FUNCTIONAL TASKS BEFORE AND AFTER AN ANTERIOR CRUCIATE LIGAMENT (ACL) RECONSTRUCTION: ARE THERE MECHANICAL DIFFERENCES?

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Saud Alarifi

ABSTRACT

Anterior cruciate ligament (ACL) injury is a major cause of significant time loss to sports participation, as well as reportedly leading to an increased risk of osteoarthritis (OA). Knee instability and functional adaptations are likely to occur following injury to the ACL, despite many ACL-deficient (ACLD) patients displaying no, or minimal, visible impairment. ACL reconstruction (ACLR) is the most common form of treatment for physically active individuals following an ACL injury. The aim of most individuals is to return to preinjury levels of physical activity after ACLR. However, most individuals experience persistent changes to lower extremity biomechanics well after completing structured rehabilitation and being cleared to return to activity. Despite this, there is little data available on individuals with an ACLD, or ACL-reconstructed ACLR knee and biomechanical alterations leading to the development of OA. Numerous studies that have investigated walking gait have found significant reductions in peak internal knee extensor moment, and small reductions in peak knee flexion angle, with individuals adopting a quadriceps avoidance gait pattern during walking. One of the most common activities pre- and post- surgery is running, and it is not known whether individuals before and after ACLR knee have different knee kinematic and kinetic patterns to healthy individuals. However, in general clinical practice, the option to ask the individual to run and to assess this is limited, and so a more space-optimised clinical assessment is needed. Therefore, the single leg squat (SLS) has been chosen as the measure to assess these individuals. No previous study has been found on kinematics and kinetics before and after ACLR during running and SLS. Therefore, the research question of this thesis is to determine whether there is an alteration in the kinematics and kinetics of hip and knee joints, along with the related risk factors for patellofemoral pain syndrome and OA, before and after ACLR during running and SLS.

This research aimed, in the first study, to establish within-day and between-days reliability for the use of 3D motion analysis to measure the biomechanical variables for running and SLS tasks. This study concludes that for between and within-day sessions, specific variables demonstrated good and excellent levels of consistency (ICC=0.80-0.99), and exhibited standard errors of measurement that have relatively low values.

The second study investigated the hip and knee joints' kinematics and kinetics six to eight months after ACLR, and compared the outcomes between the injured limb and non-injured limb (n=34), and a control group (n=34). This showed that ACLR individuals, despite a return to sport and being deemed medically fit, still have performance issues, which could be related to PF joint pathology and OA. This study found that the injured limb of the ACLR group showed a significant reduction in peak internal knee extensor moment and impulse, knee flexion angle and external knee adduction moment ($p=0.01$, $p=0.01$, $p=0.01$, $p=0.04$ respectively) compared to the control group during running. On comparing the injured and non-injured limbs in ACLR, an increase in hip internal rotation angle, coupled with a reduction in knee flexion angle, peak internal knee extensor moment and impulse ($p=0.01$, $p=0.01$, $p=0.01$, $p=0.01$ respectively) was found during running. Comparing the injured and non-injured limbs in SLS, revealed an increase in hip internal rotation angle coupled with knee adduction angle, in addition to a reduction in peak internal knee extensor moment ($p=0.01$, $p=0.01$, $p=0.04$ respectively). The control group compared to the injured limb of the ACLR group during SLS, showed reductions in peak internal knee extensor moment ($p=0.01$); whereas the non-injured limb of the ACLR group revealed an increase in hip internal rotation moment, and a reduction in peak internal knee extensor moment ($p=0.04$, $p=0.01$ respectively).

The third study investigated hip and knee joint kinematics and kinetics before, and three and six months after ACLR, during running and a SLS task to compare between the injured limb and non-injured limb (n=6), and the control group (n=6). This was to examine whether these factors develop over time, which could be related to PF joint pathology and OA. The findings show that there was a reduction in the peak internal knee extensor moment and impulse three and six months post ACLR between limbs, and in comparison to the injured limb for the ACLR group and the control group ($p=0.01$, $p=0.01$ respectively). In addition, significant differences were noted before, and three and six months after ACLR, during running ($p=0.01$, $p=0.01$ respectively), as well as SLS between limbs three months after ACLR. At the same time, within the ACLR group, there was a significant reduction in knee flexion angles during running three and six months after ACLR between limbs.

The results of this thesis show that following ACL reconstruction, individuals in this thesis showed some specifically altered knee joint kinematics and kinetics. The reduction in peak internal knee extensor moment and knee flexion angle was in an effort to reduce or avoid the contraction of the quadriceps; namely, quadriceps avoidance. These reductions may contribute towards patellofemoral joint disorders, thereby increasing the risk of degenerate joint disease commonly found post-surgery. The results of this thesis may help to guide the development of new or alternative treatment options for improving long-term joint health after an ACL injury.

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CHAPTER ONE

INTRODUCTION

1.1 Introduction

Anterior Cruciate Ligament (ACL) injury is one of the most common knee injuries; within the UK population, rates of injury are reported as being 30 per 100,000 people (Webb and Corry 2000). The re-rupture rate of those requiring surgical repair, depending upon the demands of the sport, are reported as being 2.3% to 13% within the top sporting populations (Myklebust and Bahr 2005). ACL injuries are often considered to be linked to degenerative knee changes, as well as early onset osteoarthritis (OA). This link has yet to be proven, although several studies have reported that those with an ACL injury develop OA earlier than anticipated compared to those who have not suffered an injury (Lohmander et al., 2007). The literature clearly demonstrates a link between ACL injury and early onset OA following ACL reconstruction, despite improved functionality when compared to those before ACL reconstruction (Lohmander et al., 2007). Further information relating to underlying causal mechanisms is required, with more targeted outcome measures, following an in depth objective investigation of function.

Oiestad et al., (2010) and Neuman et al., (2008) assessed patients who developed early OA following ACL injury and surgery, and they found that between 16% and 62% of patients developed radiographic OA within 15 years. The differences reported between studies may in part be related to whether removal or repair of the load bearing meniscus has occurred, with those receiving meniscus removal having higher levels of OA when compared to those with none or only minor repairs (Lohmander et al., 2004). When the femoral trochlea groove or patella facets are affected by synovitis and subchondral bone change, osteophyte formation and loss of articular cartilage tends to occur, which is characteristic of the occurrence of patellofemoral joint OA. Over 100 research investigations have published findings on tibiofemoral (TF) joint OA following reconstruction and injury related to ACL (Lohmander et al., 2007), but there have been significantly fewer studies of individuals with ACLR that have developed patellofemoral (PF) joint OA. In reports on using radiographic images between two and 15 years after surgery, an incidence of 36% appears to be typical (Hertel et al., 2005; Ahn et al., 2012; Lohmander et al., 2007; Neuman et al., 2009). A more recent study reports that five to 10 years after ACLR, 31% of participants had developed TF joint OA, and 47% had developed PF joint OA, according to radiographic images (Culvenor et al., 2013). These findings suggest that the incidence of PF joint OA could increase over time following surgery, but insufficient studies have investigated time-periods longer than five years to evaluate the incidence of PF joint OA.

It is often considered that the stabilising role of the ACL following disruption may affect the normal movement pattern in the knee, such as loading and shifting the femur's position on the tibia, which creates abnormal stresses through the knee cartilage, and may lead to cartilage degeneration (Andriacchi et al., 2009; Andriacchi et al., 2004; Barrance et al., 2006). Despite this hypothesis, no published research study has yet linked measures of function and biomechanics to degenerative knee changes linked to the longitudinal assessment of structures.

It is problematic to generalise strategies adopted in ACL injured patients, because there are differences in surgical rehabilitation and other interventions from the time that the injury and/or surgery takes place (Risberg et al., 2009; Tashman et al., 2007; Hurd and Snyder-Mackler 2007; Rudolph et al., 2001). During running, adaptations in ACL injured groups include avoiding full extension and decreasing knee extensor moments (Berchuck et al., 1990; Rudolph et al., 2001; Karanikas et al., 2009). There is a paucity of literature related to biomechanics in running and single leg squats (SLS) before and after ACL reconstruction (ACLR), which requires further evidence in order to come to firm conclusions about adaptations or compensations before and after ACLR. Many studies on running have focused only on sagittal plane kinematics and kinetics after ACLR (Sigward et al. 2015; Karanikas et al. 2009; Bush-Joseph et al. 2001). Frontal plane kinematics and kinetics after ACLR are included in the current study to provide a more detailed evaluation of adaptations within the knee, transverse and frontal plane kinematics, and kinetic adaptations within the hip. Such studies often have limitations regarding biomechanical differences between ACLR groups and control groups during running, such as small sample sizes and using a treadmill to run. As an assessment tool in physiotherapy and rehabilitation settings, the SLS is useful for assessing a patient's recovery and performance following a knee injury (Weeks et al., 2012).

The kinematics and kinetics between before and after ACL reconstruction have been compared in various studies. Knoll et al., (2004) examined the walking gait pattern of ACL injured individuals during walking before surgery, and two weeks, six weeks, four months, eight months and 12 months after surgical reconstruction. The subjects showed a quadriceps avoidance gait prior to surgery and six weeks post-surgery. The strategy of a quadriceps avoidance gait is described by Berchuck et al., (1990) as a reduction in internal extensor moment. This gait pattern was still evident five weeks post-surgery, but more similar to that of the control group. The reason for this is that before and after ACL reconstruction, the patient may have different kinematics and kinetics, which results in different performance abilities during running. Moreover, this may have consequences for long-term rehabilitation and return to sport for the individual. Currently, no research has been found that details kinematics and kinetics before and after ACL reconstruction during running. As in general clinical space, the option to request an individual to run and make an assessment is limited, a more space-optimised clinical assessment is needed. Therefore, the SLS has been chosen as the measure to assess such individuals in the current study.

Whatman et al., (2011) investigated the links between lower-limb kinematics during running and those occurring during SLS. Yamazaki et al., (2010) compared the kinematics of SLS between ACL deficient (ACLD) individuals and those in a control group. The injured leg of ACLD individuals showed less knee and hip external rotation angles, less knee flexion and more knee adduction than those of the non-injured subjects, which are risk factors for PF and TF joints OA. The investigation was carried out before and after ACL reconstruction because the patient may have different kinematics and kinetics, which results in different performance abilities during sporting activities, and this may have consequences for knee OA. To date, no studies have reported on a comparison between kinematics and kinetics before and after ACLR during an SLS task.

Therefore, the research question in this thesis is to determine whether there is an alteration in the kinematics and kinetics of the hip and knee joints, along with the related risk factors for patellofemoral pain syndrome and OA, before and after ACLR during running and SLS.

