during SLS. Another novel aspect of this study is that it is the first to compare MV between an ACL-RC limb and the contralateral side during functional sporting tasks.

## 8.8 Limitations of this study and future work

The current study has some limitations. Firstly, the investigations in this study were limited to males only, as well as solely to an ACL-RC population. Future studies are needed in order to carry out such investigations in an ACL-RC female cohort, where it is highly expected that similar results would be demonstrated. New research is also warranted to support the development of a rehabilitation programme for ACL-RC patients, which will help to reduce all these biomechanical impairments and facilitate their safe return to sport. Furthermore, new research is needed to investigate whether a relationship exists between MV and muscle strength.

## 8.9 Conclusion

The ACL-RC knee demonstrated significant differences in both MV and knee biomechanics as compared to the uninjured limb. However, these findings were not reported in all variables. The ACL-RC limb demonstrated a significantly lower peak knee extension moment as compared to the uninjured side. It also showed a lower peak knee flexion angle during both running and cutting tasks. However, the ACL-RC limb demonstrated greater MV than the uninjured one. These findings may be of assistance in the development of rehabilitation and RTS programmes for ACL-RC patients.

Having conducted these investigations, we have gained a full understanding of MV and knee biomechanics among an ACL-RC population. In addition, differences in both MV and knee biomechanics between the reconstructed and uninjured limbs have been identified. However, other research questions still require addressing, namely to compare MV and knee biomechanics in three populations related to ACL injury (ACL-RC, ACL-DF and healthy). This topic will be also investigated in the next chapter.

# Chapter 9

## 9 Differences in Movement Variability and Knee Biomechanics Between Healthy, ACL-DF and ACL-RC Individuals During Four Common Sporting Tasks

### 9.1 Introduction

In previous chapters three populations were independently investigated in relation to biomechanical risk factors for ACL injury. Significant differences were found to occur in both MV and knee biomechanics between the two limbs in ACL-DF and ACL-RC populations. However, no significant differences were reported between the two limbs in the healthy cohort. Undertaking a comparative analysis between the three populations performs a crucial role in gaining a more in-depth understanding of MV and knee biomechanics within and between groups. Such investigations can answer very important questions regarding this injury, in that they highlight differences in MV and biomechanics between the three groups. This can also help to clarify the way in which ACL injury and ACL reconstruction affects MV and knee biomechanics.

Several studies have investigated lower limb biomechanics in ACL-RC individuals during different sporting tasks. Some studies solely compared the reconstructed limb with the uninjured one, without including healthy controls (King et al., 2018; Bowersock et al., 2017; Pamukoff et al., 2017; Perraton et al., 2018; Webster et al., 2004; King et al., 2018). Other studies compared the reconstructed limb to healthy controls, without involving the uninjured limb in their investigations (Pozzi et al., 2017; Noehren et al., 2014). Furthermore, a number of studies compared the reconstructed limb to the uninjured limb, as well as to healthy controls (Herrington et al., 2017; Milandri et al., 2017; Neitzel et al., 2002; Pratt & Sigward, 2017; Sanford et al., 2016; Webster & Feller, 2012a; Webster et al., 2012; Antolič et al., 1999; Gokeler et al., 2010). Another important approach is to also compare the uninjured limb in

ACL-RC individuals to a healthy control group, as a high incidence of reinjury has been found to occur in the contralateral limb (Dekker et al., 2017; Webster & Hewett, 2019); however, only three studies conducted this type of investigation (Herrington et al., 2017; Pratt & Sigward, 2017; Antolič et al., 1999).

It has been reported that biomechanical differences exist between ACL-RC individuals and healthy controls during different functional tasks, for example: landing (King et al., 2018; Lepley & Kuenze, 2018; Schmitt et al., 2015); a cutting task (King et al., 2018); running (Pairot-de-Fontenay et al., 2019; Pamukoff et al., 2017; Pratt & Sigward, 2017; Herrington, Alarifi, & Jones, 2017); and squats (Sanford, Williams, Zucker-Levin, & Mihalko, 2016; Neitzel, Kernozek, & Davies, 2002). In addition, it has been reported that lower limb injured populations demonstrate greater MV but not in all cases (Baida et al., 2018) . This was also reported in the two previous chapters where the injured limb was compared to the uninjured one. In the literature, very few studies have investigated MV in an ACL-RC population during different tasks: running and walking (Gribbin et al., 2016b); cutting to 45 degrees (Pollard et al., 2015b); and single leg hopping (Van Uden et al., 2003). In general, in most of the variables, the reconstructed limb showed greater MV than the uninjured one.

To date, there is very limited evidence available regarding differences in MV between ACL-RC, healthy controls and ACL-DF. Furthermore, no study included healthy controls in their investigations during a cutting task. There is also limited evidence of the biomechanical risk factors associated with ACL injury when comparing the uninjured limb to healthy controls. Therefore, this study aims to investigate differences in both MV and knee biomechanics between the three groups, along with comparing the uninjured limb with healthy controls, as this provides very important information in terms of reinjury on the contralateral side.

## 9.2 Null hypothesis

1- $H0_1$: There are no significant differences in knee kinematics and kinetics between ACL-RC, ACL-DF and healthy populations during four common sporting tasks.

2- $H0_2$: There are no significant differences in MV between the ACL-RC, ACL-DF and healthy populations during these four tasks.

## 9.3 Methods

### 9.3.1 Participants

Three groups were included in this study: healthy, ACL-DF and ACL-RC. The inclusion and exclusion criteria for each group were explained in Chapter 4 (Section 4.12.4.2) for the healthy group, Chapter 7 (Section 7.3.1) for the ACL-DF group and Chapter 8 (Section 8.3.1) for the ACL-RC group. Power analysis was performed using post hoc G*Power (Version 3.1.9.4), which was based on the peak knee extension moment obtained from the results of this study during the cutting to 90 degrees task (Es = 0.32). The reason of the selection of this variable was because it is the main variable affected by this injury in both ACL-RC and ACL-DF as reported in the majority of previous studies that have investigated these two populations and compared them to a healthy control group. This will be discussed in further detail later in this chapter. The effect size of 0.32 was used to calculate the statistical power of the sample size utilised in this study (N=85), which was calculated electronically using G*Power software (see Appendix 12.8). The sample size (N=85) used in this research showed a statistical power of 92%, according to G*Power post hoc analysis, with the type 1 error alpha level set at 5% ($\alpha$ = 0.0) (Faul et al., 2007). The participants' demographic profile is presented in Table 9-1. All participants completed a consent form prior to engaging in the testing process.

*Table 9-1: Participants' demographic profiles.*

| | Healthy | ACL-DF | ACL-RC |
|---|---|---|---|
| Number (gender) | 45 (male) | 20 (male) | 20 (male) |
| Age (SD) | 25.8 (6.6) | 25.8 (5) | 26.15 (3.76) |
| Height (m) | 1.71 (0.06) | 1.72 (0.06) | 1.72 (0.07) |
| Body mass (kg) | 65.7 (10.3) | 71.6 (11.4) | 73.3 (10.33) |

### 9.3.2 Preparation and tasks undertaken

Each participant was required to do some general warm-up exercises of ten minutes' duration, followed by a further five-minute warm-up on an exercise bike, prior to beginning the test. The preparation process was explained in detail in Chapter 4 (Section 4.5). In terms of the tasks undertaken, all participants were invited to complete all four functional sporting tasks: SLS, running, cutting to 90 degrees and SLL with their two legs. Details of the requirements involved in performing these tasks were outlined in Chapter 4 (Section 4.5.1 for running, Section 4.5.2 for SLS, Section 4.5.3 for SLL and Section 4.5.4 for cutting to 90 degrees).

### 9.3.3 Procedure

The biomechanical data were collected for this study using 3D motion analysis at two laboratories. Firstly, the 3D motion analysis laboratory at the University of Salford that was described in detail in Chapter 4 (Section 4.2.1). Secondly, the 3D motion analysis laboratory conducted at Imam Abdulrahman Bin Faisal University that was outlined in detail in Chapter 4 (Section 4.2.2).

### 9.3.4 Statistical analysis

All statistical analysis in this study was carried out using SPSS (Version 24). Descriptive analysis (means and standard deviations) was performed. Prior to conducting the data analysis, its normality distribution was observed visually following application of a Shapiro-Wilk test. A variable is normally distributed if the p-value is greater than 0.05 on the latter test; however, it is not normally distributed if the p-value is equal to or less than 0.05. In addition, one-way ANOVA with post hoc tests were performed to assess differences between groups for normally distributed data that met the homogeneity assumption, while the modified version of the ANOVA (Welch ANOVA) was used for normally distributed data that did not meet the homogeneity assumption. However, the non-parametric Kruskal-Wallis H test was applied, in combination with post hoc tests, in cases where the normality assumption was violated or outliers were detected. For the healthy controls, dominant limb data were used in the comparisons, as the participants demonstrated no significant difference between the two legs, as shown in Chapter 5. In terms of MV values, a zero value indicates that no variability has occurred while, conversely, a score of one is indicative of the highest levels of variability.

In addition, in order to decrease the risk of type 1 errors occurring, a Bonferroni correction was applied to the alpha level used ($\alpha = 0.05$) for all comparison tests carried out, as there were multiple comparisons conducted in this study (corrected $\alpha = 0.005$). The effect size for each between group comparisons was also provided, along with p-values, in order to be able to fully interpret the results. The effect size was calculated using the partial eta squared ($\eta^2$) method, which involves calculating the effect size for the sample. Table 9-2 demonstrates the levels and classifications of effect size (Cohen, 2013).

*Table 9-2: Levels and classification of effect size*

| Effect size (Es) value | Classification |
|---|---|
| Es < 0.2 | Weak |
| 0.2 ≤ \| Es \|< 0.5 | Small |
| 0.5 ≤ \| Es \|< 0.8 | Moderate |
| \| Es \| ≥ 0.8 | Large |

(Cohen, 2013)

## 9.4 Results

### 9.4.1 Overground running

All participants in the three groups (n=85) performed the overground running task on the force plate with both legs. Knee kinematics and kinetics for the three groups are presented in Table 9-3. MV values for each variable are also shown in Table 9-3. The results of all post hoc tests (pairwise comparisons between groups) are displayed in Table 9-4. In general, three variables reported statistically significant differences between the groups during this task. The first one was the peak knee flexion angle, where the reconstructed leg showed a lower peak knee flexion angle than the control group ($p = 0.001$; Es = 0.10) and the ACL-DF uninjured leg ($p = 0.003$; Es = 0.10). However, these differences showed a weak effect size (Es = 0.10). The second variable was the peak knee extension moment, where the injured limb in both the ACL-RC and ACL-DF groups demonstrated lower peak knee extension moment than the control group ($p = 0.001$; Es = 0.20). Finally, significant differences were reported in peak GRF between the groups, as both legs in the ACL-RC and ACL-DF populations reported lower peak GRF than the control group ($p \leq 0.001$; Es = 0.20).

*Table 9-3: Knee kinematics, kinetics (peaks) and MV in each variable for the three groups during running.*

| Variable | ACL-RC | | ACL-DF | | Controls |
|---|---|---|---|---|---|
| | Injured leg Mean (SD) | Uninjured leg Mean (SD) | Injured leg Mean (SD) | Uninjured leg Mean (SD) | Dominant leg Mean (SD) |
| Knee flexion angle (°) | 40.19 (8.11) | 44.27 (6.07) | 43.17 (6.69) | 46.20 (7.24) | 45.9 5.74 |
| MV for knee flexion angle | 0.06 (0.05) | 0.05 (0.03) | 0.05 (0.02) | 0.05 (0.02) | 0.05 (0.03) |
| Knee extension moment (Nm/Kg) | 2.32 (0.83) | 2.76 (0.66) | 2.29 (0.69) | 2.86 (0.52) | 2.99 (0.53) |
| MV for knee extension moment | 0.15 (0.18) | 0.11 (0.06) | 0.11 (0.06) | 0.09 (0.05) | 0.09 (0.12) |
| Knee valgus angle (°) | -1.86 (4.88) | -1.87 (4.27) | -3.61 (4.14) | -2.18 (4.12) | -1.16 (5.17) |
| MV for knee valgus angle | 0.36 (0.32) | 0.30 (0.23) | 0.19 (0.15) | 0.32 (0.23) | 0.31 (0.27) |
| Knee valgus moment (Nm/Kg) | 0.23 (0.35) | 0.19 (0.23) | 0.19 (0.25) | 0.12 (0.16) | 0.20 (0.30) |
| MV for knee valgus moment | 0.54 (0.21) | 0.57 (0.28) | 0.54 (0.29) | 0.64 (0.25) | 0.54 (0.23) |
| V-GRF (*Body weight) | 1.91 (0.49) | 2 (0.37) | 2 (0.25) | 2.05 (0.23) | 2.31 (0.36) |
| MV for V-GRF | 0.06 (0.19) | 0.01 (0.003) | 0.03 (0.03) | 0.02 (0.03) | 0.04 (0.10) |

*Negative (-) represents knee valgus angle and knee varus moment.*

*Table 9-4: One-way ANOVA results with post hoc tests for running.*

| Variable | ANOVA Significance level (p value) (Between groups) | Post hoc pairwise comparison (significant differences) | P value (95% CI) | E.s ($\eta^2$) |
|---|---|---|---|---|
| Knee flexion angle (°) | 0.001* | Control vs RC injured<br>DF uninjured vs RC injured | 0.001 (0.82-10.66)<br>0.003 (0.21-11.79) | 0.10 |
| MV for knee flexion angle | 0.67 | NF | NA | 0.04 |
| Knee extension moment (Nm/Kg) | 0.001* | Control vs RC injured<br>Control vs DF injured | 0.001 (0.19-1.14)<br>0.001 (0.23-1.17) | 0.20 |
| MV for knee extension moment | 0.05 | NF | NA | 0.03 |
| Knee valgus angle (°) | 0.43 | NF | NA | 0.03 |
| MV for knee valgus angle | 0.29 | NF | NA | 0.04 |
| Knee valgus moment (Nm/Kg) | 0.31 | NF | NA | 0.02 |
| MV for knee valgus moment | 0.59 | NF | NA | 0.02 |
| V-GRF (*Body weight) | 0.001* | Control vs RC injured<br>Control vs RC uninjured<br>Control vs DF injured<br>Control vs DF uninjured | 0.000 (0.15-0.67)<br>0.001 (0.05-0.58)<br>0.000 (0.05-0.57)<br>0.001(0.002-0.53) | 0.20 |
| MV for V-GRF | 0.05 | NF | NA | 0.02 |