CHAPTER TWO

BACKGROUND

2.1 Introduction

The aim of this chapter is to present the background information on anterior cruciate ligament deficient (ACLD) and anterior cruciate ligament reconstructed (ACLR) knees. It will explore whether ACLD and/or ACLR result in altered kinematics and kinetics that may lead to tibiofemoral joint osteoarthritis (TF joint OA) or patellofemoral joint osteoarthritis (PF joint OA). According to Shabani (2015), an anterior cruciate ligament rupture will lead to instability of the knee and to biomechanical knee changes. Currently, athletes with an ACL injury generally undergo ACL reconstruction. However, it has been observed that athletes with ACLR and ACLD knees are not able to return to sport at the same level as before the injury (Paine, 2016). This research will examine the role of ACLR and ACLD in the pathogenesis of PF joint OA and TF joint OA. Because of the role of the ACL in knee joint biomechanics, it is essential to understand the biomechanical changes to ACL deficient (ACLD) and ACL reconstructed (ACLR) knees, and to examine whether these changes lead to kinematic and kinetic alterations, which in turn lead to PF joint OA or TF joint OA. The first section explores the prevalence of ACL injuries, and illustrates why this research is necessary. The second section examines the mechanisms of ACL injuries. The third section explores the reconstruction methods used for ACLD knees, and shows how OA inevitably occurs in ACLR knees. The fourth section examines whether knee biomechanics are altered in ACLR knees, and whether these altered biomechanics are a reason for the onset of OA. The fifth section examines the pathogenesis of OA in the TF joint for ACLR and ACLD knees, while the sixth section explores the pathogenesis of OA in the PF joint of ACLR and ACLD knees. The next section explores the syndrome of patellofemoral pain and its relation to ACLR and ACLD. The remaining sections include an evaluation of the functional performance tests of squatting and running; an explanation of the gaps in the literature, and the research question. The content of this chapter is set out in Figure 2.1 below.

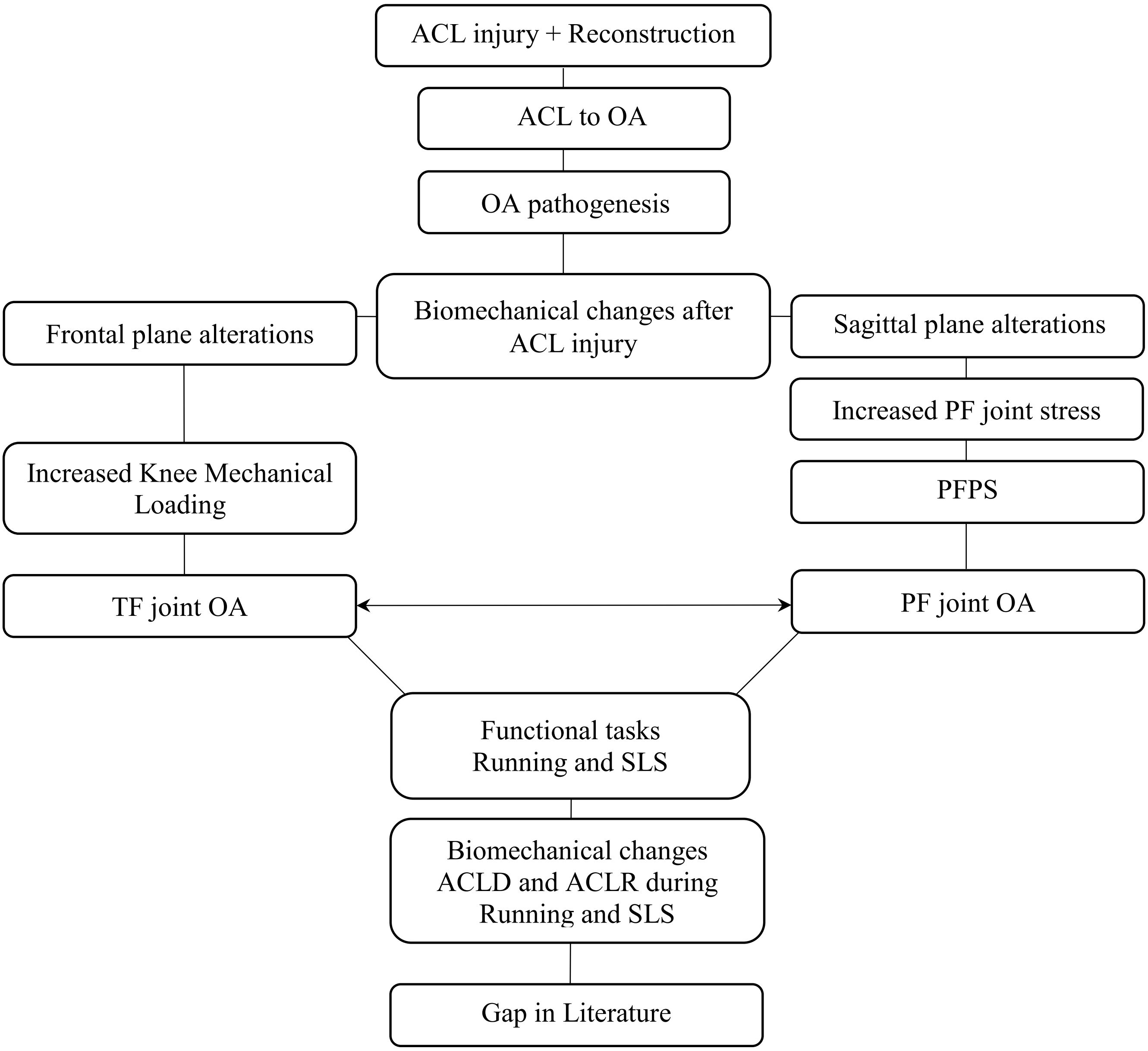


Figure 2.1 Chapter content

2.2 Prevalence of ACL Injuries

The ACL is the main knee ligament; its function is to provide stability and prevent anterior tibial translation and internal rotation. Nordenvall et al., (2012) explain that ACL rupture is one of the most common types of orthopaedic trauma that occur around the world. Frobell et al., (2007) point out that in Sweden alone, 81 out of 100,000 people suffer from an ACL rupture annually. Similarly, Nordenvall et al., (2012) states that around the world, approximately 78 per 100,000 people suffer from an ACL rupture trauma. In addition, Bates (2015) claims that the ACL rupture rate is up to 84 per 100,000 of the population in the US, with around 100,000 to 200,000 people suffering from this injury annually in the US alone.

ACL ruptures occur most frequently amongst athletes and sports persons during sporting activities, such as landing, running and cutting during football, basketball, netball, handball and volleyball matches (Mountcastle et al., 2007; Myklebust et al., 2003; Renstrom et al., 2008; Olsen et al., 2004; Boden, et al., 2000). As a result of an ACL rupture, the knee becomes unstable due to higher anterior tibial translation and anterolateral rotation (Eberhardt, 2002). Myklebust et al., (2003) point out that 58% of all handball players in Norway who have suffered an ACL injury do not play again at pre-injury levels, resulting in most players either leaving the sport altogether or continuing to play at lower levels of competition. According to Shah (2010), 37% of American football players in the US do not continue with the sport post ACL injury treatment. Lohmander et al., (2004) states that 50% of all female Swedish football players discontinue the sport post ACL trauma. Kijowski et al., (2012) and Wissman et al., (2014) explain that female soccer, football and basketball athletes are two to eight times more at risk of an ACL rupture than their male counterparts. It may be inferred, therefore, that the consequences of an ACL rupture are most acutely felt by athletes and sports persons. Thus, the consequent inability to efficiently and effectively perform sporting activities has greater career, financial and lifestyle implications for sports persons than for others.

2.3 Mechanisms of ACL Injuries

The most common mechanism involved in an ACL injury is a non-contact mechanism, as up to 70% of ACL injuries occur during non-contact incidents (Pasanen et al., 2008; Faude et al., 2005; Agel 2005). Studies by Boden et al., (2000) and Olsen et al., (2004) indicate that ACL injuries occur when reducing the speed of movement suddenly, due to a fast landing, or changing the direction of movement. According to Olsen et al., (2004), when the knee is positioned along the valgus, under condition of full extension and foot strike, there is a higher chance of an ACL injury occurring. It is because of this, that research conducted by Arendt and Dick (1995), Griffin et al., (2000) indicates that ACL injuries occur most frequently in those playing football, volleyball and basketball, all of which involve landing, reductions in speed, and quick changes in direction.

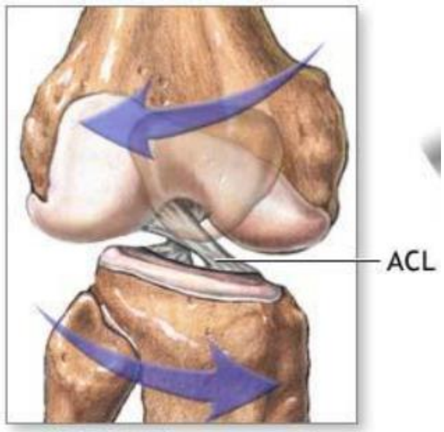


Figure 2.2 ACL Injury (Paine, 2016)

According to Shimokochi and Shultz (2008) non-contact injury occurs when there is a sudden reduction in speed while changing direction or when landing from a jump. Shultz et al., (2012) explains how extending the hip joints, along with the position of knee joints, during landing also results in higher anterior tibial shear forces occurring during the landing, which can result in a rupture, as indicated in Figure 2.2 above. Some research has used video analysis to estimate lower-limb joint angles (Olsen et al., 2004; Krosshaug et al., 2007). These studies, which define dynamic knee valgus or point of no return, show evidence of valgus knee collapse; knee close to full extension and tibia externally rotated; and hip slightly flexed, adducted and internally rotated when landing (Ireland 1999; Hewett 2005). Most ACL injuries occur with the knee in a valgus position and close to full extension, and close to foot strike (Olsen et al., 2004; Boden et al., 2000).

The net effect of ACL and its concomitant ruptures is an unstable knee joint, which can result in the 'giving way' of the knee (Shabani, 2015). This is particularly true of athletes who cannot participate in those sports where they have to change direction constantly, such as football. According to Andriacchi and Favre (2014), there is a reduction in proprioception after ACL disruption, leading to the inability of the muscles around the knee to respond properly to loads that are applied to the knee during daily activities and sports. This can lead to further knee instability and further injury. Lephart et al., (2000) state that each time the knee gives way, there is further subluxation, compression and shearing, which can lead to the menisci further tearing, along with the capsular ligaments becoming stretched and the articular surface being damaged. The inference that may be made here is that patients suffering from ACL deficiency experience alterations in knee function. According to Hurd and Snyder-Mackler (2007), only a few ACLD persons are able to perform with the same level of knee functionality without complaining of knee instability. This group is called the 'copers'. The 'adapters' are those who have resumed normal sporting activities with no occurrences of the knee giving way, while the 'non-copers' are those who experience frequent episodes of giving way at the knee, including whilst performing activities of daily living (Rudolph et al., 2001). According to Mather et al., (2014), Roos et al., (2011) and Sward et al., (2013), the non-copers form the largest group of patients suffering from the impact of ACL deficiency, and require ACL reconstruction to try and restore the stability of the knee joint.