**Significant difference between groups was found following Bonferroni correction ($p \leq 0.005$).*

*NF: Not found. NA: Not applicable. † 0.000 indicates that the SPSS reading was 0.000 and which means that $p < 0.0005$.*

### 9.4.2 Cutting to 90 degrees

All participants in each group (n=85) performed a cutting to 90 degrees task on the force plate with both legs. Knee kinematics and kinetics results for this task are shown in Table 9-5. MV values for each variable are also presented in Table 9-5. The results of all post hoc tests (pairwise comparisons between groups) are displayed in Table 9-6. Overall, of all the variables, five reported significant differences between the groups. Firstly, a significant difference between the groups occurred in the peak knee extension moment, where both legs in the ACL-RC and ACL-DF groups reported lower peak knee extension moment than the control group ($p \leq 0.004$; Es = 0.32). Significant differences were also reported in the peak knee flexion angle between the groups, as the injured limb in the ACL-DF group demonstrated a lower peak knee flexion angle than the control group ($p = 0.004$; Es = 0.10) and the uninjured limb in the ACL-RC group ($p = 0.003$; Es = 0.10). However, these differences showed a weak effect size. The peak knee valgus moment was another variable which reported significant differences, where the injured limb in the ACL-DF group showed a lower peak knee valgus moment than both legs in the ACL-RC group ($p \leq 0.005$; Es = 0.10) which, notably, that was a weak effect size. Similar to the running task, significant differences were also reported in the peak GRF between the groups, as both legs in the ACL-RC and ACL-DF groups reported lower peak GRF than the control group ($p \leq 0.001$; Es = 0.25). MV for the peak GRF was the final variable in which significant differences occurred between groups, where the uninjured limb in both the ACL-RC and ACL-DF groups demonstrated lower MV values for peak GRF than the control group ($p \leq 0.002$; Es = 0.10), with weak effect size.

*Table 9-5: Knee kinematics, kinetics (peaks) and MV in each variable for the three groups during cutting to 90 degrees task.*

| Variable | ACL-RC | | ACL-DF | | Controls |
|---|---|---|---|---|---|
| | Injured leg Mean (SD) | Uninjured leg Mean (SD) | Injured leg Mean (SD) | Uninjured leg Mean (SD) | Dominant leg Mean (SD) |
| Knee flexion angle (°) | 61.05 (12.41) | 65.51(10.83) | 56.04 (9.47) | 61.85 (11.51) | 63.60 (8.06) |
| MV for knee flexion angle | 0.08 (0.09) | 0.07 (0.07) | 0.08 (0.05) | 0.05 (0.04) | 0.07 (0.06) |
| Knee extension moment (Nm/Kg) | 1.39 (0.62) | 1.64 (0.50) | 1.42 (0.61) | 1.82 (0.45) | 2.49 (0.89) |
| MV for knee extension moment | 0.21 (0.17) | 0.17 (0.17) | 0.17 (0.10) | 0.14 (0.08) | 0.13 (0.08) |
| Knee valgus angle (°) | -4.21 (5.1) | -5.96 (5.08) | -4.18 (4.73) | -4.58 (4.42) | -4.09 (5.37) |
| MV for knee valgus angle | 0.24 (0.14) | 0.32 (0.28) | 0.43 (0.28) | 0.35 (0.27) | 0.40 (0.29) |
| Knee valgus moment (Nm/Kg) | 1.25 (0.59) | 1.31 (0.74) | 0.73 (0.53) | 1.08 (0.74) | 1.19 (0.65) |
| MV for knee valgus moment | 0.27 (0.17) | 0.21 (0.09) | 0.34 (0.18) | 0.32 (0.20) | 0.28 (0.14) |
| V-GRF (*Body weight) | 1.89 (0.26) | 1.90 (0.23) | 1.76 (0.34) | 1.99 (0.27) | 2.23 (0.37) |
| MV for V-GRF | 0.06 (0.08) | 0.05 (0.07) | 0.07 (0.05) | 0.03 (0.04) | 0.087 (0.07) |

*Negative (-) represents knee valgus angle and knee varus moment.*

*Table 9-6: One-way ANOVA results with post hoc tests for cutting to 90 degrees task.*

| Variable | ANOVA Significance level (p value) (Between groups) | Post hoc Pairwise comparison (significant differences) | P value (95% CI) | E.s ($\eta^2$) |
|---|---|---|---|---|
| Knee flexion angle (°) | 0.004* | Control vs DF injured<br>RC uninjured vs DF injured | 0.004 (0.05-15.08)<br>0.003 (0.64-18.32) | 0.10 |
| MV for knee flexion angle | 0.33 | NF | NA | 0.02 |
| Knee extension moment (Nm/Kg) | 0.000*† | Control vs RC injured<br>Control vs RC uninjured<br>Control vs DF injured<br>Control vs DF uninjured | 0.000 (0.59-1.62)<br>0.000 (0.34-1.38)<br>0.000 (0.56-1.59)<br>0.004 (0.16-1.19) | 0.32 |
| MV for knee extension moment | 0.30 | NF | NA | 0.06 |
| Knee valgus angle (°) | 0.72 | NF | NA | 0.02 |
| MV for knee valgus angle | 0.16 | NF | NA | 0.05 |
| Knee valgus moment (Nm/Kg) | 0.004* | RC injured vs DF injured<br>RC uninjured vs DF injured | 0.005 (0.01-1.03)<br>0.005 (0.0002-1.16) | 0.10 |
| MV for knee valgus moment | 0.22 | NF | NA | 0.05 |
| V-GRF (*Body weight) | 0.000* | Control vs RC injured<br>Control vs RC uninjured<br>Control vs DF injured<br>Control vs DF uninjured | 0.000 (0.10-0.57)<br>0.000 (0.09-0.56)<br>0.000 (0.23-0.70)<br>0.003(0.01-0.48) | 0.25 |
| MV for V-GRF | 0.001 | Control vs DF uninjured<br>Control vs RC uninjured | 0.000 (0.005-0.10)<br>0.002 (0.01-0.08) | 0.10 |

**Significant difference between groups was found following Bonferroni correction ($p \leq 0.005$).*

*NF: Not found. NA: Not applicable. † 0.000 indicates that the SPSS reading was 0.000 and this means that $p < 0.0005$.*

### 9.4.3 SLL

All participants in the three groups (n=85) performed SLL on the force plate with both legs. Knee kinematics and kinetics are outlined in Table 9-7. MV values for each biomechanical variable are also presented in Table 9-7. The results of all post hoc tests (pairwise comparisons between groups) are displayed in Table 9-8. In general, six variables reported significant differences between the three groups. Firstly, significant differences occurred in the peak knee extension moment, where both legs in the ACL-RC group, as well as the injured limb in the ACL-DF group, demonstrated a lower peak knee extension moment than the control group ($p < 0.0005$; Es = 0.40). In the same variable, the injured limb in the ACL-RC group and the injured limb in ACL-DF group demonstrated a lower peak knee extension moment than the uninjured limb in the ACL-DF group and the uninjured limb in ACL-RC group respectively ($p < 0.0005$; Es = 0.40). Another significant difference was reported in the peak knee flexion angle, where the injured limb in the ACL-DF group demonstrated a lower peak knee flexion angle than the control group ($p = 0.002$; Es = 0.10). However, the latter represents a weak effect size.

In addition, significant differences in MV for the peak knee extension moment were reported between the groups, where the injured limb in the ACL-DF group showed greater MV for the peak knee extension moment ($p < 0.0005$; Es = 0.10). Another significant difference was found in the peak knee valgus, where the uninjured limb demonstrated greater peak knee valgus moment than the injured limb in the ACL-DF group ($p =0.003$; Es = 0.12). However, these differences showed a weak effect size (Es = 0.10 and 0.12 respectively). Finally, significant differences were also reported in the GRF between the groups, as both legs in the ACL-RC and ACL-DF groups reported a lower peak GRF than the control group ($p \leq 0.001$; Es = 0.28).

*Table 9-7: Knee kinematics, kinetics (peaks) and MV in each variable for the three groups during SLL task.*

| Variable | ACL-RC | | ACL-DF | | Controls |
|---|---|---|---|---|---|
| | Injured leg Mean (SD) | Uninjured leg Mean (SD) | Injured leg Mean (SD) | Uninjured leg Mean (SD) | Dominant leg Mean (SD) |
| Knee flexion angle (°) | 66.06 (12.53) | 66.33 (13.89) | 54.26 (11.04) | 60.16 (10.59) | 65.81 (13.02) |
| MV for knee flexion angle | 0.07 (0.05) | 0.06 (0.03) | 0.07 (0.02) | 0.06 (0.03) | 0.056 (0.02) |
| Knee extension moment (Nm/Kg) | 2.62 (0.63) | 2.96 (0.50) | 2.39 (0.57) | 3.22 (0.56) | 3.45 (0.46) |
| MV for knee extension moment | 0.09 (0.06) | 0.07 (0.03) | 0.09 (0.04) | 0.09 (0.06) | 0.06 (0.03) |
| Knee valgus angle (°) | -0.27 (3.92) | -1.65 (3.97) | -2.52 (3.72) | -2.28 (3.83) | -0.66 (4.39) |
| MV for knee valgus angle | 0.57 (0.34) | 0.39 (0.25) | 0.47 (0.31) | 0.37 (0.25) | 0.40 (0.26) |
| Knee valgus moment (Nm/Kg) | 0.29 (0.25) | 0.72 (0.57) | 0.34 (0.36) | 0.51 (0.42) | 0.45 (0.39) |
| MV for knee valgus moment | 0.59 (0.17) | 0.38 (0.21) | 0.50 (0.22) | 0.51 (0.20) | 0.50 (0.23) |
| V-GRF (*Body weight) | 2.05 (0.25) | 2.03 (0.24) | 2.12 (0.28) | 2.24 (0.59) | 2.71 (0.63) |
| MV for V-GRF | 0.01 (0.03) | 0.01 (0.02) | 0.02 (0.03) | 0.02 (0.04) | 0.027 (0.03) |

*Negative (-) represents knee valgus angle and knee varus moment.*

*Table 9-8: One-way ANOVA results with post hoc tests for the SLL task.*

| Variable | ANOVA Significance level (p value) (Between groups) | Post hoc Pairwise comparison (significant differences) | P value (95% CI) | E.s ($\eta^2$) |
|---|---|---|---|---|
| Knee flexion angle (°) | 0.001* | Control vs DF injured | 0.002 (0.96-19.40) | 0.10 |
| MV for knee flexion angle | 0.54 | NF | NA | 0.03 |
| Knee extension moment (Nm/Kg) | 0.000*† | Control vs RC injured<br>Control vs RC uninjured<br>Control vs DF injured<br>DF uninjured vs RC injured<br>RC uninjured vs DF injured | 0.000 (0.43-1.23)<br>0.000 (0.09-0.90)<br>0.000 (0.72-1.52)<br>0.000 (0.12-1.07)<br>0.000 (0.15-1.10) | 0.40 |
| MV for knee extension moment | 0.002* | DF injured vs Control | 0.000 (0.001-0.05) | 0.10 |
| Knee valgus angle (°) | 0.30 | NF | NA | 0.05 |
| MV for knee valgus angle | 0.29 | NF | NA | 0.06 |
| Knee valgus moment (Nm/Kg) | 0.001* | RC uninjured vs DF injured | 0.003 (0.03-0.29) | 0.12 |
| MV for knee valgus moment | 0.06 | NF | NA | 0.07 |
| V-GRF (*Body weight) | 0.000* | Control vs RC injured<br>Control vs RC uninjured<br>Control vs DF injured<br>Control vs DF uninjured | 0.000 (0.35-0.97)<br>0.000 (0.37-0.98)<br>0.000 (0.27-0.93)<br>0.001(0.004-0.93) | 0.28 |
| MV for V-GRF | 0.007 | NF | NA | 0.06 |

**Significant difference between groups was found following Bonferroni correction ($p \leq 0.005$).*

*NF: Not found. NA: Not applicable. † 0.000 indicates that the SPSS reading was 0.000 and this means that $p < 0.0005$.*

### 9.4.4 SLS

All participants in each group (n=85) performed a cutting to 90 degrees task on the force plate with both legs. Knee kinematics and kinetics results for this task are shown in Table 9-9. MV values for each variable are also presented in Table 9-9. The results of all post hoc tests (pairwise comparisons between groups) are displayed in Table 9-10. Overall, no significant difference between the groups was found during this task.