2.4 ACL Injury, Reconstruction Methods and OA

Conservative physiotherapy treatment is one option for treating an ACL injury, for both copers and non-copers. Freedman et al., (2003) explain that the copers are those without concomitant injuries who no longer wish to engage in physical activity. For the copers, conservative treatment is recommended, consisting of physiotherapy with a physical therapist. However, researchers such as Jacobsen (1977), McDaniel and Dameron (1983) and Satku et al., (1986) all indicate that conservative treatment of ACL is associated with an increase in OA. Hence, researchers such as Küllmer et al., (1994), Johnson et al., (1984) and Maletus and Messner (1999), point out that if ACL injuries are left untreated, this will predispose the person to the early onset of OA. Tashman et al., (2004) note that without surgical reconstruction, the knee will remain unstable and susceptible to OA. However, some researchers claim that there is a lack of evidence that ACLR reduces the risk of OA (e.g. Frobell et al., 2013; Øiestad et al., 2009). Indeed, higher rates of OA have been reported following surgical compared to non-surgical management of the ACLD knee (Daniel et al., 1994; Fithian et al., 2005; Kessler et al., 2008; Neuman et al., 2008). This has led to the theory that an ACLR may propagate the development of knee OA, possibly due to the return to high-impact cutting and pivoting sporting activity (Ajuied et al., 2014), which non-operative treatment algorithms often advise against (Grindem et al., 2012; Neuman et al., 2008). However, there is little evidence to suggest that individuals following a non-operative management strategy should avoid high-impact activities. Following a structured and well-controlled rehabilitation and education program to optimise equipoise has been shown to be beneficial. For example, a recent RCT (which is typically absent in other studies and clinical practice) revealed that activity levels and OA rates do not differ between conservative or surgical management strategies (Frobell et al., 2013).

The most common treatment for ACL injury is ACL reconstruction. According to Mall et al., (2014), there are as many as 100000 to 150000 ACL reconstruction surgeries performed every year in the United States. This is despite the lack of information to support the assumption that ACLR restores the pre-operative functionality of the knee. Starman et al., (2008) indicate that all ACLR surgeries replace the ruptured ACL using auto / allografts, even with artificial tissue. The aim is to restore the stability of the knee and its functionality to before injury levels. In this regard, Crawford et al., (2007) point out that the latest type of graft reconstruction method is to introduce double bundle grafts that resemble the front-medial and the back-lateral features of the original ACL. However, Kongtharvonskul et al., (2013) and Zhang et al., (2014) claim that there is no evidence that the double bundle grafting technique is more effective than the single bundle grafting technique in restoring the stability of the knee or its functionality.

The most common graft types for ACL reconstruction are bone patellar tendon bone (BPTB) and hamstring tendon (HT) grafts. Researchers such as Pinczewski et al., (2007), Spindler et al., (2004) and Roe et al., (2005), all state that grafting has produced good clinical outcomes, with most patients achieving functionally stable knees immediately after surgery. However, research by Pinczewski et al., (2007) and Roe et al., (2005) also found that patients who had undergone BPTB developed OA five to seven years after surgery. Roe et al., (2005) state that 14% of all the respondents in their study developed OA seven years after HT surgery. These views are corroborated by Hui et al., (2011) and Bourke et al., (2012), who indicate that 51% of their respondents who had undergone BPTB, and 7% of HT treated patients, suffered from OA after 15 years. Thus, graft reconstructions can also result in OA, just as it may occur in conventionally treated and in single or double bundle reconstructions.

In a single bundle reconstruction, either the posterior lateral (PL) bundle or the Anterior Medial (AM) bundle of the ACL is reconstructed. However, Ristanis et al., (2003) and Tashman et al., (2004) found that in single bundle reconstructed knees, OA has been found to occur more frequently than in a normal knee, and they attribute the onset of OA to abnormal knee rotations, which result in abnormal knee kinematics. In the double bundle reconstruction technique, both the AM and the PL bundles are reconstructed. Researchers such as Seon et al., (2009), Kondo et al., (2008) and Hemmerich et al., (2011) all found significant improvement in the rotational stability of the double bundle reconstructed ACL knee compared to the single bundle reconstruction. Hemmerich et al., (2011) found that there was much less external rotation shifts in the double bundle reconstruction compared to single bundle reconstructions. Lam et al., (2011) found that the tibial rotation of intact and double bundle reconstructed knees was the same, which means that rotational knee stability is restored after double bundle reconstruction. However, Fu and Lin (2013) state that even in double bundle reconstructed knees, the rate of occurrence of OA is similar to ACLD knees. Ventura et al., (2012) also point out that while double bundle reconstructions are associated with superior clinical treatment of ACL, there is no difference in the rates of occurrence of OA when compared to single bundle reconstructed knees. This reveals the occurrence of OA, even following a reconstruction that purportedly restores the functionality of the ACL and the knee. There is mixed evidence concerning whether this more complex and technically demanding surgery can restore knee biomechanics more effectively than a traditional single-bundle approach (Hemmerich et al., 2011; Seon et al., 2010). In addition, there do not appear to be any differences in symptomatic or functional outcomes (Kongtharvonskul et al., 2013; Muneta et al., 2007; Streich et al., 2008; Zhang et al., 2014). However, as the double-bundle technique is a relatively new procedure, there are limited reports on the long-term outcomes.

Neumann et al., (2008) and Louboutin et al., (2009) claim that ACL injuries are the main cause of the development of osteoarthritis (OA) in the patellofemoral (PF) and tibiofemoral (TF) joints. According to Egloff (2012), an ACL rupture leading to OA is the most common cause of disability and mobility impairment across the world. Al-Hadithy et al., (2013) state that there are 100,000 new cases of ACL induced OA being reported globally every year. Fu and Lin (2013) explain how an ACL injury causes considerable social, psychological and financial problems for patients, and in France alone the cumulative health costs of treating ACL induced OA doubled over the period 2003 to 2013. The implication of the above evidence is that ACL rupture is a frequently occurring injury around the world. It leads to OA and subsequent mobility impairment, with considerable financial and health burdens for patients.

Knee OA is a burdensome condition associated with considerable economic, health and personal costs (Hunter et al., 2014). Knee OA results in increased pain, reduced physical function and impaired quality of life (QoL), and is responsible for over 600,000 knee arthroplasties in the US annually (Losina et al., 2012), which is an effective but expensive procedure for end-stage OA (Callahan et al., 1994; Katz et al., 2007; Paxton et al., 2010). Younger adults with OA, such as those following ACLR, can face a range of challenges not typically associated with OA in the older population, including from work and parental responsibilities, as well as competitive sporting careers. This may cause younger adults with OA to experience greater psychological distress than their older counterparts (Gignac et al., 2006). The socio-economic costs of an ACL injury over the lifetime of a patient can also be substantial (Mather et al., 2013), due largely to the long-term disability associated with the development and progression of knee OA. Indeed, an ACL injury is a contributing factor in up to 30,000 knee arthroplasties annually in the US (Mather et al., 2013). However, the long-term survival of a prosthesis is a major concern in the younger, more active patient (Weng & Fitzgerald, 2006); not only is a revision arthroplasty often inevitable (Rand et al., 2003), but post-operative outcomes are poor in those aged less than 60 years (Elson & Brenkel, 2006). It is not surprising that younger individuals with post-traumatic OA are more likely to be advised to wait, albeit while suffering considerable pain and symptoms, until a joint replacement is a more viable management strategy. Clearly, research resources need to focus on changing this long-term trajectory.

From the above analysis, it may be inferred that ACL ruptures increase the chance of OA occurring. Lohmander et al., (2007) found that OA occurred in 50% of respondents who had undergone ACL injury up to two decades previously. Hui et al., (2011) and Von Porat et al., (2004) also indicate that in the case of adolescents and young adults who have suffered from an ACL injury, they will almost inevitably get OA before they turn 40 years old. Therefore, it may be inferred that an ACL injury can result in the early onset of OA. Frobell et al., (2013) and Oiestad et al., (2009), also state that ALCR does not necessarily stall the onset or the progression of OA. In regard to this, Daniel et al., (1994), Neuman et al., (2008), Kessler et al., (2008) and Fithian et al., (2005) explain that there are more instances of OA occurring in ACLR knees compared to ACLD knees. It may be inferred that with post-ACL reconstruction, there is a significant increase in the development of OA in comparison to the general population. While OA occurs in untreated knees, it also occurs in reconstructed knees. ACL reconstruction, therefore, does not offer significant protection against OA. These inferences are corroborated by Shrier et al., (2006), Gillquist and Messner (1999) and Claes (2013) who point out that ACL reconstruction in fact increases the risk of the occurrence of OA. Reconstructive treatments only restore the mechanical constraints of the ACL, but are not able to prevent OA. Thus, further analysis is required to identify the factor(s) that cause OA in reconstructed ACL patients.

The cohort studies conducted by Lohmander (2007), Von Porat et al., (2004) and Neumann (2008), all of which used similar radiographic methods and OA criterion, found the prevalence of OA to be uniform, at 51% for female and 41% for male athletes 12 to 14 years after their ACLs were ruptured. In these cohort-based studies, where approximately two thirds of the respondents agreed to a radiographic examination, there is the possibility that the occurrence of OA has been overestimated. This is because those respondents who bear all the symptoms of OA may be more interested in participating in such research in comparison to healthy respondents. As Lohmander's (2007) study indicates, the relationship between knee OA detected through radiographic methods and symptoms is tenuous at best. Lohmander (2007) also points out that radiography has always been used to show outcomes for studies on OA progression. Nevertheless, Hannan et al., (2000) explain that in radiographic studies, there is a limited relationship or association between sensitivity to change and to relevant outcomes for patients. For an accurate association to be made, it is necessary to standardise image acquisition and assessment processes (Shabani, 2015). Kotecha et al., (2013), claim that it may be more efficient to use magnetic resonance imaging systems that can visualise joint structures, tissue composition and sensitivity to change. Another limitation of the studies by Lohmander (2007) and Neumann (2008) is that they were not able to identify the factors leading to the onset of OA. It may be noted here that the lack of statistically significant variations are because of type 2 errors, which can result in subjective interpretations of negative findings. Other deficiencies in the studies by Lohmander (2007) and the Neumann (2008) are their poor methodological quality and lack of critical analysis. There is a requirement for significant improvements to be made in the quality of methodologies in order to understand the actual efficacy of treatments being performed on ACLD knees.

2.5 ACL Reconstruction, Altered Knee Biomechanics and OA

According to Fu and Lin (2013), ACL injuries will initiate OA, with current treatment options unable to stop its progression. Cameron et al., (2000) state that ACL reconstructions lead to joint instability, which triggers the onset and progression of OA. Injury to the ACL increases the anterior sliding motion between joints, which results in altering the mechanics of articular contacts (Fritschy, 1993). The cartilage of the knee is thickest where contact pressures are greater (Hui et al., 2011). When the ACL is injured, the amount of joint motion increases, resulting in a tibiofemoral offset that transfers contact forces to areas where the thinner cartilage is less able to offer support to such forces, increasing the shear stresses at the interfaces of the cartilage and bones (Cameron et al., 2000). This in turn results in larger external adduction moments, which increase medial compartment loading, resulting in the accelerated progression of OA. Joint instability also results in higher stress on the secondary joint stabilisers, further exacerbating the problem of OA. Chaudhari et al., (2008) point out that ACLD knees suffer from knee instability, with altered levels of compression and tension occurring in different parts of the cartilage after the rupture of the ACL, which results in premature OA.

These views seem to suggest that ACL reconstructions lead to changes in the biomechanics of the knee joint, which lead to OA. This inference is corroborated by Sajovic et al., (2006) who claim that ACL injuries can alter knee biomechanics, which may lead to OA. Thus, reconstruction procedures only restore ligament function, without restoring the original biomechanics of the uninjured knee. Dejour et al., (2013), Feller (2004), Webster et al., (2014) and Allen et al., (1999) carried out biomechanical studies to establish whether the kinematics and kinetics of ACL reconstructed knees match with those of normal knees. They all conclude that while current reconstruction techniques restore knee stability along one plane, this may not be possible along all planes of motion. Bourke et al., (2012) attribute this to changes in the structure of the grafts; their intra-articular placement, and their tension levels when compared to normal ACLs.

The idea of the altered kinematics of ACL reconstructed knees leading to OA is proposed by Almedkinders et al., (2004). They found that despite reconstruction to re-establish the anterior and posterior stability of the knee joint, the neutral tibiofemoral contact areas are located towards the anterior of the tibial plateau. This results in kinematics that differ from those of the normal knee. Logan et al., (2004) found that grafts result in altered tibiofemoral contact conditions, which leads to continuous anterior subluxation of the lateral tibial plateau under weight bearing conditions, and when the knee is flexed between 0° and 90°. Tashman et al., (2004) found that after ACL reconstruction, patients ran with their knees externally rotated by 3.8° and adducted 2.8° more than the control knee, a year after their operations. Jonsson et al., (2004) also found a positive shift in the knee pivots of reconstructed knees when compared to normal knees. It may be inferred from these results that ACL reconstructive techniques do not restore normal kinematics, and this must be responsible, at least in part for the onset of OA and its progression.

According to Herzog et al., (2004) and Andriacchi et al., (2004), a factor causing OA in ACL reconstruction knees is the altered loading of the joints. ACL reconstructions lead to altered joint motion, causing changes in loading and muscle function. There is a decrease in the force of the muscles, with the control mechanisms of the knee becoming disrupted, and these changes result in the knee cartilage becoming weak. Thus, joint unloading and weak muscle control are primary factors that result in OA formation in ACL reconstructed knees. Williams et al., (2004), explain that ACL injuries and reconstructions cause alterations in muscular function, and this results in the start of OA. This is either due to too much loading in areas not designed to support such loads (Andriacchi et al., 2004; Carter et al., 2004), or because loads are not sufficient enough to maintain metabolic homeostasis (Herzog et al., 2004; Carter et al., 2004).

Altered kinematics as a cause of OA has been proposed by Almedkinders et al., (2004). Tibial subluxation results in altered knee kinematics, and this causes alterations in the rolling motion of the tibiofemoral joint, leading to an increased chance of OA occurring. Almedkinders et al., (2004) suggest that while such subluxation may not be the cause of alterations in knee kinematics, it is most likely that these alterations in knee biomechanics, induced by changed kinematics, are the most important reason for the onset of OA post ACL injury. According to Chaudhari et al., (2008), ACLD knees exhibit altered load bearing, causing an alteration in gait kinematics that in turn results in thinning of the cartilage in the knee.

Gao et al., (2012) investigated whether ACL reconstructed knees exhibit biomechanical changes, using 3D gait analysis. There were two sets of respondents- those with normal knees and those with ACL reconstructed knees, and they were requested to ascend and descend stairs. It was found that there was a residual varus deviation of the tibia, which caused a statistically significant reduction in the range of extension, amongst the patients with ACL reconstructed knees. Hall et al., (2013) found that people with ACL reconstructed knees compensate for the reduced extensor moments of the knee by increasing hip extensor moments whilst climbing stairs. They assume that this movement results in greater loading on the cartilage of the knee joints post ACL reconstruction surgery. Tashman et al., (2004) examined the kinematics of ACL reconstructed knees under conditions of dynamic functional loading as respondents ran downhill, post ACL reconstruction surgery. It was found that there was a statistically significantly greater external angle of rotation of the tibia when compared to normal knees. It was therefore suggested that excessive tibial rotation results in extra load being placed on the cartilage of the knee, resulting in a predisposition to OA. Hauser et al., (2013) claim that while reconstruction reduces stress in the posterior medial compartment of an ACLD knee, it results in very high contact stresses in other parts of the knee due to multi-planar variations in knee kinematics, which results in OA.

Pahnabi et al., (2014) contrasted the uninjured legs of ACLR group with the healthy legs of a control group. It was found that in those individuals with ACLR knees, there was overloading of the uninjured leg. This led to greater stress being placed on the uninjured leg, which in turn led to compromised performance. Bonfim et al., (2008) point out that a reduction in the neural signal transmissions in the injured leg because of an ACL injury, can lead to the malfunctioning of the motor control system, which cannot control two limbs using different sensory inputs. In order to avoid asymmetric control, the performance of the uninjured leg is affected. The research by Bonfim et al., (2008) also found changes in performance when ACL group respondents moved about using their uninjured legs. In addition, the research by Callaghan et al., (2002) and Evans et al., (2004) indicates compromised performance of the uninjured leg in those with an ACL injury. It is inferred that sensory information transmission is impaired in the injured leg, which in turn leads to poor motor control in both legs. This shows that the uninjured leg is affected as well.

There are several inferences that may be made from the above analysis. While the primary injury to the ACL can contribute towards the development of OA, it is the secondary injuries in the form of instability and alterations in the normal kinematics of the knee that further progress OA. Alterations in knee kinetics result in biomechanical changes, which in turn, cause shifts in loading to areas that are not able to cope with the increased stress, resulting in degradation of the articular knee cartilage. Once OA is initiated, loading due to altered kinematic and kinetics results in further progression and the degradation of the cartilage, making the knee more susceptible to further injury, instability, and degenerative changes in the long term. Current surgical reconstruction techniques can reproduce the original ACL anatomy, but fall short of restoring the original kinematics and kinetics of the knee joint, resulting in a higher incidence of OA compared to normal knees. Moreover, a comparison between the uninjured and operated on knees of the same individuals has revealed considerable differences in kinematics, and this is manifested in altered biomechanical postures. Osteoarthritis of the knee may be divided into Patellofemoral joint OA (PF joint OA) and Tibiofemoral joint OA (TF joint OA).

2.6 Pathogenesis of OA at the Tibiofemoral (TF) Joint

This section illustrates how altered kinematics and kinetics at the TF joint can be a cause of the onset of OA.

2.6.1 Kinematics & Kinetics of the Normal TF Joint

An analysis of the kinematics and kinetics of the normal TF joint is necessary to understand the maximum limits of angles of rotation and movement. This information can be used to show how kinematics and kinetics become altered in ACLR knees, and how this leads to the onset of OA.

The tibia moves freely with respect to the femur, and is therefore able to move in anterior– posterior, central – side and proximal – distal directions (Jakob and Staubli, 1992). In addition, the tibia can also turn around in the flexion – extension, varus – valgus and internal – external directions. This means that the TF joint can move in all three translations and three rotations of the knee joint, collectively termed six degrees of freedom, as shown in Figure 2.3. Lawrence et al., (2008) explain that these six degrees of freedom are important to the functioning of the TF joint, for flexing the knee and bearing the weight of the body.

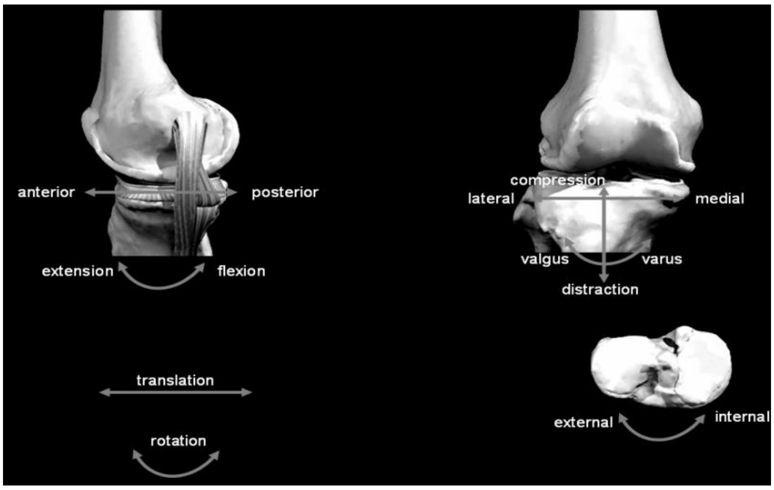


Figure 2.3 Movements of the TF Joint (Thambyah, 2004)

Flexion – Extension Rotation: Most knee motion happens along the plane where the flexion-extension of the TF joint occurs. During walking on level ground, the range of motion in the sagittal plane is up to 25 degrees during the standing phase and 50 degrees during swing (Nadeau et al., 2003). During stair climbing, this angle has been found to be 75 degrees when the feet make initial contact with the steps, and 100 degrees in swing (Nadeau et al., 2003). During squatting, knee angles can reach a peak of 160^0 (Nagura et al., 2002). Karholm et al., (2000), Nakagawa et al., (2000) and Hill et al., (2000) conducted in-vivo studies using MRI and found that flexion-extension rotations are accompanied by a shift in TF joint contact points. The medial femoral condyle shifts by 4mm, with shifts of 15mm in lateral backward movements. Komistek and Dennis (2003) found that during gait motions, the lateral condyle moves 4.3mm posteriorly. The implication of the aforementioned studies is that the maximum flexion angle at the normal TF joint is $120^0 – 150^0$ and extension angle is $5^0 – 10^0$.

Anterior – Posterior Translation: Both the ACL and the PCL offer primary restraint against anterior posterior translation motion. The ACL is taut during flexion, allowing for some internal rotation and limiting anterior tibial translation from 5 mm to 10 mm (Dennis, 1996).

Varus – Valgus Rotation or Abduction / Adduction: There is no varus – valgus rotation possible at the TF joint. Because of this, when the knee is fully extended, frontal plan motion is impossible (Cheng et al., 2010). However, Robbins and Marly (2009) point out that for up to 30^0 of knee flexion, there is passive adduction of a few degrees only, with maximum adduction occurring during the swing phase of the knee flexing motion, and the maximum abduction occurring during heel strike. The adduction / abduction range was observed to be a maximum of 11 degrees. Hurwitz et al., (2002) found that during stair-climbing, when the angles of the knee flexion are larger, the knee adduction angle is a maximum of 5 degrees vis-à-vis maximum knee internal valgus moments, and this is more than during flat ground walking, when it is a maximum of 2.5 degrees.