*Table 9-9: Knee kinematics, kinetics (peaks) and MV in each variable for the three groups during SLS.*

| Variable | ACL-RC | | ACL-DF | | Controls |
|---|---|---|---|---|---|
| | Injured leg Mean (SD) | Uninjured leg Mean (SD) | Injured leg Mean (SD) | Uninjured leg Mean (SD) | Dominant leg Mean (SD) |
| Knee flexion angle (°) | 82.42 (12.85) | 84.73 (8.65) | 78.25(11.76) | 83.22 (7.74) | 85.81 (11. 61) |
| MV for knee flexion angle | 0.04 (0.01) | 0.03 (0.02) | 0.06 (0.05) | 0.04 (0.03) | 0.05 (0.04) |
| Knee extension moment (Nm/Kg) | 1.15 (0.43) | 1.26 (0.46) | 1.12 (0.44) | 1.15 (0.53) | 1.22 (0.45) |
| MV for knee extension moment | 0.18 (0.16) | 0.21 (0.14) | 0.20 (0.14) | 0.22 (0.13) | 0.21 (0.12) |
| Knee valgus angle (°) | 0.67 (2.87) | -0.64 (3.08) | -0.35 (2.45) | 0.76 (3.08) | -0.78 (2.90) |
| MV for knee valgus angle | 0.42 (0.29) | 0.40 (0.30) | 0.57 (0.27) | 0.41 (0.31) | 0.43 (0.32) |
| Knee valgus moment (Nm/Kg) | 0.03 (0.04) | 0.08 (0.08) | 0.04 (0.03) | 0.05 (0.07) | 0.06 (0.06) |
| MV for knee valgus moment | 0.61 (0.23) | 0.50 (0.28) | 0.63 (0.23) | 0.62 (0.22) | 0.47 (0.27) |
| V-GRF (*Body weight) | 1.12 (0.07) | 1.14 (0.08) | 1.13 (0.08) | 1.15 (0.07) | 1.15 (0.12) |
| MV for V-GRF | 0.03 (0.02) | 0.03 (0.01) | 0.03 (0.02) | 0.04 (0.02) | 0.03 (0.02) |

*Negative (-) represents knee valgus angle and knee varus moment.*

*Table 9-10: One-way ANOVA results with post hoc tests for the SLS task.*

| Variable | ANOVA Significance level (p value) (Between groups) | Post hoc Pairwise comparison (significant differences) | P value (95% CI) | E.s ($\eta^2$) |
|---|---|---|---|---|
| Knee flexion angle (°) | 0.21 | NF | NA | 0.06 |
| MV for knee flexion angle | 0.05 | NF | NA | 0.08 |
| Knee extension moment (Nm/Kg) | 0.08 | NF | NA | 0.01 |
| MV for knee extension moment | 0.06 | NF | NA | 0.01 |
| Knee valgus angle (°) | 0.28 | NF | NA | 0.04 |
| MV for knee valgus angle | 0.24 | NF | NA | 0.04 |
| Knee valgus moment (Nm/Kg) | 0.06 | NF | NA | 0.08 |
| MV for knee valgus moment | 0.05 | NF | NA | 0.08 |
| V-GRF (*Body weight) | 0.06 | NF | NA | 0.02 |
| MV for V-GRF | 0.07 | NF | NA | 0.02 |

**Significant difference between groups was found following Bonferroni correction ($p \leq 0.005$).*

*NF: Not found. NA: Not applicable. † 0.000 indicates that the SPSS reading was 0.000 and this means that $p < 0.0005$.*

## 9.5 Discussion

The main aims of this study were to compare MV and knee biomechanics between ACL-RC, ACL-DF and healthy control groups during four common sporting tasks. In general, the results showed that there were significant differences between groups in at least one variable in each task. SLL showed the highest number of variables (7) reporting significant differences between the three groups, as compared to the other tasks. Beginning with the biomechanical variables, both the ACL-RC and ACL-DF groups showed a significantly lower peak knee extension moment than the control group during running, cutting to 90 degrees and SLL ($p \leq 0.001$, Es = 0.20; $p \leq 0.004$; Es = 0.32; $p < 0.0005$; Es = 0.40, respectively). These findings are in line with some studies which investigated knee extension moment in an ACL-RC population and included healthy controls during similar tasks: running (Pamukoff et al., 2018; Herrington et al., 2017; Noehren et al., 2014; Saxby et al., 2016); SLL (Pozzi et al., 2017; Gokeler et al., 2010; Ortiz et al., 2008); and cutting task (Saxby et al., 2016). All studies reported a significant decrease in extension moment in the ACL-RC knee, as compared to the healthy controls. Herrington et al. (2017) also compared the uninjured limb in an ACL-RC group to a control group. In contrast, results of a study by Pratt and Sigward (2017) did not concur with these findings, where they reported no significant difference in knee extension moment between the uninjured limb in the ACL-RC group as compared to the control group during running and SLL tasks. These differences could possibly be attributed to the fact that the latter authors included both genders in their results, where they did not present the data for each gender independently. This could be due to the study's small sample size (n=15; 7 males and 8 females).

Unfortunately, no studies to date have investigated knee extension moment in ACL-DF individuals during these functional tasks ; however, two studies did report similar results during walking and jogging activities (Kawahara et al., 2012; Lewek et al., 2002). Hence, the current study is the first of its kind to compare knee extension moment between legs, as well as to include a healthy control group during the four common functional sporting tasks. The injured limb in the ACL-RC group also demonstrated a significantly lower peak knee extension moment than the uninjured limb in the ACL-DF group during SLL. Similarly, the injured limb in the ACL-DF group reported significantly lower peak knee extension moment than the uninjured limb in the ACL-RC group during the latter task, with a moderate effect size (E.s = 0.40). In fact, these findings indicate that the injured knee experienced the same deficits prior to and following surgical interventions. This may be due to the ACL-RC group undertaking insufficient rehabilitation programmes.

The explanation of the reduction in extension moment in the injured limb in both ACL-RC and ACL-DF groups could potentially be attributable to quadriceps muscular weakness, in that these muscular contractions generally represent knee extension moment (Shimokochi et al., 2009). Furthermore, Ingersoll et al. (2008) has highlighted that prolonged quadriceps weakness or restriction frequently occurs among ACL-DF and ACL-RC cohorts. However, in the absence of appropriate treatment, Roos (2005) contended that early-onset degeneration at the knee joint can result. In seeking to address this problem, as well as to aid normalisation of moment at the knee joint, the restoration of quadriceps muscle strength should be the primary focus of attention, particularly when implementing rehabilitation programmes. A significant difference did not occur in the peak knee extension moment when undertaking the SLS task. Unlike the other functional tasks, the latter was carried out using the same foot position and, therefore, it was not necessary to engage in any step activities. Therefore, if this task was used in rehabilitation as an exit RTS strategy, then this could provide a false reading.

Importantly, it should be noted that both legs in the ACL-RC and ACL-DF groups demonstrated a significantly lower peak knee extension moment than in the control group during cutting to 90 degrees and SLL ($p \leq 0.004$; Es = 0.32; $p < 0.0005$; Es = 0.40, respectively). In fact cutting to 90 degrees and SLL tasks are considered to be the riskiest tasks in terms of ACL injury, as highlighted in Chapter 4. Such results indicate that alterations in knee extension moment occurred in both legs in the ACL-RC and ACL-DF groups as compared to the healthy cohort, which could be a main source of reinjury in any limb. These findings also support the contention that an uninjured limb does not automatically represent the healthy control group, as reported in previous studies (Chmielewski, 2011; Di Stasi, Myer, & Hewett, 2013; Wright, Magnussen, Dunn, & Spindler, 2011) and that when assessing an individual return to sport or discharge from rehabilitation, assessment against standardized healthy populations are needed. Furthermore, the results of the current study have shown that the injured limb in the ACL-DF group demonstrated a significantly lower peak knee flexion angle than the control group in both cutting and SLL tasks ($p = 0.004$; Es = 0.10; $p = 0.002$; Es = 0.10, respectively); however, its weak effect size could be due to the sample size, as our sample was not very large. This was also found to occur in the uninjured limb in the ACL-RC group during a cutting task. The injured limb in the ACL-RC group reported a significantly lower peak knee flexion angle than the control group during a running task, and it was also lower than the uninjured limb in the ACL-DF group during the same task. Similar findings were reported in several studies which investigated the knee flexion angle in an ACL-RC population, and which included healthy controls, during different tasks: running (Pamukoff et al., 2018; Herrington et al., 2017; Noehren et al., 2014; Saxby et al., 2016; Lewek et al., 2002); SLL (Pozzi et al., 2017; Webster & Feller, 2012b; Webster et al., 2012; Antolič et al., 1999); and a cutting task (Saxby et al., 2016). All aforementioned studies reported a significant decrease in the peak knee flexion angle in the injured knee in the ACL-RC group, as compared to the healthy controls. These results are also consistent with previous studies which investigated ACL-DF individuals and where healthy controls were included during: jogging (Lewek et al., 2002; Rudolph et al., 2001; Chmielewski et al., 2001); and walking tasks (Kawahara et al., 2012; Rudolph et al., 2001; Chmielewski et al., 2001). However, no studies compared ACL-DF individuals to a healthy control group during the four common sporting tasks. Hence, the current study is the first to compare the knee flexion angle in an ACL-DF cohort to healthy controls during functional sporting tasks.

A decrease in knee flexion functioning when conducting common sporting tasks could be as a result of quadriceps impairment, which manifests as a significant reduction in the knee extension moment. Leppänen et al. (2017a) identified that a reduction in the knee flexion angle was a risk factor for ACL-related injury. Furthermore, according to Li et al. (2005) and Markolf et al. (1995), it can result in a greater likelihood of anterior tibial shear occurring. Consequently, if a reconstructed knee remains in an extended position, the knee joint is of heightened risk of sustaining another ACL injury on the ipsilateral side, particularly when engaging in a cutting task. Thus, this highlights the benefits of enhancing knee flexion as part of a rehabilitation programme. In contrast, SLS did not show significant differences in the peak knee flexion angle between the groups. Previous research also reported no significant differences between ACL-RC individuals and a healthy control during parallel squats (Neitzel et al., 2002). In addition, Yamazaki et al. (2010) reported no significant difference in the peak knee flexion angle between an ACL-DF cohort and a control group during SLS . One reason why the SLS did not report significant differences could be because this task is less complex and less stressful than the other tasks.

In addition, both legs in the ACL-RC and ACL-DF groups demonstrated a lower peak V-GRF than the control group ($p \leq 0.003$) in all tasks except SLS. Similar results were reported in previous studies (Miranda et al., 2013; Mohammadi et al., 2013; Vairo et al., 2008) where a significant difference between the injured limb in the ACL-RC group and the control group occurred during a SLL task. However, these findings do not concur with those revealed in a study by Milandri et al. (2017) who reported no significant difference in peak GRF between the ACL-RC and control groups during a running task. The difference between our findings and theirs could arise due to the fact that they included ACL-RC individuals who were a number of years post-surgery (2–6 years), while the current study was solely restricted to those who were 6–12 months post-surgery. It is highly likely that the length of the time frame following surgery can affect knee biomechanics (Webster & Hewett, 2019). The findings of another study by Pozzi et al. (2017) do not concur with the present study regarding GRF during a SLL task. The source of the difference between the latter study and all aforementioned research and the current study in terms of the reported significant difference between the injured limb in the ACL-RC and control groups could be due to the inclusion of both genders (17 female, 3 male) in each group in the other studies, while the current research included males only. Another possible explanation is that they recruited ACL-RC individuals who were more than 12 months post-surgery, while the current study only included those who were 6–12 months post-surgery.

In terms of GRF in an ACL-DF group as compared to a control group, Rudolph et al. (2001) and Chmielewski et al. (2001) also reported similar results to the current study during walking. Unfortunately, no studies investigated the GRF in ACL-DF individuals as compared to a healthy cohort during functional sporting tasks. In addition, no studies compared GRF for the uninjured limb to a control group. Hence, the current study is the first to bridge this gap. Such comparisons are very important from a reinjury perspective, as there is high rate of reoccurrence, as evidenced in several studies (Webster & Hewett, 2019; Dekker et al., 2017; Morgan et al., 2016; Paterno, et al., 2014). Interestingly, peak GRF is another important variable in the uninjured limb in that both ACL-RC and ACL-DF groups failed to demonstrate normal values similar to healthy controls. In fact, these findings also support the premise that the uninjured limb does not represent a healthy control group, as highlighted in previous studies (Chmielewski, 2011; Stasi, Myer, & Hewett, 2013; Wright, Magnussen, Dunn, & Spindler, 2011). Importantly, these findings strongly indicate that the uninjured limb must be targeted in ACL injury rehabilitation programmes following either surgical or conservative treatment.

With regard to knee valgus, the results of this study have shown that there was a significant difference with low effect size in knee valgus moments between the groups in two tasks, namely SLL and cutting to 90 degrees. In both tasks, the uninjured limb in the ACL-RC group demonstrated greater valgus moment than the injured limb in the ACL-DF group ($p = 0.003$; Es = 0.12; $p = 0.005$; Es = 0.10, respectively). In addition, the injured limb in the ACL-RC group demonstrated significantly greater knee valgus moment than the injured limb in the ACL-DF group ($p = 0.003$; Es = 0.12). Notwithstanding that the knee valgus moment was greater in the uninjured limb in the ACL-RC group than the control and ACL-DF groups during SLL and cutting tasks, the earlier findings did not reach a significance level. The results also showed that no significant difference occurred between the ACL-DF and control groups in the knee valgus variables in all tasks.

In fact, these results imply that the uninjured limb in ACL-RC individuals is of higher risk of sustaining a second ACL injury than the injured one, which may explain why the likelihood of ACL reinjury occurring on the contralateral side is high, as the knee valgus was identified as a risk factor for ACL injury (Hewett et al., 2005; Myer et al., 2015; Numata et al., 2018; McLean, Lipfert, & Van Den Bogert, 2004). Clinicians should consider this and target the contralateral limb during ACL rehabilitation and RTS programmes. Furthermore, such findings where the ACL-RC group demonstrated greater knee valgus moment than the ACL-DF and control groups indicate that this deficit has developed post-surgery.

Few studies have investigated the knee valgus in ACL-RC individuals, as compared to a healthy control group. Ortiz et al. (2008) reported similar results to the current study during SLL, where they found that the reconstructed limb demonstrated greater knee valgus moment than the healthy control group; however, it did not reach significance level. Our findings regarding the knee valgus angle remain consistent with earlier studies by Webster et al. (2012) and Kuenze et al. (2014) who reported no significant difference between the reconstructed leg and the control group during SLL and jogging, respectively. However, Stearns and Pollard (2013) reported conflicting results as compared to our findings during a cutting task. This may be attributed to the fact that they included females only in their study and their sample size was 11 and could be small, without providing a justification for same. Unfortunately, no studies have investigated knee valgus in a male ACL-RC cohort as compared to healthy controls during cutting and SLS tasks. In addition, no studies have examined knee valgus in an ACL-DF cohort as compared to a control group or in an ACL-RC population as compared to an ACL-DF group during all four functional sporting tasks. Hence, the current study is the first to conduct such investigations.

Several points need to be taken into consideration regarding the aforementioned studies which investigated either ACL-RC or ACL-DF populations as compared to healthy cohorts during any of the four sporting tasks. Firstly, few authors have investigated both knee kinematics and kinetics values associated with the risk of ACL injury, with the exception of Webster et al. (2012), Ortiz et al. (2008) (SLL) and Milandri et al. (2017), (running). Furthermore, only three studies, namely Pratt et al. (2017) (running and SLL), Webster et al. (2012) (SLL) and Saxby et al. (2016) (running and cutting) performed power calculations to determine the sample size, while the remaining authors failed to do so. Adequate sample sizes are necessary as small samples bias the results and inflate type I errors. In addition, they are needed for multiple statistical tests of dependent variables (Knudson, 2017). Unfortunately, some researchers included small number of participants and could be  considered as sample sizes, without providing sufficient justification for this decision. Gokeler et al. (2010) studied SLL among ACL-RC individuals, where they included only nine participants in total. Another study by Sanford et al. (2016) focusing on SLS comprised only a very small number of participants (n= 8).