Medial – Lateral Translation: Mundermann et al., (2005) state that medial – lateral translation in the knee is quite small, and ranges from 1 mm to 2 mm.

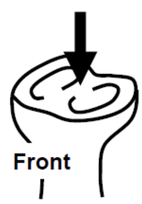
Compression / Distraction translation: The translation along the proximal – distal axis includes the amount of space between the tibia and femur when the knee is hanging free, and the allowable cartilage deformation. Shakoor et al., (2008) note that during compression – distraction, the displacement ranges from 2 mm to 5 mm to reduce the impact between the tibia and the femur from the effects of meniscus compression.

The studies above highlight the importance of the range of motion along the six degrees of freedom in the TF joint, and they detail the maximum angles of rotation / movement that are possible at the TF joint. The angles mentioned in the above studies also describe the ranges and reproducible patterns of motion along the six degrees of freedom for a healthy, adult human knee. The inference that may also be made here is that kinematic movements that breach the limits of angles of rotation / movement at the TF joints can also alter the kinetic forces' incident on the TF joint.

Flexion & Compression Load: Mow and Hayes (1991) have summarised the key flexion angles of the TF joint for various activities and their corresponding compressive loads, as illustrated in Table 2.1.

Table 2.1 Flexion Angles and Compressive Loads (Mow and Hayes, 1991)

• **Cycling**	**60-100 deg**	**1.2BW**
• **Walking**	**15**	**3.0**
• **Stairs**	**60**	**3.8**
• **Stairs**	**45**	**4.3**
• **Squat-rise**	**140**	**5.0**
• **Squat-down**	**140**	**5.6**



From Table 2.1, it can be observed that the flexion angle changes for different activities. The TF joint has to bear three times the person's body weight for a simple activity such as normal walking, while for activities such as squatting, the TF joints have to bear up to six times the body weight.

2.6.2 Alterations in the kinematics & kinetics of the TF Joint leading to OA

According to Butler et al., (2007), the loading in the knee joint can be as much as three times that of the person's body weight during walking, with the medial compartment of the TF joint bearing more load than the lateral compartment. The implication here is that determining the TF joint's contact area is important in order to evaluate the weight bearing capacity of the TF joint. Kettelkamp and Jacobs (1972) point out that the average contact area in the medial plateau is 1.5 times greater than the area on the lateral plateau. Thus, notwithstanding the fact that kinetic moments in the medial compartment are greater than those in the lateral compartment, the contact forces may not vary much between the two compartments, when there is greater force distribution in the medial compartment due to the enhanced contact area.

According to Andriacchi et al., (2009), as the femur turns over the articular tibial surface, the movements exhibit rotational as well as anteroposterior translation. The thickness of the cartilage over the TF joints varies due to the loading that occurs during normal walking. Hamai et al., (2009) point out that cartilage is thickest on the lateral side of the tibia, and least thick on its medial side. Carter et al., (2004) and Andriacchi et al., (2009) opine that whenever there is an alteration in knee kinematics, there is a shift in load from the common articular contact areas to infrequently loaded articular areas that are not frequently loaded. This altered loading results in fibrillation of the collagen network; loss of matrix surface, and more surface friction and shear stresses, all of which lead to cartilage degradation, which in turn leads to the onset of OA.

These views are corroborated by Vincent et al., (2012) who point out that the two tibial surfaces at which the lateral and the medial femoral condyles make contact are shaped differently, as illustrated in Figure 2.4.

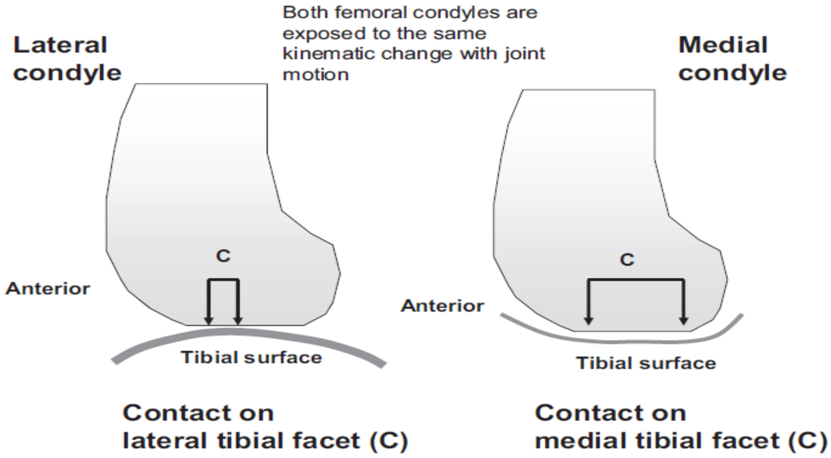


Figure 2.4 Contacts on Lateral and Medial Tibial facets (Vincent et al., 2012)

Vincent et al., (2012) state that when there are any kinematic changes that lead to the same joint movement, the medial femoral condyle will interact more with the concave tibial surface that has less thick cartilage thickness than the lateral condyle, resulting in reduced contact with the convex tibial facet which has thicker cartilage. This daily loading, along with increases in internal rotation, results in kinetics that adversely affect the areas of thick cartilage, first causing it to break down. In addition, prolonged exposure of these thin cartilage areas to more internal rotation will result in greater cartilage wear, which will in turn induce OA. According to Wise et al., (2012), changes in tibiofemoral kinematics that result in OA include the amount of internal turning, the point of turning, and the quantum of exposure on the medial compartment.

According to Hunter et al., (2007), malalignment of the knee, which places the TF joint in a varus position, results in excessive loading on the medial portion of the joint. An increase in varus malalignment results in degeneration of the joint, causing OA. Figure 2.5 shows how during dynamic kinematic movements, the medial joint loading can be measured using the peak external knee adduction moment.

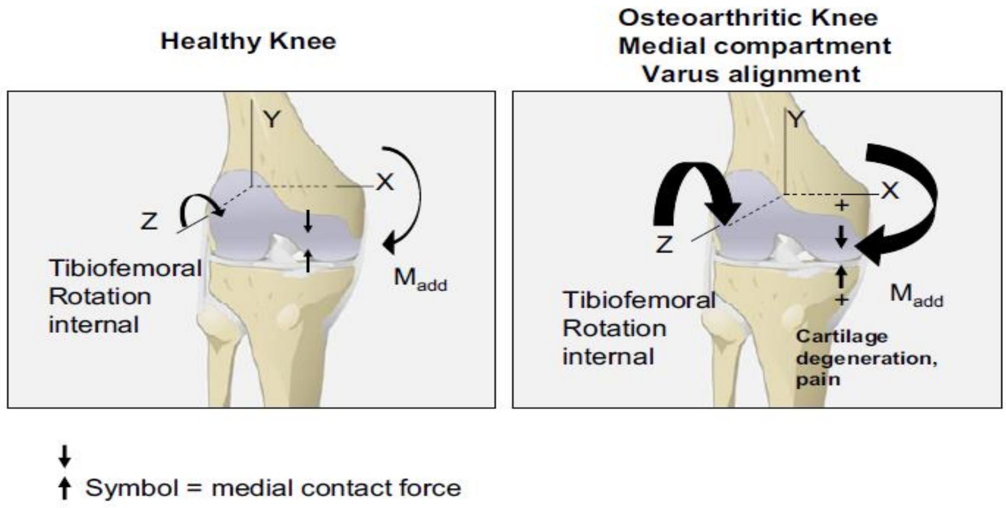


Figure 2.5 External Knee Adduction moment for the TF Joint in Healthy and OA knees (Vincent et al., 2012)

Peak knee adduction moment occurs whenever the feet touch the ground, with the line of action of the kinetic force vector passing medial to the knee joint. The greater the distance between the vector force and the knee joint, the greater the level of adduction forces and the higher the loading on the medial joint (Hunter et al., 2007). Varus alignment in turn causes the knee joint to move laterally to the position of the feet on the ground, which also increases adduction moment. In Figure 2.5, the thicker the arrows, the greater the rotation moment. A constantly higher adduction, along with increased internal rotation, results in higher medial contact forces over time. This results in degeneration of the cartilage and the development and progression of OA. Mundermann et al., (2004) point out that in order to compensate for the pain of OA, and to unload the medial compartment of the knee, people with TF joint OA walk more slowly than those with normal knees, showing that kinematic changes in gait can be an indicator of TF joint OA. From the studies discussed above, it may be inferred that altered kinematics result in changes in kinetics, which damages the cartilage. This is the initial step that leads to the onset of OA.

Hart and Spector (1993), Anderson and Felson (1988) and Cooper et al., (1994) examined TF joint OA among Asians and Westerners, and concluded that differences in knee kinematics predisposed the former respondents to a higher incidence of TF joint OA. Their research indicates that wherever people engage in activities involving frequent deep flexion, such as knee bending and squatting, coupled with excessive weight, this results in higher incidences of TF joint OA. Zhang et al., (2004) point out that prolonged squatting predisposes the knee to TF joint OA. Nagura et al., (2002) states that squatting results in higher external flexion moment around the knee than normal ambulatory activities where the peak moments generated are between 90° and 150° of flexion angle. This corresponds with the outer limits of the maximum deep flexion angles (Nadeau et al., 2003; Nakagawa et al., 2000). The implication here is that with such large extensor moment occurring about the knee at deep flexion angles, the contact forces and stresses occurring at the TF joint have to be considered in any study related to the onset of OA.

Cartilage is most thick at points of high contact in the TF joint. Studies by Kurz et al., (2005), Li et al., (2013), Andriacchi et al., (2009) all indicate that the thickness of the cartilage is dependent on loading. Excessive loading changes the thickness of the cartilage. In general, the amount of cartilage in the TF plateau is more than that in the back part of the lateral condyle and the front part of the medial condyle. Koo and Andriacchi (2007) point out that this differential cartilage thickness corresponds to contact patterns during normal human movement such as walking. Andriacchi et al., (2005) also explain that one of the differences between the front and side concerning the thickness of the cartilage is due to the internal rotations of the TF joint that occur during walking.

The external knee adduction moment should be considered when studying loading of the knee (Noyes et al., 1992). Andriacchi (2007) claims that the external peak knee adduction moments that occur when walking impact both the middle and side parts of the knee. Because of this, Sharma et al., (1988) and Mundermann et al., (2008) point out that OA that occurs in the medial compartment is linked to an increase in external peak knee adduction moments. According to research conducted by Sharma et al., (1998) and Mundermann et al., (2008), the external knee adduction moment may also be used to estimate the outcomes of OA treatment, and the magnitude of OA disease and its progression. Their research also indicates that variations in loading between healthy and injured OA knees means that cartilage responds variably once it starts becoming degraded, revealing that the ability of the knee joint to adapt to repetitive loading during walking activities is reduced.