Furthermore, gender is another significant influencing factor. While most previously cited research included both male and female samples, five studies focused specifically on one gender only. Two drew on a female cohort to perform running (Noehren et al., 2014)  and SLL (Ortiz et al., 2008) tasks, while the three remaining studies comprised a male-only sample carrying out SLL (Webster et al., 2012) and running (Milandri et al., 2017) activities. As highlighted in Chapter 4, there is evidence to support significant gender variation in biomechanical functioning. Consequently, data should be differentiated by gender, and failure to do so has been identified as a limitation of all research undertaken in this area. In addition, great disparity has also occurred in female-male participation ratios in the studies conducted to date: 2:18 (Webster et al., 2004); 32:3 (Webster et al., 2012a); 10:24 (Herrington et al., 2017); 38:66 (Saxby et al., 2016) and 3:5 (Sanford et al., 2016). This significant difference could bias the results toward one gender.

One further factor worthy of note is the proximity of the post-surgery period to the testing phase. In most cases, testing took place within 6 to 12 months, while six studies examining an ACL-RC cohort were conducted in excess of two years post-surgery (Milandri et al., 2017; Pamukoff et al., 2018; Sanford et al., 2016; Ortiz et al., 2008; Saxby et al., 2016; Stearns & Pollard, 2013). Conversely, one study examined an ACL cohort within 4.6 ± 1.4 months of surgery, which could indicate that some participants may still have been attending rehabilitation programmes during the testing period. As previously highlighted, biomechanical functioning is impacted by the length of the post-surgery period, as confirmed by Neitzel et al. (2002) where they noted that the loading responses in uninjured and reconstructed limbs in an ACL-RC cohort can be affected by the time span post-ACL reconstruction.

With regard to MV, the findings of the current study have shown that there were significant differences in MV between the three groups. However, this was not found to occur in all MV variables. Firstly, the uninjured limb in both the ACL-RC and ACL-DF groups showed significantly lower MV for GRF than the control group. In contrast, the injured limb in the ACL-DF group showed significantly greater MV for knee extension moment than the control group ($p < 0.0005$). In the literature, MV has been linked to injury (Brown et al., 2009; Konradsen, 2002; Konradsen & Voigt, 2002). It has also been proposed that MV could be associated with risk of developing musculoskeletal injuries, including ACL injuries (McLean et al., 1999).

In addition, based on our investigations (Chapter 5), MV has been found to be significantly correlated with ACL biomechanical risk factors. The investigations outlined in Chapter 4 demonstrated that if MV increases then ACL injury-related risk factors decrease. Such findings when the ACL-RC population demonstrated lower MV in two variables that considered as risk factors of ACL injury during risky tasks (cutting to 90 degrees and SLL), it indicates that the biomechanical risk factors may increase and lead to increased risk of second ACL injury occurrence in either leg. Importantly, despite the findings set out in Chapters 6 and 7 which showed that the injured limb in both ACL-RC and ACL-DF groups demonstrated greater variability in some variables than the uninjured limb, both limbs failed to show greater MV than the healthy cohort, with one exception. Only MV for the knee extension moment during SLL demonstrated that the injured limb in the ACL-DF group showed greater MV than the control group ($p < 0.0005$). This deficit in MV, in particular in the ACL-RC group could be one possible reason of reinjury. Clinicians and researchers should take this into account and should target deficits in MV occurring in both legs in ACL-injured populations so as to avoid reinjury. Therefore, improving MV is very important and this can be achieved by specifically targeting neuromuscular control exercises.

In the literature, few studies have investigated MV in ACL-RC individuals while they are performing different tasks. Unfortunately, no studies to date have investigated MV in an ACL-RC group as compared to a healthy cohort using similar methods to the current studies. However, a few studies have investigated coordination variability based on the dynamic systems theory (see Section 2.11.4.3) in ACL-RC individuals as compared to a healthy control group. Gribbin et al. (2016) investigated a walking task, whereby they reported that most joint couplings in the ACL-RC group showed greater coordination variability than the healthy cohort. In addition, Pollard et al. (2015a) investigated coordination variability using the dynamic systems theory and reported significant differences in some variables where the ACL-RC cohort showed increased coordination variability than the control group. However, it is not possible to compare our findings to these studies as they adopted different methods and approaches. In addition, to date no study has investigated MV in ACL-RC individuals during SLL and SLS tasks. The reason why ACL-RC and ACL-DF demonstrated less MV than healthy controls could be due to the provision of insufficient rehabilitation programmes, resulting in inadequate neuromuscular control, and which, ultimately, leads to poorly controlled movement (Schmidt, 2003). In addition, this could be due to the new biomechanical constraints that have been caused by the injury, which, in turn, affect biomechanical flexibility. In their systematic review of ACL injury reduction training programmes, Webster and Hewett (2019) highlighted that just less than one quarter (23%) of all ACL-RC patients successfully completed RTS test batteries. Furthermore, despite a 43% failure rate being recorded, many still resumed sporting activities even though many may continue to display poorly controlled movement. Van Melick et al. (2016) also noted that athletes who succeed in one specific criterion on a RTS battery test, may subsequently not pass it at another point in time.

One other less plausible explanation may be that ACL reconstruction does not succeed in improving biomechanics in injured individuals, in that a number of deficits in knee biomechanics presenting prior to surgery may persist post-surgery, which can include deficits in MV and a reduction in knee extension moment. While surgery undeniably offers greater functional stability, Smith et al. (2014) have acknowledged that uncertainty remains in terms of identifying the optimal interventions to adopt.

Clearly, further improvements are required in relation to the rehabilitation programmes being provided, along with the RTS criteria selected for people following ACL-RC. Addressing these biomechanical deficits will reduce the likelihood of reinjury, as well as assisting this cohort to re-engage in sporting activities in a safe manner. Care should be taken to ensure that neuromuscular control exercises are integrated into programmes in order to help prevent the emergence of biomechanical deficits or reinjury risk. An alternative explanation mooted by Hodges and Tucker (2011) proposes that new MV acts as a compensatory movement, in an effort to reduce the load exerted on painful tissues. Therefore, Hamill, Palmer and Van Emmerik (2012) and Stergiou, Harbourne and Cavanaugh (2006) have warned that persistent and untreated MV-related deficits could lead to further damage or injury. Focusing specifically on neuromuscular control would appear to be the optimum approach to adopt.

### 9.6 Clinical implications

A number of different clinical implications have been identified in this study. Firstly, clinicians and researchers should be aware of significant differences in a number of knee biomechanical functions between ACL-RC, ACL-DF and healthy populations. These differences can help them to identify and target biomechanical and MV deficits in ACL-injured populations. Importantly, they should also know that knee extension moment is significantly lower in both ACL-RC and ACL-DF populations as compared to healthy controls, which indicates that a weakness can occur in the quadriceps muscle and this should be targeted when implementing a rehabilitation programme. In addition, they should be aware that the uninjured knee in ACL-RC population showed the greatest knee valgus moment during the SLL task as compared to the other groups. Unfortunately, as increased knee valgus moment is a risk factor in sustaining ACL injury, clinicians must take this into consideration when developing ACL reinjury risk mitigation and ACL-RC rehabilitation RTS programmes. In addition, both ACL-RC and ACL-DF groups showed a lower peak knee flexion angle than the healthy group during running, SLL and cutting tasks. Improving knee flexion during functional tasks should be targeted so as to avoid further damage or injury, which may occur due to altered knee flexion. Clinicians should be aware that both sides in ACL-injured populations demonstrate lower MV than the healthy ones, which could be attributable to a reduction in neuromuscular control. Hence, practitioners should include exercises which can improve this deficit. Finally, a biomechanical assessment should comprise one of the RTS criteria in order to ensure that biomechanical and MV deficits are eliminated at this point, especially in those demonstrating knee biomechanics impairments, namely knee extension moment, knee flexion angle, knee valgus angle/moment and GRF.

## 9.7 Study novelty

As highlighted throughout this chapter, this novel study is the first of its kind to compare knee biomechanics between ACL-RC and ACL-DF populations, as well as to compare knee biomechanics for the uninjured leg in both ACL-RC and ACL-DF populations to a healthy control group in the same study during functional sporting tasks. In addition, it compared MV between ACL-RC, ACL-DF and healthy control groups during functional sporting tasks.

## 9.8 Limitations of this study and future work

The current study has some limitations. Firstly, the investigations in this study were confined to males only. Future studies would need to carry out such investigations in a female cohort in order to establish whether they would demonstrate similar results or display more or less impairments in MV and knee biomechanics. New research is also warranted to support the development of a rehabilitation programme for ACL-RC patients, which would help to reduce all these biomechanical impairments and MV deficits, thus facilitating their safe return to sport.

## 9.9 Conclusion

Significant differences emerged in both knee biomechanics and MV between ACL-RC, ACL-DF and healthy groups in all four sporting functional tasks. However, these findings were not reported in all variables. Both ACL-RC and ACL-DF groups demonstrated a significantly lower peak knee extension moment as compared to healthy controls. They also showed a lower peak knee flexion angle than the control group during running, SLL and cutting tasks. In addition, both ACL-RC and ACL-DF groups demonstrated lower MV than the healthy controls. The uninjured limb in ACL-RC group showed greatest knee valgus moment between groups during SLL and cutting tasks. These findings may be of assistance in the development of rehabilitation and RTS programmes for ACL-injured patients who either undertook surgical or conservative treatment. In addition, they will be of assistance when developing ACL injury/reinjury risk mitigation programmes.

# Chapter 10

## 10 Overall Summary, Conclusions, Impact of the thesis and Recommendations for Future Research

### 10.1Summary

Anterior cruciate ligament (ACL) has a crucial function in relation to mechanical stability and proprioception (Abram et al., 2019). However, ACL injury can frequently occur, where Moses et al. (2012) recorded an annual median incidence per person of approximately 0.03%, with this figure rising to 3.7% in the case of professional athletes. A longitudinal UK-based study spanning from 1997 to 2017 revealed that the age-standardised and sex-standardised ACL reconstruction rate rose twelvefold in that time period, while there was 2.4 times more meniscal repair carried out (Abram et al., 2019). Moreover, there are substantial costs associated with this injury, most notably interventions such as imaging, reconstructive surgery, postoperative bracing and rehabilitation. These have been calculated to amount to in excess of two billion dollars annually (Gottlob et al., 1999; Malek, DeLuca, Kunkle, & Knable 1996). Other more intangible and individual costs also need to be factored in, such as the psychological impact of injury, in particular, when elite athletes sustain injury. In most situations when athletes suffer injury, they will be forced to sit out a competition for 6–9 months. This can have very significant repercussions for them, depending on the degree to which they are engaging in competitive play for monetary gain.

In order to minimise the impact of this very serious problem, it is important to identify the various conditions from different perspectives. Biomechanics are one of the most important aspects to understand in ACL injury, as several biomechanical risk factors have been identified as causing these types of injuries. Movement variability (MV) is another very important aspect, as it has been suggested that it could increase the likelihood of injury occurring (Baida, Gore, Franklyn-Miller, & Moran, 2018; James, 2004; Konradsen, 2002; Konradsen & Voigt, 2002). In addition, it is proposed that it correlates with risk of further musculoskeletal injuries, such as ACL rupture (McLean et al., 1999). In fact, intra-individual variability between repeated movement patterns, such as a SLL task or stride cycle, is deemed to comprise a core component of all motor tasks (König et al., 2016) and, therefore, it cannot be avoided. However, it is important to study the levels of MV in healthy and ACL-injured populations in order to fully understand the differences between normal and injured individuals. Another important consideration is to acquire an understanding of the relationship between MV and biomechanical risk factors, as the former may play a role in the injury.

In the literature, the most commonly investigated tasks in terms of sport injuries are landing, running, SLS and cutting. All these tasks were investigated in this thesis as most non-contact sport injuries occur during these activities, particularly cutting and landing. The inclusion of both risky and non-risky tasks is important in order to acquire an understanding of the differences between them, thus minimising injury. Furthermore, it is crucial in terms of shaping the development of rehabilitation and return to sport (RTS) programmes for ACL-injured patients who either undertook surgical or conservative treatment. In addition, it is important to undertake a comparative analysis of all tasks so as to rank them in order according to level of risk. Such investigations can assist in the development of ACL injury risk mitigation and ACL reconstruction rehabilitation return to sport (RTS) programmes.

The involvement of all ACL injury-related populations in the investigation can offer a more comprehensive understanding of the condition. Hence, three populations have been aligned with this injury and all have been included in this thesis. Firstly, a healthy cohort was included in order to determine knee biomechanics and MV values in a normal population. Secondly, an

ACL-DF population was selected so as to understand the effects of the injury on biomechanics and MV. Thirdly, an ACL-RC cohort was included so as to understand how surgery can alter the deficits in both knee biomechanics and MV caused by this injury. In addition, it is important to carry out different levels of comparison between the limbs and these groups. Leg dominance in a healthy population was investigated to establish its effect. Another comparative level examined was the injured side in both ACL-DF and ACL-RC groups, which can assist in determining the effect of injury or surgery on both affected and non-affected limbs.

Between-group comparisons are also very important as this level of comparison can allow us to identify differences in knee biomechanics and MV between all populations in relation to this condition. Notwithstanding that a number of studies have investigated an ACL-injured population and compared them to healthy controls, there is still a lack of information regarding the uninjured limb as compared to a control group in terms of both biomechanics and MV during different functional sporting tasks. Such investigations are very important in seeking to gain an understanding of reinjury, especially on the contralateral side. In addition, comparing ACL-RC to ACL-DF populations is important when seeking to ascertain how surgery can affect biomechanics and MV. Such comparisons can also improve our knowledge of the efficacy of current rehabilitation programmes following ACL reconstruction and during the RTS stage.

Due to the importance of all the aforementioned points highlighted above, this thesis was conducted in order to achieve the specific aims listed below:

1) Examine the reliability of 3D motion analysis as a measure of knee biomechanics and CV as a measure of MV during four common sporting tasks.

2) Investigate MV and knee biomechanics in healthy population to establish reference values for MV and knee biomechanics and to establish the relationship between MV and knee biomechanical risk factors of ACL injury during these four tasks.
3) Investigate MV and knee biomechanics in ACL-DF and ACL-RC populations during the four tasks.
4) Compare MV and knee biomechanics between the three groups during the four sporting tasks.

**10.2 Conclusion**

A review of the literature was conducted on ACL injury, knee biomechanics and MV, based on the identified need to address a multi-level research gap. The findings to emerge were subsequently used to set the aims and objectives of this thesis. These were outlined in Chapter 3. Following on from this, the next step was to test the methods set out in Chapter 4. They were selected for all the investigations that would be conducted in the study, and involved examining the reliability of 3D motion analysis, as well as the use of CV as a measure of MV during the four selected sporting tasks. The results showed that all 3D lower limb kinematic and kinetic variables emerging as part of the investigation process were reliable in terms of the running, SLS, SLL and cutting to 90 degrees tasks. However, the levels of reliability were found to differ, ranging from fair to excellent. In addition, all variables reported low SEM values, which is indicative that the change was accepted as accurate. In addition, the results demonstrated that CV (second order) is a reliable measure of variability in lower limb kinematic and kinetic variables.

The next step was to investigate MV and knee biomechanics in a healthy population in order to achieve the second aim, which was to establish reference values for MV and knee biomechanics, as outlined in Chapter 5. The latter chapter also set out two sub-aims relating to this population, as explained earlier in the thesis. These two sub-aims were: firstly, to investigate the effect of leg dominance on MV and knee biomechanics in healthy individuals during the four sporting tasks; and, secondly, to investigate the effect of task on MV and knee biomechanics in healthy individuals during these tasks. The reference values for all investigated variables were provided for both MV and knee biomechanics in this chapter. The results showed that no significant differences in either MV or in knee biomechanics between dominant and non-dominant legs occurred during all four tasks. However, significant differences were noted in the between-task pairwise comparisons for most of the biomechanical variables and for a number of the MV variables. In addition, of all four tasks, cutting to 90 degrees demonstrated the highest values in terms of knee biomechanical risk factors, following by SLL. However, the former task demonstrated lower MV, as compared to the other tasks.

The second element of Aim 2 was set out in Chapter 6, whereby the relationship between MV and ACL risk factors was established. The findings outlined in this chapter showed that a significant negative relationship was found to occur between MV and knee biomechanical risk factors for ACL injury during these sporting activities. That implies that an inverse relationship exists between MV and biomechanical risk factors, whereby if MV values increase then the biomechanical risk factors decrease. However, this thesis suggests that the MV should be in the normal average range, in accordance with the reference values outlined in Chapter 5.

The third aim of this thesis is to involve all relevant ACL injury populations. Chapters 7 and 8 focused on this aspect, by investigating MV and knee biomechanics in both ACL-DF and ACL-RC populations, respectively. In Chapter 7, findings of a comparison undertaken between an ACL-DF knee and the contralateral side were outlined. The results demonstrated that there were significant differences between the ACL-DF limb and the uninjured one in both knee biomechanics and MV values. However, this was not found to occur in all variables. In addition, the ACL-DF limb demonstrated significantly lower peak knee extension moment as compared to the uninjured side. It also showed both a lower peak knee flexion angle and peak GRF. However, the ACL-DF limb demonstrated greater MV than the uninjured one.

The final population impacted by ACL injury is an ACL-RC cohort, which was also included in investigations undertaken in this thesis. This was the focus of Chapter 8, where this particular cohort was examined in relation to the second element of Aim 3. A comparison was also carried out between the reconstructed knee and the contralateral side. The findings set out in this chapter revealed that significant differences emerged between the ACL-RC and the uninjured limb for both knee biomechanics and MV. However, these outcomes were not reported in all variables. The ACL-RC limb demonstrated a significantly lower peak knee extension moment as compared to the uninjured side. It also showed a lower peak knee flexion angle during both running and cutting tasks. However, the ACL-RC limb demonstrated greater MV than the uninjured one.

Finally, aim 4 was the focus of Chapter 9, whereby comparisons were undertaken of MV and knee biomechanics between the three populations. In this chapter, the uninjured limb in both ACL-RC and ACL-DF populations were included in the comparisons. The results of this chapter showed that significant differences emerged in both knee biomechanics and MV between ACL-RC, ACL-DF and healthy groups in all four sporting functional tasks. However, these findings were not reported in all variables. Both ACL-RC and ACL-DF groups demonstrated a significantly lower peak knee extension moment as compared to healthy controls. They also showed a lower peak knee flexion angle than the control group during running, SLL and cutting tasks. In addition, both ACL-RC and ACL-DF groups demonstrated lower MV than the healthy controls. Of all the groups, the uninjured limb in the ACL-RC cohort showed greatest knee valgus moment during SLL and cutting tasks. These findings may be of assistance in the development of rehabilitation and RTS programmes for ACL-injured patients, in the case of those who either underwent surgical or conservative treatment. Furthermore, they will be of benefit when developing ACL injury/reinjury risk mitigation programmes.

Finally, the findings of this thesis may pose concerns for clinicians and researchers regarding ACL injury in all relevant populations. The recommended clinical implications are highlighted in every chapter. These findings may make a significant contribution to the development of rehabilitation and RTS programmes for ACL-RC/ACL-DF patients. They may also be of assistance when developing ACL injury risk mitigation programmes.

### 10.3 Impact of the thesis

The findings of this thesis have several important clinical implications in terms of assisting all three populations in relation to non-contact ACL injury. Beginning with the healthy population, this thesis has established the reliability of knee biomechanics using 3D motion analysis during all four common sporting tasks, as well as establishing the reliability of CV as a measure of MV. In addition, the thesis has provided reference values for MV, as well as for knee kinematics and kinetics, associated with the risk of ACL injury for both legs (dominant/non-dominant) among this population group. These values are very important as they will allow clinicians and researchers to carry out biomechanical assessments, having gained a solid understanding of how these variables typically present in healthy individuals. It will also assist them to differentiate between normal and abnormal characteristics, in particular, in relation to the biomechanical variables that are considered to be risk factors for knee injury.

In addition, the thesis has revealed that no significant differences in either the knee biomechanics or in MV between dominant and non-dominant legs for a healthy male population, during all four common sporting tasks. Consequently, clinicians could deal with both legs in the same way, when assigning exercises. Comparisons drawn between tasks have shown that cutting to 90 degrees differs significantly from all three other tasks, as it places the knee joint in a greater valgus pattern. Therefore, it is advisable not to begin with this sporting task when engaging in either prevention or rehabilitation programmes for ACL injury. Instead, it is recommended to defer it to a more advanced stage, in order to avoid any complications that may occur. Furthermore, of the four tasks, SLL demonstrated the greatest GRF. Having a greater peak GRF means that there is an increased load placed on the knee joint. Clinicians should take this into consideration when developing either prevention or rehabilitation programmes. Another very important implication of this thesis is that it has established the relationship between MV and knee biomechanical risk factors for ACL injury, revealing that a negative relationship exists between both factors, whereby if MV increases then the biomechanical risk factors decrease. This reinforces the point that increased MV could be a primary goal when developing ACL injury risk mitigation programmes and returning to sport following ACL reconstruction and rehabilitation. However, this study proposes that MV should remain within the average reference values recommended for healthy individuals in the previous chapter.

This thesis has also provided useful information in terms of the ACL-DF population, in that it has revealed significant differences in some knee biomechanics values between the ACL-DF side and the uninjured one. In particular, clinicians and researchers should be aware that knee extension moment is significantly lower in the injured limb, which means that a weakness occurs in the quadriceps muscle and this should be targeted when implementing a rehabilitation programme. In addition, this thesis has revealed that no significant difference occurred in knee extension moment between the ACL-DF and the uninjured side during the SLS task, which implies that this activity could be the best option to select to reduce imbalance in the extension moment between the two sides. In addition, the ACL-DF side showed a lower peak knee flexion angle than the uninjured one. Improving knee flexion during functional tasks should be targeted to avoid further damage or injury, which may occur due to altered knee flexion, as decreased knee injury is an indication of risk of further damage. In terms of MV, clinicians should be aware that the ACL-DF side demonstrated greater MV than the uninjured one, which could be attributable to a reduction in neuromuscular control. Hence, practitioners should include exercises which can improve this function, such as strengthening and movement control exercises.

This thesis has also provided important information for an ACL-RC population, as it has revealed significant differences in some knee biomechanical functions between the ACL-RC side and the uninjured one. Clinicians and researchers should be aware that the knee extension moment is significantly lower in the reconstructed limb, which indicates that a weakness can occur in the quadriceps muscle and this should be targeted when implementing a rehabilitation programme. In addition, this thesis has revealed that the ACL-RC side showed a lower peak knee flexion angle than the uninjured one during running and cutting tasks. Improving knee flexion during functional tasks should be targeted so as to avoid further damage or injury, which may occur due to altered knee flexion. Clinicians should be aware that the ACL-RC side demonstrated greater MV than the uninjured one, which could be attributable to a reduction in neuromuscular control. Hence, practitioners should include exercises which can improve this deficit.

The final clinical implication derived from this thesis emanates from the comparisons drawn between all three ACL injury-related groups. Firstly, the thesis has revealed that significant differences occurred in a number of knee biomechanical functions between ACL-RC, ACL-DF and healthy populations. These differences can assist clinicians and researchers to identify and target biomechanical and MV deficits in ACL-injured populations. Importantly, it has revealed that knee extension moment is significantly lower in both ACL-RC and ACL-DF populations as compared to healthy controls, which indicates that a weakness can occur in the quadriceps muscle, and this should be targeted when implementing a rehabilitation programme. In addition, they should be aware that the uninjured knee showed the greatest knee valgus moment during the SLL task as compared to the other groups. Unfortunately, as increased knee valgus moment is a risk factor in sustaining ACL injury, clinicians must take this into consideration when developing ACL reinjury risk mitigation and ACL-RC rehabilitation RTS programmes.

In addition, both ACL-RC and ACL-DF groups showed a lower peak knee flexion angle than the healthy group during running, SLL and cutting tasks. Improving knee flexion during functional tasks should be targeted so as to avoid further damage or injury, which may occur due to altered knee flexion. Clinicians should be aware that both sides in ACL-injured populations demonstrate lower MV than the healthy ones, which could be attributable to a reduction in neuromuscular control. Hence, practitioners should include exercises which can improve this deficit. Finally, a biomechanical assessment should comprise one of the RTS criteria in order to ensure that biomechanical and MV deficits are eliminated at this point, especially in those demonstrating knee biomechanics impairments, namely knee extension moment, knee flexion angle, knee valgus angle/moment and GRF.

### 10.4 Recommendations for future research

While this thesis conducted a wide-ranging investigation of MV and knee biomechanics on all populations related to ACL injury, nevertheless, new questions have arisen which are worthy of further consideration. Firstly, based on the findings and discussion presented in Chapter 5, it is recommended that similar investigations be conducted on a female cohort, as significant gender differences in knee biomechanics during these four specific sporting tasks have been reported in several studies: cutting task (Weinhandl, Irmischer, Sievert, & Fontenot, 2017; James et al., 2004; Ford, Myer, Toms, & Hewett, 2005); landing task (Hughes, 2019; Weinhandl, Irmischer, Sievert, & Fontenot, 2017; Holden, Boreham, & Delahunt, 2016; Jenkins et al., 2017); and SLS (Dwyer et al., 2010; Graci et al., 2012; Weeks et al., 2015; Yamazaki et al., 2010; Zeller et al., 2003).

In addition, based on the findings and discussion set out in Chapter 6, new research is also needed to establish whether clinical assessments would help clinicians to assess MV within clinical practice, as it is not possible to conduct 3D motion analysis in every clinical assessment, due to both the cost and time constraints associated with the application of this method. Further research is recommended to investigate MV prospectively in order to establish whether it is a risk factor for ACL injury. Further recommendations for future research identified in Chapter 8 would be to conduct the same investigations on female ACL-RC individuals in order to determine whether they demonstrate the same deficits in MV and knee biomechanics. This would allow clinicians and researchers to fully discern the deficits that exist in both genders, especially given that significant gender differences have been found to occur. One further recommendation for future research, also identified in Chapter 8, is to establish whether MV changes occurred following a period of rehabilitation.

Finally, according to the findings and investigations described in Chapter 9 of this thesis, it is recommended that future studies be conducted in order to carry out similar comparisons between the three populations relating to ACL injury in a female cohort, where it is highly likely that similar results would emerge. This would help to ascertain whether they demonstrate the same deficits in MV and knee biomechanics. It would also allow clinicians and researchers to identify the broad deficits in both genders, especially given the significant differences that have been found to occur between genders. In addition, new research is warranted in order to improve the current rehabilitation programmes on offer to ACL-RC patients, which would help to reduce all these biomechanical impairments and MV deficits, as well as to facilitate their safe RTS activities. Most importantly, such studies would help to reduce the reinjury rate.