There are several inferences that may be made from the above studies. Whenever there is malalignment of the TF joint, contact stresses on the surfaces of the TF joint are altered. This condition is exacerbated according to the nature of the kinematic activity performed by the person, and the fact that the TF joint bears almost three times the body weight of the person during normal walking, and six times the body weight during activities such as squatting. The weight incident on the TF joint for all of these activities increases with other factors conventionally attributed to OA onset, such as obesity. The TF joint is essentially a weight bearing structure, with the location of its centre of gravity determining the distribution of these forces. With varus malalignment, the centre of gravity shifts medially, as does the centre of maximum joint pressure, resulting in increased / excessive loading on the medial compartment of the TF joint. These kinematic changes manifest primarily through increases in knee flexion angle, increasing the kinetic force of the knee adduction on the TF joint cartilage, and resulting in its gradual wear and tear and the onset of OA. The external manifestation of these phenomena is a decrease in the knee extension capability of the individual, indicated by slower walking speeds. It is necessary to examine whether these kinematic and kinetic changes correspond with those that occur in ACLR knees, and if ACL can therefore be considered one of the key causes of TF joint OA. The next section examines PF joint OA, as Culvenor et al., (2013) point out that ACLR also results in PF joint OA along with TF joint OA.

2.7 Pathogenesis of OA at the Patellofemoral (PF) Joint

This section illustrates how altered kinematics and kinetics at the PF joint can be a cause of the onset of OA.

2.7.1 Kinematics & Kinetics of the Normal PF Joint

The most important function of the PF joint is to facilitate knee extensions by improving the efficacy of the quadriceps muscle, through increasing the moment arm of the muscle force with respect to the centre of rotation of the knee (Haxton, 1945). The patella improves the distribution of patellofemoral compressive forces on the femur by increasing the contact areas during flexion, as illustrated in Figure 2.6. Moreover, the patella controls the extensor mechanism through centralisation of the divergent pull from the other four quadriceps muscles, and transmitting these forces to the patella tendon (Schindler and Scott, 2011).

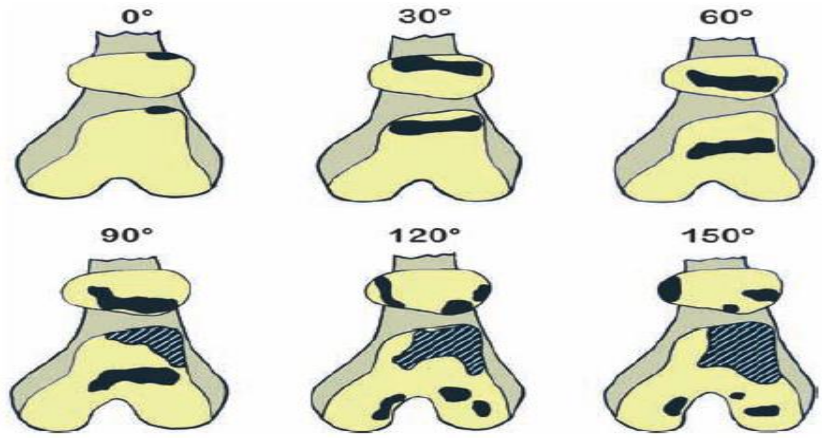


Figure 2.6 PF Contact Areas at Various Degrees of Knee Flexion (Schindler and Scott, 2011)

When fully extended, the patella is out of contact with the trochlea groove. As can be observed in Figure 2.6, depending on the patellar tendon length, the patella gets drawn into the trochlea and gains contact with the femur between 10° to 20° (Elahi et al., 2000). The contact starts with the inferior margin of the patella, and then moves proximally as the flexion angle increases, as shown in Figure 2.7.

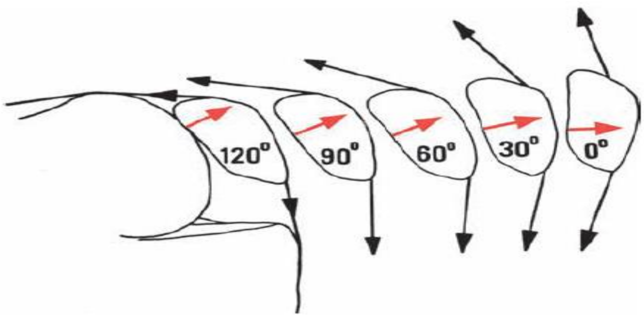


Figure 2.7 Contact Point Movements of the Patella during Knee Flexion (Walker, 2006)

For flexion angles over 30°, the patella settles into the deepening trochlea groove where it is stabilised by the quadriceps and patellar tendon forces. Niu et al., (2005) explains that the PF joint contact area extends from the medial margin of the medial facet to the lateral margin of the lateral facet, with the band of contact moving from distal to proximal. As Figure 2.7 illustrates, as the flexion angle moves from 30^0 to 60^0, the contact is at the centre. At a flexion angle of 90^0, the contact moves towards the superior pole, and at flexion angles greater than 90^0, the patella sits across the medial and the lateral condyles, forming two separate contact areas.

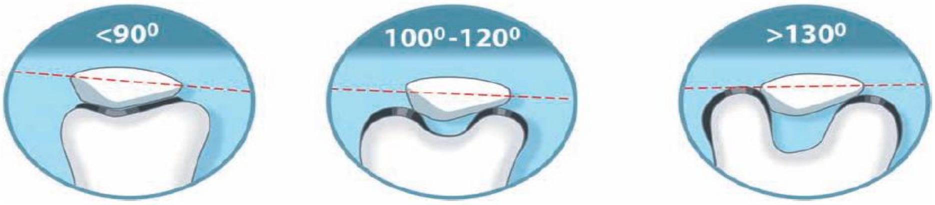


Figure 2.8 Movement of PF Contact Area beyond 100° of knee Flexion (Schindler and Scott, 2011)

According to Sharma et al., (2003), the patella can rotate 12° to 15° with respect to the femur, with most of the rotation occurring at more than 50^0 of knee flexion. Moreover, the patella tilts in a mediolateral direction, depending on knee flexion and the amount of internal and external rotation, and the varus / valgus alignment of the TF joint. Hinman et al., (2003) point out that the patella can be medially displaced by 5 mm in the coronal plane, with most of this displacement occurring during the first 30^0 of the knee flexion. Powers et al., (2004), state that the length of the patella tendon, and the angle between patellar tendon and quadriceps tendon, determines the load bearing area of the patella. According to Dixon (2006), the size of the contact areas of the PF joint is also dependent on the position of the knee. As can be observed in Figure 2.8, as the angle of flexion moves from 20^0 to 60^0, the average contact area increases linearly from 150 mm^2 to 480 mm^2, and it remains constant up to 90^0 of flexion after which it linearly reduces. At 120^0, the contact area drops to 360 mm^2. Table 2.2 sets out the compressive force incidents on the patella during various daily activities.

Table 2.2 Compressive Loads on PF Joint (Masouros et al., 2010)

Activity	Force	% Body weight
Walking	**850 N**	**1/2 x BW**
Bike	**850 N**	**1/2 x BW**
Stair ascend	**1500 N**	**3.3 x BW**
Stair descend	**4000 N**	**5 x BW**
Jogging	**5000 N**	**7 x BW**
Squatting	**5000 N**	**7 x BW**

From Table 2.2 it can be seen that the amount of compressive force on the patella changes dynamically according to different activities. During normal walking, when the knee flexion is lowest, the largest force of reaction on the PF joint is half of the body weight. This can increase to three times the body weight during stair climbing, and to seven times the body weight during activities such as running and squatting.

From the above analysis, it may be inferred that the reaction force at the PF joint is a measure of the compressive force exerted by the patella on the femur. During weight bearing activities, the sum of the force of the quadriceps muscles and patellar ligament, along with an increase in knee flexion angle, increases the PF joint's reaction forces. This reaction force can be more than three times the body weight while climbing stairs, and up to seven times the body weight during squatting activities. The alignment and motion of the patella within the femoral trochlea determines the distribution of the PF joint's reaction forces. The implication is that malalignment of the PF joint will lead to an increase in contact pressure on individual facets of the joint, and this increased contact pressure leads to cartilage deformation and the onset of OA.

2.7.2 Alterations in the Kinematics & Kinetics of the PF Joint leading to PF Joint OA

PF joint malalignment primarily manifests in the form of a lateral tilt of the patella or lateral displacement of the patella, or a combination of both, as indicated in Figure 2.9.

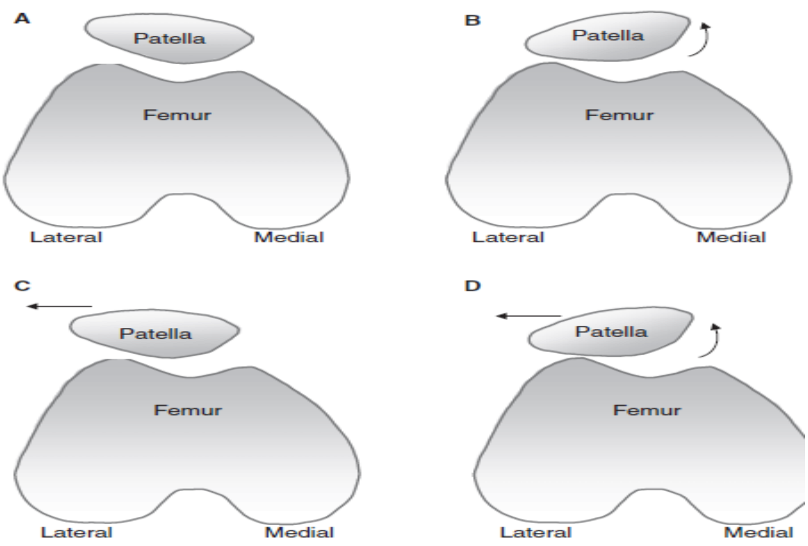


Figure 2.9 Patella Malalignment

In Figure 2.9, (A) shows the normal or ideal relationship between the femoral trochlea and the patella from an axial point of view. (B) illustrates an increase in lateral tilt where the lateral facet of the patella is tilted towards the lateral femoral trochlea. (C) shows the increased lateral displacement where the patella becomes laterally displaced, and (D) reveals a combination situation where the patella is both tilted as well as laterally displaced. Niu et al., (2005) point out that PF joints with laterally positioned patella and increased patella tilt show a higher incidence of PF joint OA when compared to normal knees. Hunter et al., (2005) indicate that the severity of OA induced knee pain is directly proportional to the amount of patella tilt; the severity of the disease increases as the patella becomes tilted, whether medially or laterally (Harrison et al., 1994). Medial tilt has been found to be associated with medial PF joint OA, while lateral tilt has been found to induce OA in the lateral compartment of the TF joint. Harrison et al., (1994) claim that up to 28% of varus knees show medial and lateral displacement, while 47% of valgus knees show lateral displacement. Iwano et al., (1990) found that patients suffering from PF joint OA alone demonstrated more patella lateral tilt compared to those with concurrent TJ joint OA. This study suggests that patella dislocation is therefore a predisposing factor for PF joint OA. According to Hinterwimmer et al., (2005), lateral patellar tilts are associated with a 50% reduction and lateralisation of the PF joint contact area. This increases lateral stress on the PF joint, and this stress may be a cause of PF joint OA onset.