## The end

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# 12 Appendices

## 12.1 Supportive course and modules

| Course | Date |
|---|---|
| ○ **Completing a Learning Agreement & the PhD Progression Points** | 26-11-2015 |
| ○ **Intro to Endnote X7** | 16-12-2015 |
| ○ **Critical Writing** | 23-11-2015 |
| ○ **Neuroscience and neurological rehabilitation** | 11-02-2016 |
| ○ **Doing a Literature Review** | 15-12-2015 |
| ○ **Research Ethics for PGRs** | 18-02-2016 |
| ○ **Qualisys Workshop** | 17-02-2016 |
| ○ **Word scope** | 29-02-2016 |
| **LEAP Higher** | 24-02-2016 |
| ○ **Planning & writing a thesis** | 17-03-2016 |

## 12.2 International Conferences

# CERTIFICATE

This is to certify that

***Abdullah Alyami***

Presented an accepted poster during the congress:

**Relationship Between Movement Variability and Maximum Knee Valgus Angle during Functional Sports Tasks**

**SCANDINAVIAN SPORTS MEDICINE CONGRESS 2019**

**31 January - 2 February, 2019 in Copenhagen, Denmark**

| | |
|---|---|
| For DIMS: Kristoffer W. Barfod<br>Chair for the Danish Society of Sports Medicine | For DSSF: Karen Kotila<br>Chair for the Danish Society of Sports Physical Therapy |


Certificate of Recognition

conferenceseries.com

Conference Series and the Editors of Journal of Novel Physiotherapies, Journal of Yoga & Physical Therapy and Journal of Palliative Care & Medicine wish to thank

Prof/Dr/Mr/Ms. Abdullah Alyami

University of Salford, UK

for his Poster Presentation on

"Reliability of lower limb biomechanical outcome measures among healthy subjects using a 3D motion analysis during five specific sports tasks: Single-leg squats, single-leg landings, running, cutting 135 and cutting 90"

at the "5th International Conference on Physiotherapy"

held during November 27-29, 2017 in Dubai, UAE

Anand Shetty
University of St. Mary, USA

## 12.3 Awards during PhD time period

conferenceseries.com

Conference Series LLC
Best Poster Award

Awarded to Abdullah Alyami
University of Salford, UK

for presenting the poster entitled:
Reliability of lower limb biomechanical outcome measures among healthy subjects using a 3D motion analysis during five specific sports tasks: Single leg squats, single-leg landings, running, cutting 135 and cutting 90°

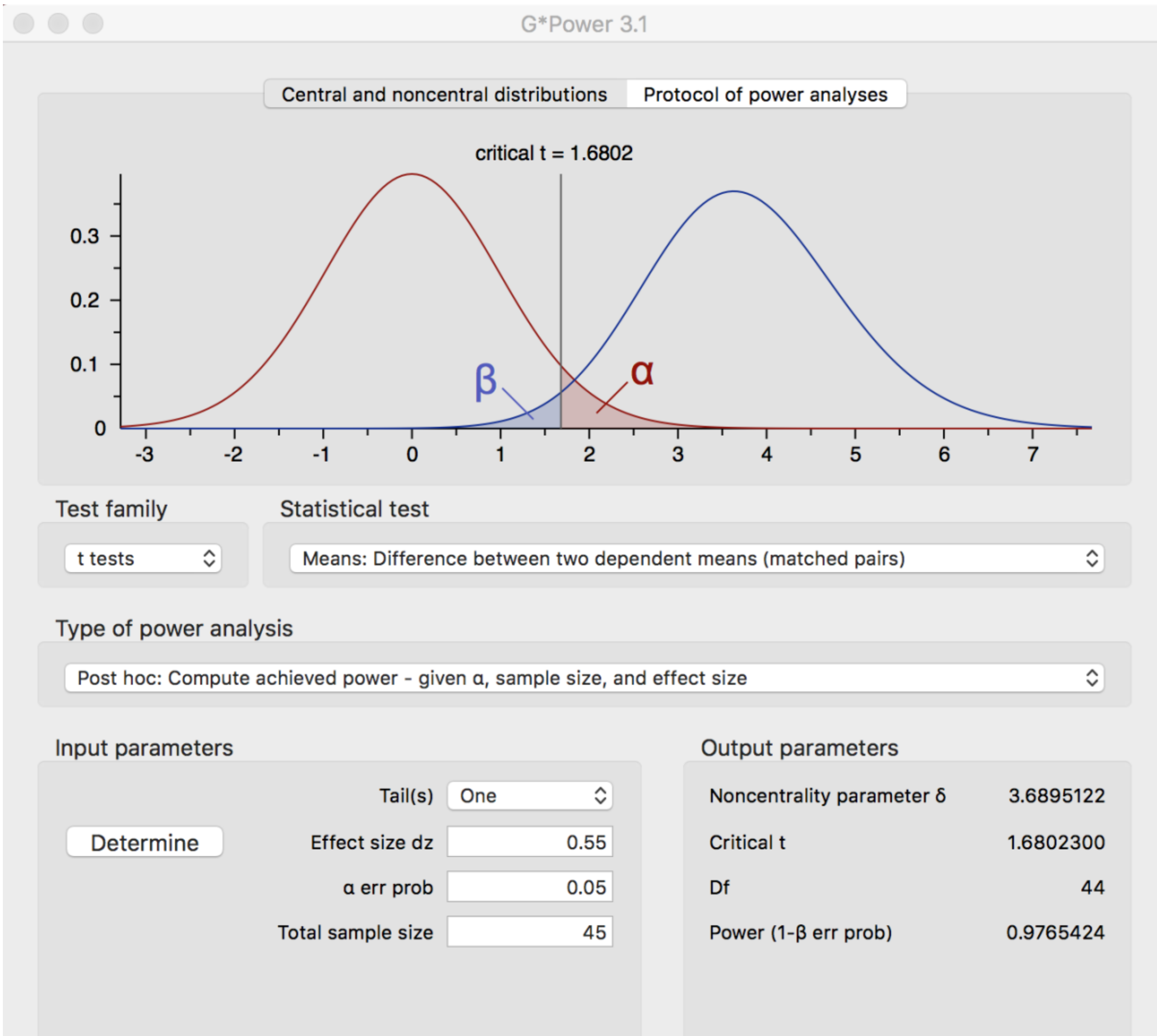
at the "5th International Conference on Physiotherapy"
held during November 27-29, 2017 in Dubai, UAE

The award has been attributed in recognition of research paper quality, novelty and significance.

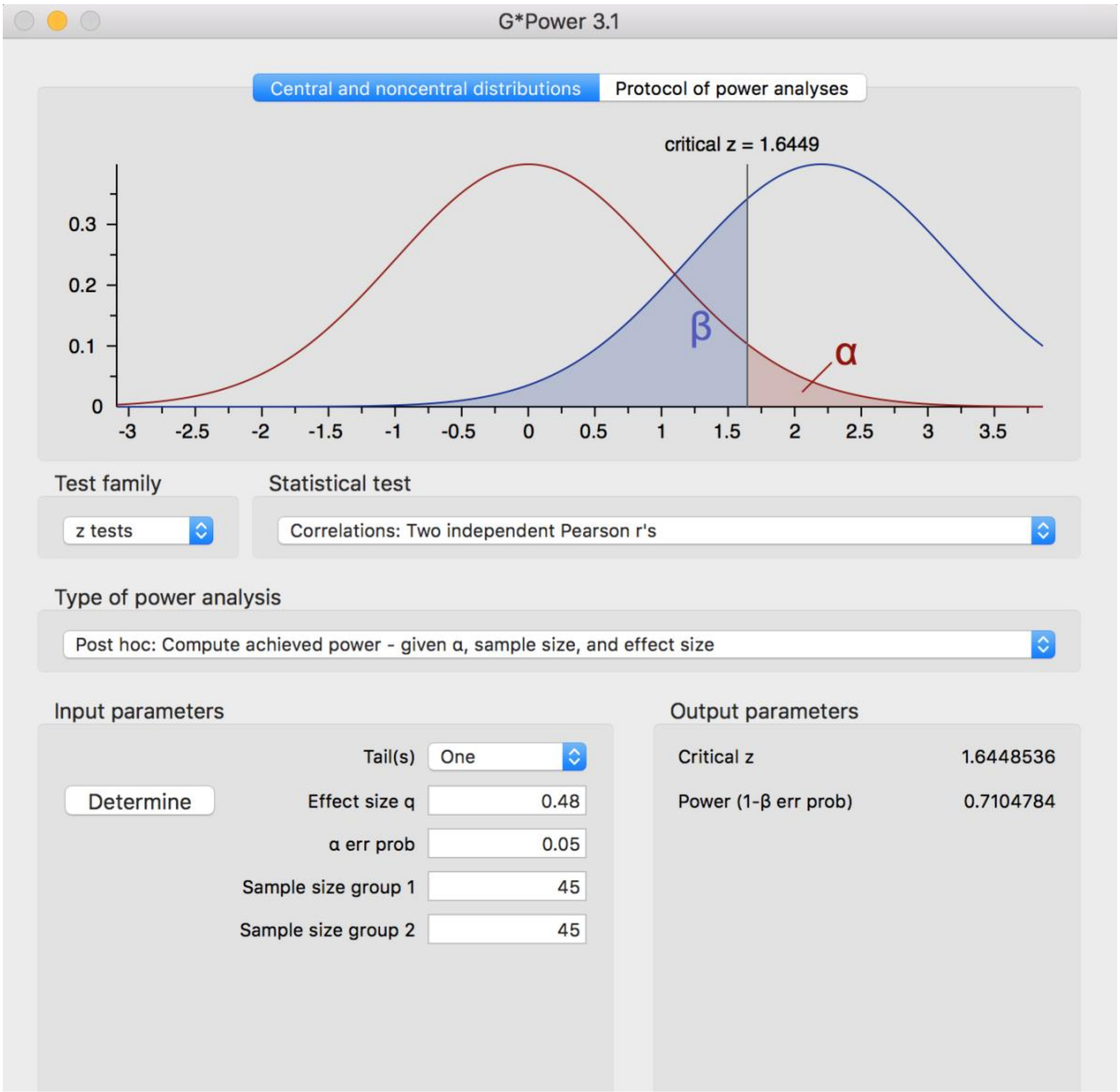
(PROFESSOR AREERAT SUPUTTITADA, MD)

Anand Shetty
University of St. Mary, USA

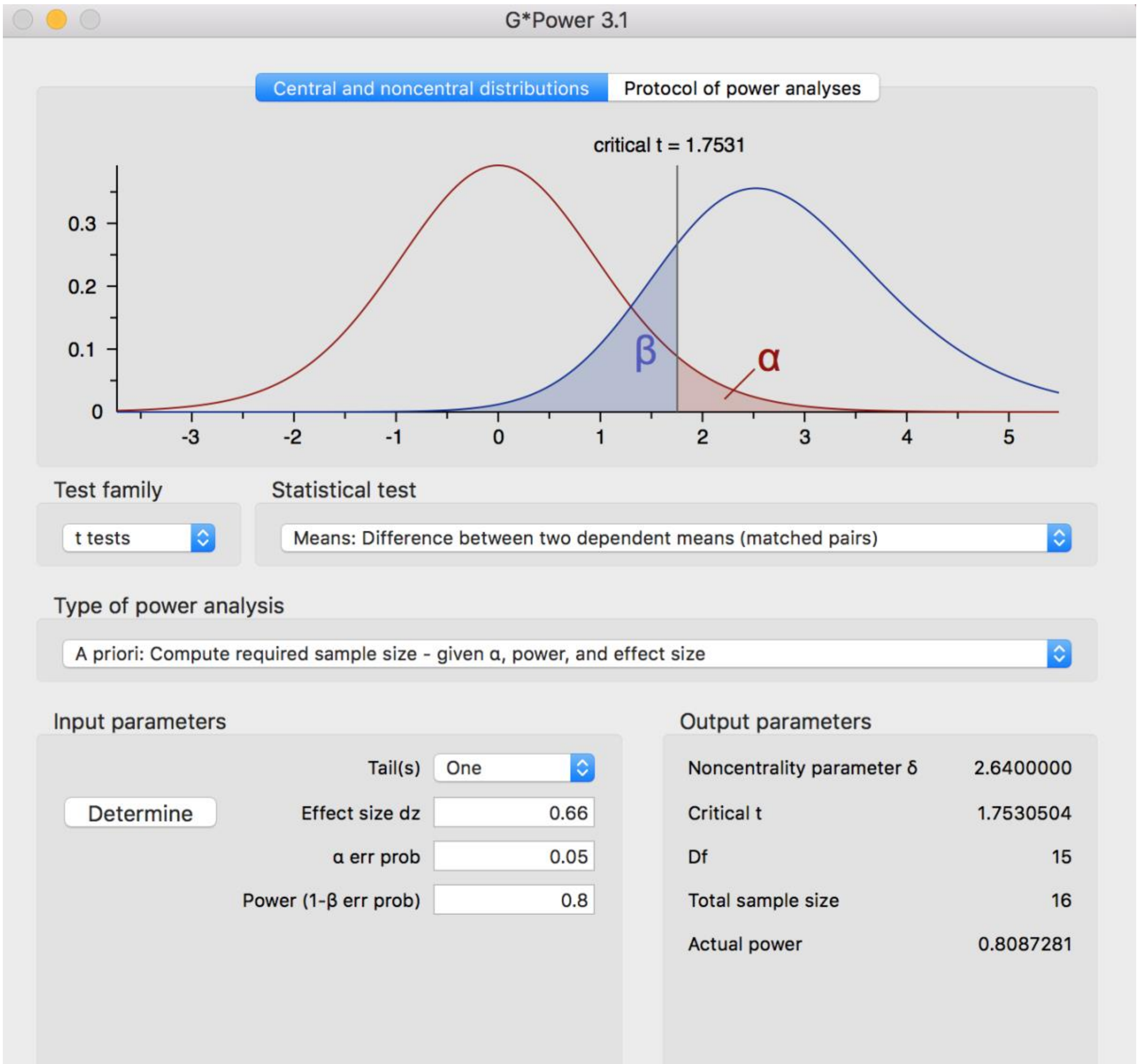
## 12.4 Power analysis using post hoc G*Power (Version 3.1.9.3) for Chapter 5