According to Elahi et al., (2000), a valgus knee malalignment also results in an increased incidence of OA. Cahue et al., (2004) point out that as the frontal plane alignment determines the Q-angle, varus malalignment results in an increase in the Q-angle, and increased stress on the lateral patella facet. Cahue et al., (2004) further state that valgus malalignment precedes OA and that progressive valgus malalignment progressively increases the odds of isolated lateral PF joint OA. The aforementioned studies suggest that PF joint malalignment results in altered kinematics and altered kinetics on the PF joint, resulting in the onset of OA.

PF joint OA is accompanied by muscle weakness, particularly weakness of the quadriceps muscles. According to O'Reilly et al., (1998), the strength of the quadriceps muscle determines the severity of the pain and the physical functioning of patients with PF joint OA. Slemenda et al., (1998) and Thorstensson et al., (2004) point out that weakness of the quadriceps muscles precedes the onset of PF joint OA. Baker et al., (2004) also explain that there is a relationship between weakness of the quadriceps muscle and PF joint OA. Sharma et al., (2003) claim that the weaker the quadriceps muscle, the more likely it is that PF joint OA will progress in malaligned knees. According to Wluka et al., (2002), balanced activity in both the medial and lateral quadriceps is important to maintain PF joint alignment. Weakness of the quadriceps muscles results in alterations in medial and lateral quadriceps force, and in PF joint malalignment, which increases the chances of PF joint OA onset.

The literature suggests that the kinematics of other supporting structures of the PF joint can result in TF joint malalignment. The alignment of the patella at the local level depends on passive structures such as osseous configuration and soft tissue restraints, as well as active structures such as the medial and lateral quadriceps (Grelsamer, 2000). Powers (2000) points out that a shallow femoral trochlea groove and patella alta also impact on the alignment and motion of the patella. Farahmand et al., (1998) explain that tension within soft tissues, the medial and lateral retinaculae, distal expansions of the iliotibial bands, joint capsules and ligaments, all maintain the alignment of the patella. According to Heegard et al., (2001), the alignment of the lower limbs affects patellar tracking by altering the relative position of the femoral trochlea, and changing the tension in soft tissues. Lee et al., (2003) point out that femoral internal rotation is associated with an increase in lateral tilt and rotation of the patella, with higher lateral PF joint pressures being exerted.

Hinman (2005) explains that those quadriceps muscles that are most necessary for maintaining the optimal alignment of the patella are the vastus medialis obliquus (VMO); the distal medial quadriceps, and the vastus lateralis (VL). Dixon (2006) points out that healthy individuals exhibit synchronous VMO and VL activity during a variety of activities. Conversely, Sakai (2000) indicates that reductions in VMO activity or increases in VL activity result in malalignment of the lateral patella, and greater lateral PF joint contact pressures. Huberti and Hayes (1984) explored the impact of alterations in the Q angle on PF joint OA. The Q angle is formed by the intersection of the line of application of quadriceps muscle force with the centre line of the patellar tendon. It indicates the orientation of the resultant force of the four components of the quadriceps muscles that act on the patella on the frontal plane. Hehne (1990) describes how this laterally directed force vector causes the lateral facet of the patella to receive 60% more force than the medial facet. According to Mizuno et al., (2001), an increase in the Q-angle results in a shift in the PF joint contact area laterally, and this in turn, further increases the pressure inside the lateral facet. Hurwitz et al., (2002) point out that increases in lateral forces, or lateral patella malalignment, significantly affect contact pressures on the lateral facets, which in turn results in increases in PF joint OA.

According to Boling et al., (2009), increases in femoral internal rotation can lead to lateral patellar malalignment, and the subsequent patellofemoral contact stresses could increase due to smaller angles of knee flexion. This is illustrated in Figure 2.10 below.

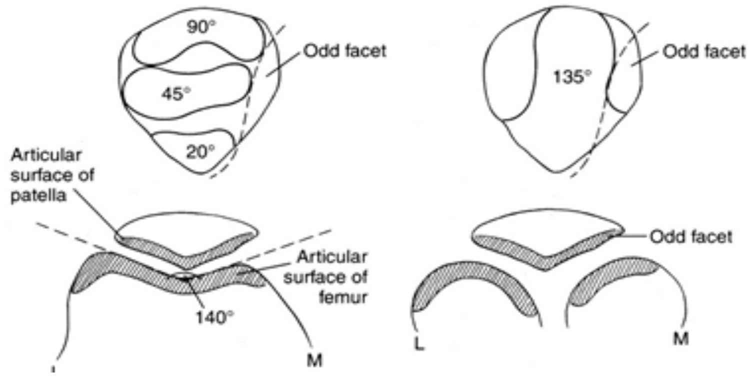


Figure 2.10 PF Contact Area Size during Knee Flexion Angle (Boling et al., 2009)

According to Boling et al., (2009), with smaller knee flexion angles at the TF joint, there is a decrease in the contact area across the PF joint. This then leads to increased contact stress, which in turn could result in decreased knee extensor moments, causing reduced dynamic control of the patella. This study confirms the research by Schnmitz et al., (2007) and that of Yu and Garrett (2006), which also reports that decreased knee flexion angles at the TF joint increase the ground reaction forces, and the consequent stresses on the facets of the TF joint.

From the above analysis, it may be inferred that altered kinematics and kinetics, as a result of PF joint malalignment and decreased strength of the knee musculature, often due to compensations by individuals with PF joint OA, are risk factors for the onset of PF joint OA. PF joint malalignment leads to an increase in the loading of the PF joint. Similarly, weakness of the hip and thigh musculature alters patella alignment within the femoral trochlea, which in turn leads to abnormalities in patellar loading. It is necessary to examine whether these kinematic and kinetic changes correspond to those that occur in ACLR knees, and if ACLR can therefore be considered as one of the key causes of PF joint OA. Therefore, the risk factors that predispose the PF joint to the onset of patellofemoral pain syndrome (PFPS) need to be examined first.

2.8 PFPS – Onset and Risk Factors

This section will examine the reasons for the onset of, and the risk factors for, PFPS. The reason for this is to examine whether biomechanical changes in ACLR knees present a critical risk factor for the onset of PFPS, and whether the risk factors are due to the biomechanical changes in ACLR knees.

2.8.1 PFPS and its impact on Athletes

According to Fulkerson (2002), PFPS manifests in the form of an unpleasant and steadily worsening pain occurring in the knees. It has been variously termed as runners / jumpers knee, patellar subluxation, patellar arthralgia, intra-articular patellar chondropathy and chondromalacia patellae (Witvrouw et al. 2005). Fulkerson (2002) points out that PFPS occurs even after ACLR. Anterior cruciate ligament reconstruction restores the knee's stability but does not prevent the development of PFPS. The studies by Brushoj et al., (2008) and Lankhorst et al., (2012) indicate that participants in their control groups experienced pain when they squatted and ascended or descended stairs. These are activities where there is considerable sagittal plane loading of the knees repeatedly by strong forces, and hence PFPS is also termed an overuse or overload injury because it occurs due to the constant loading of the knee due to strong forces. PFPS has a significant impact on the lifestyles and careers of athletes and sportspersons. Starkey (2000), states that PFPS pre-empts athletic activity, causing athletes to limit or stop their sporting activities altogether. Blond and Hansen (1998) point out that 74% of all athletes who suffer from PFPS alter their level of participation in sports by stopping participation altogether, playing at lower intensity levels, or taking a break from sports. In any case, such persons are not subsequently able to undertake physically demanding jobs and are hence compelled to seek alternative employment. Utting et al., (2005) point out that the PF joint is more likely to demonstrate symptomatic knee OA than the TF joint, and that young athletes who suffer from PFPS also eventually develop PF joint OA.

Fairbank et al., (1984) describe how PFPS includes those disorders characterised by pain and tenderness around the PF joint. PFPS is devastating to the careers of athletes, as it limits their physical activity levels. In a study conducted by Stathopulu and Baildam (2003), it was found that the majority of patients suffer from PF joint pain, and this limits the extent of their physical activity. Utting et al., (2005) claim that PFPS is associated with the development of PFOA, as their study reveals that 22% of the patients who exhibited PF joint OA also exhibited PFPS symptoms when they were adolescents. Yu et al., (2005), state that because of the impact of PFPS on physical activity and its association with PF joint OA, it is a public health concern, and preventive measures must be taken to decrease its occurrence. A decrease in the occurrence of PFPS will in turn decrease the rate of occurrence of PFPS (Yu et al., 2005). Therefore, there is a need to understand the risk factors associated with PFPS.

The implication is that PFPS can negatively affect the quality of life of athletes, as well as their participation in sports in the short term, and their employment prospects in the long term. Hence, there is a need to examine the causes of PFPS, and whether these causes ultimately lead to PF joint OA, in order to develop pre-emptive measures and treatments.

2.8.2 Risk factors for PFPS

There are several risk factors for PFPS, which are described below:

2.8.2.1 Patellofemoral Joint Injury

Injury to the knee has been quoted as being the singular most important reason for the onset of PFPS. This inference is supported by Brushoj et al., (2008) who indicate that PFPS originates from the portion of the knee that is composed of the central region of the patella, the distal patella pole and the peripatellar region. Powers (2003) and MacIntyre et al., (2006) explain that when injured knees are flexed or extended, maltracking of the patella on a stable femur results in PFPS. According to Barton et al., (2012), patella maltracking results in high levels of PF joint contact pressures, and this ultimately results in PFPS. However, the positioning of the tibia or the femur in relation to the patella also influences PF joint contact forces, and this can result in PFPS.

The link between PF joint contact pressures and decreased PF joint contact areas due to external rotation of the tibia, internal rotation of the hop and hip adduction, and the onset of PFPS, have been investigated by Salsich and Perman (2007), Souza et al., (2010) and Lee et al., (2003). It has been found that such patella maltracking and changes in femoral / tibial positioning result in changes in the patella's position, which in turn increases PF joint contact pressures and lowers load-bearing surfaces for the patella. According to Dye et al., (1999), excess PF joint stress in injured knees, and altered distribution of forces, results from the abnormal motion of the tibia, femur, and patella, and this decreases the load-bearing surface of the patella. Continuous overload of the PF joint results in the loss of peripatellar tissue, leading to pain. Farrokhi et al., (2011) point out that due to the reduced PF joint contact area, those suffering from PFPS demonstrate higher levels of PF joint stress during squatting and walking activities.