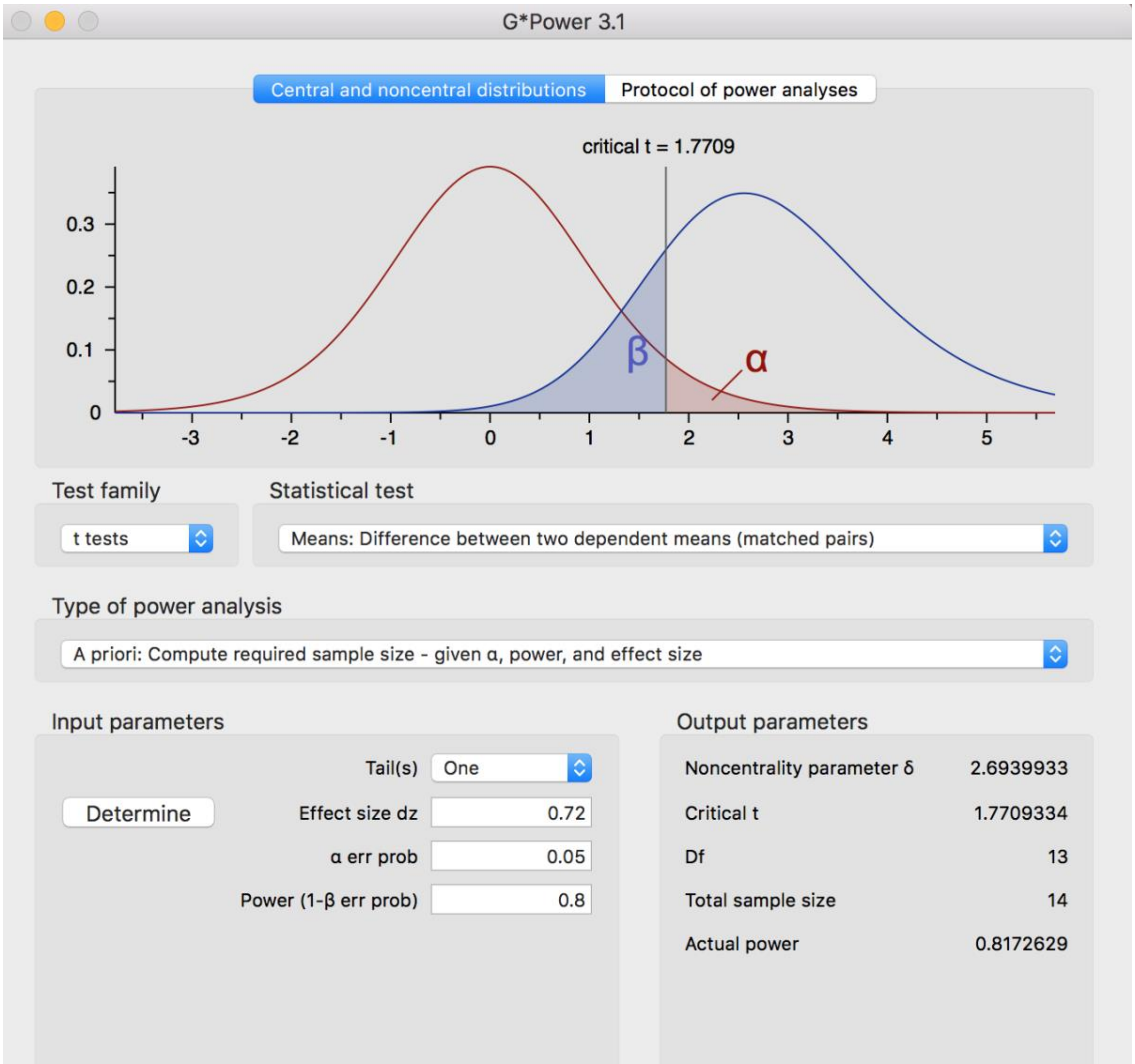
## 12.5 Power analysis using post hoc G*Power (Version 3.1.9.3) for Chapter 6



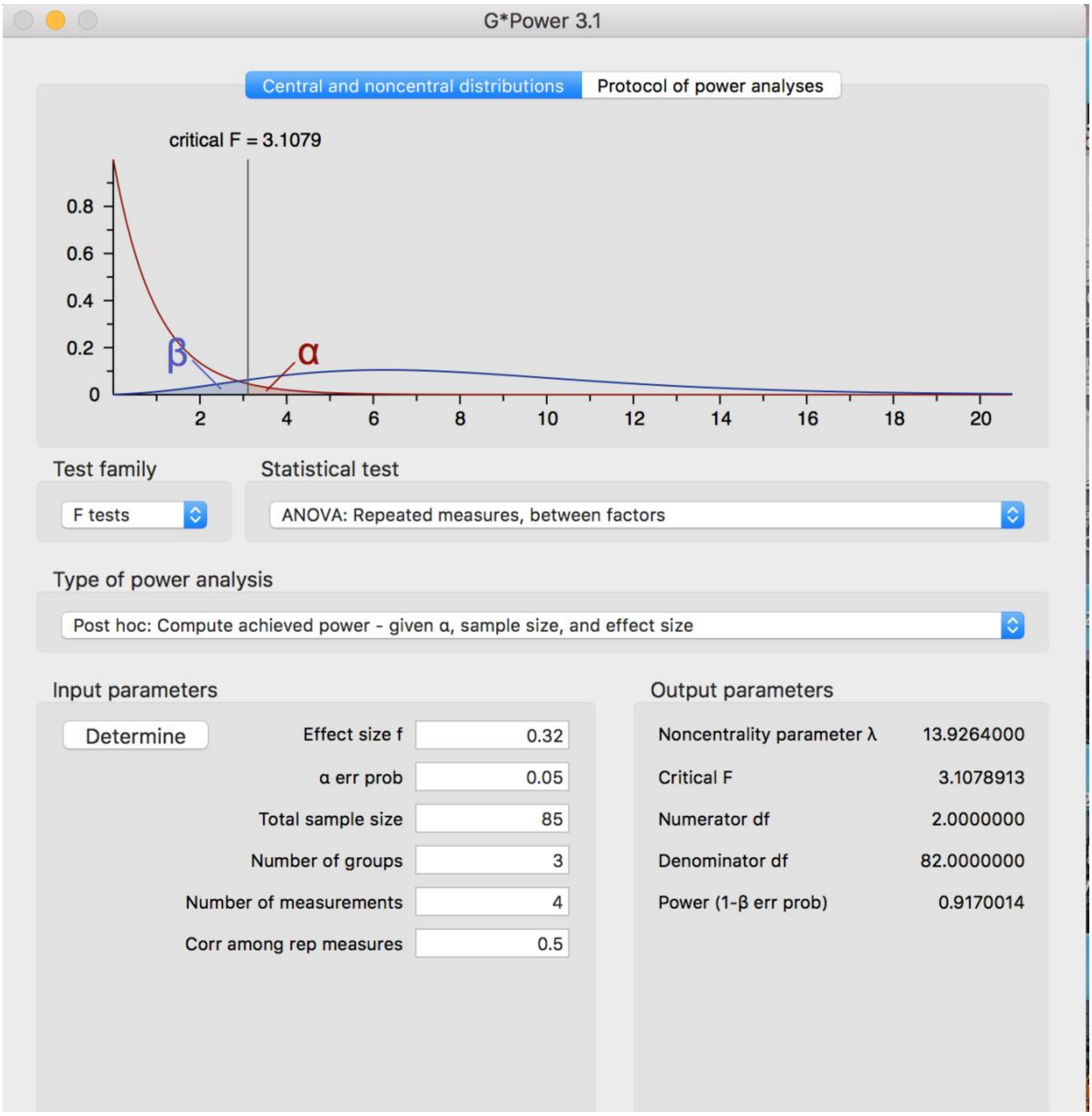
## 12.6 Power analysis using G*Power (Version 3.1.9.3) for Chapter 7



## 12.7 Power analysis using G*Power (Version 3.1.9.3) for Chapter 8



## 12.8 Power analysis using post hoc G*Power (Version 3.1.9.3) for Chapter 9



## 12.9 Research participant consent form

University of Salford MANCHESTER

**Research Participant Consent Form**

**Study title:**

Movement variability of biomechanical outcome measures when doing four specific sport tasks; single-leg squat, single-leg landing, running and cutting 90, among healthy subjects and individuals with ACL injuries Using 3D system.

**Ethics Ref No**:

| | | |
|---|---|---|
| 1. | I confirm that I have read and understood the participant information sheet for the above study (version3 11.3.18) and what my contribution will be. | Yes / No |
| 2. | My involvement in the study and its purpose has been fully explained to me. I have been given the opportunity to ask questions and any questions that I have asked have been answered to my satisfaction. | Yes / No |
| 3. | I understood that my participation in this test will include a number of tests which are detailed in the information sheet (version3 11.3.18) | Yes / No |
| 4. | I have the right to withdraw from the research at any time without objection from the researcher and I understand that my data may be used unless I request not to do so. | Yes / No |
| 5. | I have been informed that I will be compensated for my participation if i complete both testing sessions. | Yes / No |
| 6. | I understand how the researcher will use my responses, who will see them and how the data will be stored. | Yes / No |
| 7. | I understand that the results of this research may be published but that my name and identity will not be revealed at any time. All information as numbered codes in computer files will be kept confidential on the computer of Abdullah Alyami, that is only available for him. | Yes / No |
| 8. | I agree to take part in the above study as a volunteer | Yes / No |

Variability informed consent V3 11.03.2018

9. I have been informed that any questions I have at any time concerning the research of my participation, will be answered by researcher and I can contact them via e-mail address: a.m.n.alyami@edu.salford.ac.uk — Yes | No

10. I agree to have my details kept on a contact list and may be sent ethically approved information sheets for other studies, related to this project, to see if I would be interested in taking part. — Yes | No

Name of participant ……………………………………………………………..…………

Signature …………………..…………… Date ………………………….

Name of researcher taking consent

…………………………………………………………………….…………………

Signature …………………………………... Date …………………………

Variability informed consent V3 11.03.2018

## 12.10 Participant information sheet

**Participant Information Sheet**

**Study title:** Variability of biomechanical outcome measures when doing five specific sport tasks; single-leg squat, single-leg landing, running and cutting 90, among healthy subjects and individuals with ACL injuries Using 3D system.

You are invited to take part in a research study which will provide important information in regard with the understanding and prevention for Anterior cruciate ligament (ACL) injuries, which is a crucial ligament in the knee joint.

**What is the purpose of the study?**

The purpose of the study is to investigate movement variability and knee biomechanics during four sport tasks which includes single leg landing and sidestep cutting where most ACL non-contact injuries happen. This study will provide very important information regarding knee biomechanics and movement variability and will be used when developing ACL injury risk mitigation and ACL reconstruction rehabilitation RTS programmes.

.

**Why have I been chosen?**

You have been chosen as you are either a healthy individual or you are affected by ACL injuries and you have indicated that you are happy to consider taking part in the study.

**What will I have to do?**

You will be asked to wear a loose pair of shorts to expose your lower limb and a short top to visualise the trunk in order to allow the video cameras to record you during the study. Please note that no actual video images are stored as the cameras record reflective marker information so only dots will be saved on the screen.
Your age, height and mass will be measured by the researcher and following this, the researcher will attach 38 retro-reflective markers to the skin of your lower limb on both legs, as shown in the figure below.

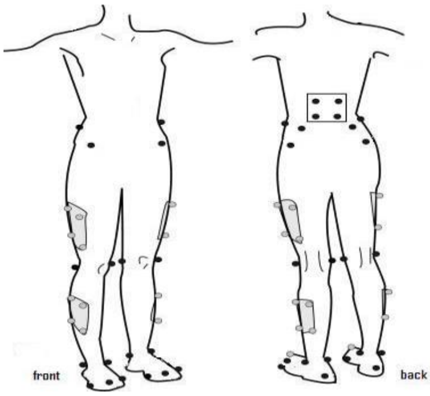



*Figure: Overview of the kinematic markers.*

You will be required to undertake 5 tests: a static standing assessment, single leg squat, single leg landing, running and cutting 90. These tests will be recorded in a controlled laboratory environment at the University of Salford or at the Imam Abdulrahman Bin Faisal University.

After the testing all of the markers will be removed. The measurement will be conducted over a single session and will take around 2 hours in duration.
Your identity will not be revealed: the video recording will only capture the markers which are on your lower limbs. All information obtained from these measurements will remain confidential and anonymous. All data will be directly coded with a study number, so you will remain anonymous to all but the researcher. The data will be stored on the computer of the researcher which is password protected. On completion of the study, the video-recordings will be archived but only the non-identifying information will be kept.

**<u>What benefits are involved in participating in the study?</u>**

You may not personally benefit from the study but you are providing essential data in regard to prevention and understanding ACL injuries among specific sport actions.

**If I participate in this study, can I also take part in other studies?**

As the study is not including a treatment or long-term assessment, taking part should not affect any other studies that you are involved in. However, if you are taking part in other research or would like to do so, please discuss it with the researcher.

**Is there an inherent risk involved?**

There is an inherent risk with any type of testing. However, this risk is reduced to a minimum by testing in a controlled laboratory environment.

**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 Research Centres Manager at the University of Salford (Mr Anish Kurien (a.kurien@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.

**What if I want to leave the study earlier?**

You are free to not participate in this study as well as drop out of the study at any time. If you want to withdraw or if you have any further questions about the project, you can contact the researcher.

**Who will see my details and results?**

All personal information will be kept confidential. The final results of the study will be available to you and may be published.

**What will happen to the results of the research study?**

A summary of the research findings will be sent to everyone who participates in the study. Important findings will be published in clinical and engineering journals.

**Contact information:**
Email: a.m.n.alyami@edu.salford.ac.uk
Phone number: 07446946209
Address:
University of Salford
Frederick Road Campus,
Salford, M6 6PU

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

## 12.11 Participant's details form

University of Salford MANCHESTER

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

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

1. **Personal information**

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

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

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

2. **Additional information**

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

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

Inactive ☐ moderately active ☐ highly active ☐

c. Give an example of a typical weeks exercise:

……………………………………………………………………………………….

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

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

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

15 Health Questionnaire/Exclusion Criteria:

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

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

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

## 12.12 Research participant consent form (Arabic)

جامعة الإمام عبدالرحمن بن فيصل
IMAM ABDULRAHMAN BIN FAISAL UNIVERSITY

**نموذج طلب الموافقة بعد التبصير للمشاركة في بحث**

- أنت مدعو من قبل (عبدالله محمد اليامي) للمشاركة في بحث علمي.
- عنوان الدراسة: تقييم ميكانيكا الركبه و التغير الحركي أثناء أداء أربعة أنشطه رياضيه في الرياضيين الأصحاء و الرياضيين الذين لديهم إصابة بالرباط الصليبي و الذين أجروا عملية الرباط الصليبي.
- اسم المنشأة التى اعتمدت البحث : جامعة سالفورد ببريطانيا.
- برعاية: جامعة سالفورد ببريطانيا.
- الباحث الرئيسى: عبدالله محمد اليامي طالب دكتوراه في العلاج الطبيعي.
- هذه دراسة بحثية يتم اجراءها على الافراد الذين سيتم اختيارهم للمشاركة فقط.
- لذا نرجو أخذ الوقت الكافى لمناقشة هذا الامر مع عائلتك وأصدقائك قبل اتخاذ قرار المشاركة.
- سبب اختيارك لهذه المشاركة فى هذه الدراسة هو اما انك شخص رياضي سليم او شخص رياضي لديه إصابة بالرباط الصليبي و خضع لعلاج غير جراحي و رجع لممارسة الرياضه او رياضي لديه إصابة بالرباط الصليبي و خضع لعلاج جراحي و رجع لممارسة الرياضه.
- الغرض من هذه الدراسة هو معرفة التغير الحركي و ميكانيكا الركبه أثناء أداء أربعة أنشطه رياضيه في الرياضيين الأصحاء و الرياضيين الذين لديهم إصابة الرباط الصليبي و الذين أجروا عملية الرباط الصليبي.
- طريقة الدراسة: يتم فحص المشارك بواسطة جهاز التحليل الحركي ثلاثي الابعاد أثناء أداء أربعة أنشطه رياضيه:
  1- نشاط تغيير الاتجاه اثناء الجري.
  2- نشاط الجري بشكل مستقيم.
  3- نشاط القفز على قدم واحدة.
  4- نشاط النزول على قدم واحدة.
- حيث متوقع أن يشارك حوالى 90 فرد فى هذه الدراسة والتى ستستمر لمدة ثلاثة أشهر
- يمكنكم التوقف عن المشاركة فى هذه الدراسة فى أى وقت دون أى عقوبة أو فقدان منفعة .
- المخاطر المحتملة:لا يوجد مخاطر محتملة خلال فحص المشارك بواسطة جهاز التحليل الحركي ثلاثي الابعاد.
- الاجراءات المتخذة لتقليل المخاطر:عن طريق إعطائك كل ما يكفي من الوقت للتدريب، الاحماء. بالإضافة إلى ذلك، في حال الإصابة لا قدر الله، سوف يتم تقديم الإسعافات الأولية المتوفرة في قسم العلاج الطبيعي. ومع ذلك، هذه الدراسة سوف تتم في قسم العلاج الطبيعي، وبالتالي لا يوجد مخاطر محتملة لحدوث أي إصابة لا قدر الله.

الفوائد المرجوة (مباشرة /غير مباشرة): تستطيع الاستفادة شخصيا من الدراسة عن طريق معرفة وضع الركبه الميكانيكي الحالي, بالإضافة للفائدة العامة من البحث وتحسين الحالة الوظيفية للمصابين بالرباط الصليبي.

جامعة الإمام عبدالرحمن بن فيصل
IMAM ABDULRAHMAN BIN FAISAL UNIVERSITY

- وصف لطرق العلاج البديلة المتوافرة خارج نطاق البحث ان وجدت: لا يوجد طرق بديلة سوى في المراكز الرياضية المتخصصة في تأهيل الرباط الصليبي للاعبين المحترفين وتقييمهم قبل عودتهم لممارسة الرياضة.
- كما أننا سنحافظ على سرية المعلومات الشخصية الخاصة بك ولن يتم ذكر أى بيانات شخصية أو عامة تدل على تحديد شخصيتك ولن يتم استخدام اسمك أو ذكر هويتك فى أى تقرير أو اصدار حول هذه الدراسة .
- لن تتحمل أى تكاليف متعلقة بهذه الدراسة ان وجدت ولكن الباحثين هم المسئولين عن تمويل هذه الدراسة .
- المشلركة في البحث أمر طوعى وان رفض المشاركة لن يترتب عليه أية عقوبة أو خسارة لمنفعة تستحقها بسبب اخر, كما أن لك الحق في الانسحاب من البحث في أية مرحلة من مراحله دون أن تتعرض لخسارة أو فوات منفعة تستحقها لأى سبب.
- توضيح المخاطر أو الأضرار التى يمكن أن تترتب على الانسحاب من البحث ان وجدت: لا يوجد
- أتعهد أنا الباحث الرئيسى بأنك ستحاط علما بجميع المعلومات التى قد تستجد خلال فترة اجراء البحث والتى يمكن أن تؤثر معرفتك لها في استمرار مشاركتك في البحث.
- في حال وقوع ضرر ناتج عن اجراء البحث عليك فان طريقة التعويض هى توفير العلاج لك من اي إصابة قد تحدث لك.
- اذا كان لديك أى استفسار أو شكوى حول هذه الدراسة الرجاء الاتصال برئيس قسم العلاج الطبيعي: د/ قاسم المعيدي

تلفون: 013333 / 00000 جوال: 0550033666

ملاحظة:

من خلال توقيعكم على هذا النموذج تكون قد وافقت على المشاركة فى هذه الدراسة البحثية.

وتكون قد أعطيت الفرصة الكاملة لفهم ماسيتم وأعطيت الفرصة لأى سؤال يتعلق بهذه الدراسة كما أنه سيتم وضع صورة من هذا الاقرار فى ملفك الطبى.

(يتم التوقيع من قبل المشارك)

| | |
|---|---|
| ( ) أوافق فى المشاركة فى هذه الدراسة | توقيع الشخص الذى حصل على الموافقة |
| الاسم : .................................. | الاسم : .................................... |
| التوقيع:.................................. | التوقيع :.................................... |
| التاريخ:.................................. | التاريخ:...................................... |

## 12.13 Participant information sheet (Arabic)

جامعة الإمام عبدالرحمن بن فيصل
IMAM ABDULRAHMAN BIN FAISAL UNIVERSITY

**معلومات عن الدراسه للمشاركين**

- **عنوان الدراسة** : تقييم ميكانيكا الركبه و التغير الحركي أثناء أداء أربعة أنشطه رياضيه في الرياضيين الأصحاء و الرياضيين الذين لديهم إصابة بالرباط الصليبي و الذين أجروا عملية الرباط الصليبي.

- **الباحث الرئيسي:** عبدالله محمد اليامي طالب دكتوراه في العلاج الطبيعي
- **العنوان جامعة سالفورد – بريطانيا**

**رقم الهاتف 0556779669**

- سيشرح لك عضو من فريق البحث محتويات هذه الدراسة وتأثيرها عليك. و يصف هذا الإقرار إجراءات الدراسة، والمخاطر والفوائد من المشاركة، وكيفية الحفاظ على سرية المعلوم ات. الرجاء اخذ الوقت الكافي في طرح الأسئلة لكي تتخذ قرارك ما إذا كنت ستشارك أم لا. وهذه الموافقة تسمى الموافقة المستنيرة. إذا قررت المشاركة في هذه الدراسة، سيطلب منك التوقيع على هذا الإقرار وستعطي نسخة لسجلاتك. وطوال هذا الإقرار اللفظ، "أنت" سوف يشير إليك.

- **لماذا تجري هذه الدراسة؟**

لمعرفة ميكانيكا الركبه و التغير الحركي أثناء أداء أربعة أنشطه رياضيه في الرياضيين الأصحاء و الرياضيين الذين لديهم إصابة بالرباط الصليبي و الذين أجروا عملية الرباط الصليبي.

- **كم عدد المشاركين في هذه الدراسة؟**

90 مشارك

- **ماذا سيحدث إذا شاركت في هذه الدراسة؟**

سيتم فحص المشارك بواسطة جهاز التحليل الحركي ثلاثي الابعاد أثناء أداء أربعة أنشطه رياضيه.

- **موقع الدراسة؟**

قسم العلاج الطبيعي – جامعة الامام عبد الرحمن بن فيصل بالدمام.

- **ما هو المتوقع مني خلال هذه الدراسة) ما هي مسؤولياتي)؟**

سيتم فحص المشارك بواسطة جهاز التحليل الحركي ثلاثي الابعاد أثناء أداء أربعة أنشطه رياضيه:

1- نشاط تغيير الاتجاه اثناء الجري.
2- نشاط الجري بشكل مستقيم.
3- نشاط القفز على قدم واحدة.
4- نشاط النزول على قدم واحدة.

- **ما هي مدة مشاركتي في هذه الدراسة؟**

120 دقيقه.

- **هل أستطيع إنهاء مشاركتي في هذه الدراسة؟**

نعم. يمكنك أن تقرر التوقف في أي وقت. فقط أخبر الباحث إذا قررت التوقف. ليوضح لك كيفية إنهاء مشاركتك بأمان.

لا أحد سيحملك على تغيير رأيك.

- **هل هناك مخاطر متوقعة إذا أنهيت مشاركتي في هذه الدراسة؟**

لا

- **ما هي المخاطر أو الآثار الجانبية التي يمكن حدوثها من جراء مشاركتي في هذه الدراسة؟**

لا يوجد مخاطر محتملة خلال اختبار قوة العضلات أو اختبار تحليل الحركة.

- **هل هناك فوائد من مشاركتي في هذه الدراسة؟**

تستطيع الاستفادة شخصيا من الدراسة عن طريق معرفة وضع الركبه الميكانيكي الحالي بالإضافة للفائدة العامة من البحث وتحسين الحالة الوظيفية للمصابين بالرباط الصليبي.

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- **ما هي الخيارات الأخرى؟**

لديك خيارات أخرى بدلا عن المشاركة في الدراسة:
لا يوجد طرق بديلة سوى في المراكز الرياضية المتخصصة في تأهيل الرباط الصليبي للاعبين المحترفين.

- **وما هي تكاليف المشاركة في الدراسة؟**

لن تتحمل تكاليف أي من أنشطة الدراسة.

- **هل سأتقاضى اجر نظير المشاركة في هذه الدراسة؟**

لن يكون هناك اجر.

- **هل سيتم الحفاظ علي سرية المعلومات الطبية الخاصة بي؟**

سنبذل قصارى جهدنا للتأكد من أن المعلومات الشخصية في سجلك الطبي تحظى بالسرية. ومع ذلك ، لا يمكننا أن نضمن الخصوصية التامة. قد يفصح عن معلوماتك الشخصية إذا اقتضى الأمر ذلك بموجب القانون. لن يتم الإفصاح عن اسمك أو معلوماتك الشخصية إذا تم نشر أو عرض نتائج هذه الدراسة.

- **ما هي حقوقي إذا وافقت على المشاركة في هذه الدراسة؟**
- قرار المشاركة في هذه الدراسة من اختيارك. لك حرية اختيار المشاركة في هذه الدراسة أو لا. كما يمكنك إنهاء المشاركة في أي وقت. مهما كان قرارك، لن يكون هناك أي عقوبة. الباحث قد يستخدم المعلومات التي تم جمعها قبل أن تترك لدراسة.

وفي حال الإصابة الناتجة عن هذه الدراسة بتوقيع هذا الإقرار لن تفقد أيا من الحقوق القانونية في طلب التعويض.

- **بمن يمكنني الاتصال إذا كانت لدي أي أسئلة أو مشاكل؟**

- قبل أن توافق على المشاركة هذه الدراسة، ستتحدث إلى أحد أعضاء فريق الدراسة المؤهلين ليخبرك عن هذه الدراسة. يمكنك أن تطرح الأسئلة حول أي جانب من جوانب البحث. إذا كان لديك المزيد من الأسئلة عن الدراسة، يمكنك السؤال في أي وقت. يمكنك الاتصال الباحث الرئيس علي الرقم: **0556779669**.

## 12.14 The ethical approval

University of Salford MANCHESTER

**Research, Enterprise and Engagement**
**Ethical Approval Panel**

Doctoral & Research Support
Research and Knowledge Exchange,
Room 827, Maxwell Building,
University of Salford,
Manchester
M5 4WT

T +44(0)161 295 2280

www.salford.ac.uk

13 September 2016

Dear Abdullah,

**RE: ETHICS APPLICATION–HSCR16-61 – 'Variability of biomechanical outcome measures when doing five specific sport tasks; single leg squat, single leg landing, running, cutting 180, and cutting 90 among healthy subjects and individual with ACL injuries Using 3D system.'**

Based on the information that you have provided, I am pleased to inform you that ethics application HSCR16-61 has been approved.

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

Yours sincerely,

Professor Sue McAndrew
Chair of the Research Ethics Panel

## 12.15 Participant's details form (Arabic)

معلومات الانشطه البدنية

العمر:

الوزن:

الطول:

| النشاط | عدد الساعات في الاسبوع |
| --- | --- |
| المشي | |
| الجري | |
| ركوب الدراجه | |
| السباحه | |
| تمارين رفع الاثقال | |
| حلقات التدريب | |
| فنون الدفاع عن النفس | |
| الرياضات الجماعية مثل(كرة القدم, كرة اليد, كرة السلة) | |

هل لديك إصـابه سابقة بالركبة　　نعم　　لا

هل لديك إصـابه سابقة بالطرف السفي　　نعم　　لا

## 12.16 Data collection completion letter from Imam Abdulrahman Bin Faisal University.

وزارة التعليم
Ministry of Education
043

جامعة الإمام عبدالرحمن بن فيصل
IMAM ABDULRAHMAN BIN FAISAL UNIVERSITY

المملكة العربية السعودية
Kingdom of Saudi Arabia

### إلى من يهمه الأمر

**اسم الطالب**: عبدالله محمد ناصر اليامي

تفيد كلية العلوم الطبيه التطبيقيه في جامعة الإمام عبدالرحمن بن فيصل ( جامعة الدمام سابقا) في الدمام بأن المدون اسمه أعلاه, قد قام بدراسة ميدانيه في مقر الجامعة بمعامل قسم العلاج الطبيعي في مدينة الدمام وتحت إشراف الأستاذ المشارك بالقسم سعادة الدكتور/ قاسم بن ابراهيم المعيدي والتي تتعلق بمجال بحثه لدراسة الدكتوراه في علم الحركة والإصابات الرياضية الجارية في جامعة سالفورد في بريطانيا. حيث أن الدراسة بدأت في ٠٢-٠٥-٢٠١٧م وانتهت في ٣١-٠٧-٢٠١٧م. هذا وقد تم إصدار هذا الخطاب بناء على طلبه لتقديمه إلى الملحقيه الثقافية السعوديه بلندن.

وتقبلوا خالص الشكر والتقدير،،،

**عميد كلية العلوم الطبية التطبيقية**

**د/ علي بن متعب الشامي**

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