According to Salsich and Perman (2007), changes in the PF joint contact areas result in articular cartilage damage, but they point out that damaged articular cartilage is not a source of PFPS as it contains no nerve endings. Biedert et al., (1992) instead attribute the onset of PFPS due to articular cartilage damage to excessive loading on the subchondral bone. Farrokhi et al, (2011) points out that degeneration of patella cartilage manifests in all patients suffering from PFPS. Ho et al., (2014) state that female athletes suffering from PFPS were found to have higher water content in their patella structures, venous engorgement, and higher levels of extracellular fluids, all of which contribute towards more intraosseous pressure and pain.

Surgery of the ACL following a rupture can be followed by PFPS even though such symptoms did not occur before the surgery. Pinczewski et al., (2007) examined patients 10 years after they underwent ACLR surgery and found that 40% of individuals reported that during physical activities they experienced anterior knee pain. This suggests that with a BPTB autograft and after ACLR, it is common for patients to experience patellofemoral pain. Shino et al. (1993) report that after ACLR, patients often experience serious complications that manifest as patellofemoral pain. This pain is not thought to be due to PF joint degeneration, but rather to removing the central third of the patella tendon and bone associated with it, which contributes towards graft site morbidity. Patellofemoral pain is also thought to be strongly associated with loss of knee extension or knee flexion contracture, as this increases PF joint contact forces (Sachs et al. 1989).

From the above analysis, it may be inferred that injury to the knee resulting in an ACL rupture has been found to be associated with higher loading of the patella; reduced contact surface area of the patella; high incidence of cartilage wear and tear, and abnormal loading. It is this that has been found to be the cause of PFPS symptoms, and which ultimately leads to a decrease in the levels of activity in patients. A comparison of these findings with those in sections 2.7.1 and 2.7.2 indicates that they correspond to the consequences of alterations in the kinematics and kinetics of the PF joint that lead to PF joint OA. It may therefore be hypothesised that injury to the knee leading to ACL rupture results in altered kinematics and kinetics of the PF joint, and this in turn is manifested in the form of PFPS. This is true even of ACLR knees after reconstruction. PFPS is thus the precursor of PF joint OA.

2.8.2.2 Predisposing Structures

According to researchers such as Keller and Levine (2007) and Natri et al. (1998), there are structural causes to the pathogenesis of PFPS. These researchers point out that PFPS is associated with joint immobilisation due to frequent haemorrhaging, protracted synovitis, anomalies of the patella, patella misalignment, and knee dysfunction, on account of higher than normal body mass resulting in extensor mechanisms. This is due to structural misalignments occurring as a result of heredity, increased retinacular, subchondral and cartilage stress, overuse, instability, surgery / injury of the knee ligament, and acute trauma.

Patellofemoral pain may develop from anomalies of the knee structure. The patella acts as a fulcrum, resulting in static and dynamic stabilisation of the human body. The patella is not connected to the trochlear groove when fully extended, and hence to maintain stability requires support from the soft tissues of the knee located at the lateral aspect. Flexing of the knee requires action from various forces: quadriceps force, patellar tendon force and contact force, which together pull the patella posteriorly, providing stability. The contact force increases with the patella tension as the knee flexes. How deep the trochlear groove is has been shown to be unimportant when compared to the factor of stability (Amis 2007). The implication here is that there must be perfect alignment of the patella and the trochlear groove during flexing movements of the knee, and that malalignment will result in instability.

According to Keller and Levine (2007), malalignment of the patella and trochlear groove occurs due to abnormal sulcus angle, lateral patellofemoral angle, congruence angle and patellofemoral index. Where the patella is located too high, malalignment conditions such as patella alta occur (Malek & Mangine 1981). The sulcus angle is defined as the angle posterior to the patella, and is found at the articular facets between the slopes (Amis, 2007).

According to Powers (2000), the posterior condyle interval, at its midpoint, is overlaid by the intercondylar groove at its deepest level in a subject’s knee, and this is defined as normal; also, where the intercondylar sulcus is at its lowest location, and femoral condyles at the highest lateral points and highest medial points form the angle. The depth of the trochlear groove and the steepness of the slopes have a combined effect on stability (Amis 2007). The displacement of the patella along the medial and lateral planes can be analysed using bisect offset. This method of measurement involves using a line to connect the back femoral condyle with another perpendicular line drawn anteriorly from the tip of the trochlear groove. According to Powers et al., (2000), side alignment of the patella when compared to a perpendicular line is indicated as a percentage of the total width of the patella. Keller and Levine (2007) state that a congruence angle ranging from – 6 degrees to +6 degrees can also be used to compare the lateral alignment of the patella with the perpendicular line. Powers et al., (2000) state that as the sulcus angle increases, the displacement of the patella also increases. The tilt between the middle and the side patella can be measured in terms of side patellofemoral angle. This angle is formed between lines drawn over the apex of the condyles and a line drawn through the side facet of the patella (Keller & Levine 2007). Powers et al., (2000) suggest that individuals suffering from patellofemoral pain exhibit more patella tilt angle when flexing their knees at 45, 36, 27, 18, 9 and 0 degrees (10.7º vs 5.5 º) compared to those without patellofemoral pain; they used a 2 x 6 (groups x angle) analysis of variances for repeated measure on angle ($p = 0.02$).

The inference that can be made is that structural misalignments in the patella region alter the kinematics of the PF joint, and this alteration in kinematics in turn results in PFPS. This supports the suggestion that PFPS is the precursor of PF joint OA.

2.8.2.3 Valgus and Varus Orientation

Medial knee displacement is also known as knee valgus (abduction), and lateral knee displacement is referred to as knee varus (adduction). The knee is twisted outward in knee abduction, and the opposite in knee adduction. Greater adduction alignment may contribute towards increased tibiofemoral joint pressure in the medial compartment (Mizuno et al. 2001). The research suggests that females commonly present knee abduction, where there is adduction of the femur and adduction of the tibia. This causes the knee abduction angle to be misaligned, and the external abduction moment to increase. This is because a wider pelvis has been shown to move the centre of mass of the body to the hip joint at the medial point. In addition, Powers (2003) points out that when subjects are involved in dynamic tasks, weakness in the hip abductor could contribute towards excessive femoral adduction.

According to Stefanyshyn et al. (2006), PFPS and excessive knee abduction moment are linked, as after six months of running tasks, patients that developed PFPS were observed to have greater knee abduction impulse in baseline measurements when compared to a similar age group that did not develop PFPS. A further study investigated young adults with PFPS and a healthy control group of similar ages to compare the frontal plane kinetic patterns of the knee during a 10m walking task at a self-selected speed on a level surface (Paoloni et al. 2010). Three-dimensional (3D) kinetic analysis was used in this study, which reports that during the loading of the stance leg, patients with PFPS showed significantly greater knee abduction moments than those in the control group. Research by Stefanyshyn et al. (2006) and Myer et al. (2010) indicates that during landing and running tasks, the presence of high knee abduction loads can predict PFPS. During knee abduction postures, Noehren et al. (2012a) found increased lateral patellar displacement, which could be related to PFPS. In addition, the possibility of lateral PF joint OA could be increased by greater PF joint stress as a result of knee abduction alignment (Shultz et al. 2010).

From the above information, it may be inferred that dynamic hip adduction can lead to increased knee external abduction moment. This causes the individual to adopt compensatory strategies, demonstrating increased knee external adduction moment, which adds stress to the medial side of the knee and lateral ligaments. These factors contribute towards joint diseases, such as patellofemoral pain and osteoarthritis. These inferences match with the findings of Elahi et al., (2000) who point out that valgus knee malalignment also results in an increased incidence of PFPS leading on to PF joint OA. Cahue et al., (2004) explain that as frontal plane alignment determines the Q-angle, varus malalignment results in an increase in the Q-angle, and this places increased stress on the lateral patella facet, resulting in PFPS. In other words, valgus malalignment results in altered kinematics. This is manifested in altered biomechanics of movement, which exert abnormal kinetic forces on the PF joint, leading to the onset of PFPS and ultimately PF joint OA.

2.8.3 PFPS and PF joint OA

The analysis in section 2.8.2 indicates the risk factors for PFPS and PF joint OA. These risk factors are a result of altered kinematics, kinetics. These alterations match those induced by ACLR, as discussed in sections 2.7 and 2.8. It may be inferred that ACLR induces changes in kinematics and kinetics of the PF joint are the cause of PF joint OA. This inference is supported by Altman et al., (1986) who state that PF joint OA of the knee can be determined in terms of PFPS and radiographic abnormalities. Of these, PFPS is the most important symptom of PF joint OA. Nevertheless, in the literature, it is shown that PF joint OA is not afforded the same importance as TF joint OA. As the knee consists of three distinct compartments (medial and lateral tibiofemoral joints, and patellofemoral joint), and radiographic knee OA can present in various distributions across these compartments. Studies of non-traumatic knee OA have focussed on the tibiofemoral compartment alone due to radiographic assessment focussing on the posteroanterior view, yet as lateral and skyline X-rays have become more common in clinical practice and research, a greater awareness of patellofemoral OA has emerged. This has led to more recent population based studies reporting that patellofemoral joint pathology is common (Szebenyi et al., 2006), and is present in approximately 65% of people with symptomatic knee OA (Duncan et al., 2006). Most importantly, the patellofemoral joint is more likely than the tibiofemoral joint to result in knee OA symptoms (Duncan et al., 2008). Even isolated mild radiographic patellofemoral OA can cause symptoms that impact considerably on activities of daily living (Duncan et al., 2008, 2009). However, surprisingly little is known about patellofemoral OA, particularly the factors that contribute towards its development and, most importantly, how to manage this common and potentially debilitating condition effectively.

Similar to community-based studies, investigations following ACLR have typically evaluated rates of tibiofemoral OA alone, or have not described the compartmental distribution of OA. This is in spite of the extensor mechanism being surgically disrupted through a BPTB harvest, causing patellar malalignment (Van de Velde et al., 2008). Indeed, typical biomechanical features observed during the first two years following ACLR, such as increased tibial external rotation angle (Scanlan et al., 2010; Webster and Feller, 2011) and knee abduction angle (Webster and Feller, 2011), have been experimentally associated in vitro with altered patellofemoral kinematics and kinetics. These include increased lateral patellar tilt and rotation (Heegaard et al., 2001); a lateral shift in patellofemoral contact area (Huberti and Hayes, 1984; Mizuno et al., 2001), and increased lateral patellofemoral pressure (Lee et al., 2003; Mizuno et al., 2001). These altered loading patterns may give rise to patellofemoral joint degeneration. Considering that patellofemoral OA is capable of being a potent source of knee symptoms and affects the ability to perform activities of daily living, insights into the prevalence of patellofemoral OA after ACLR is warranted. Knowledge of factors that may be associated with this condition could facilitate more efficacious rehabilitation regimens, and thus significantly reduce the morbidity associated with ACLR if patellofemoral OA is implicated.

Powers et al., (2003) point out that some of the risk factors that can progress PFPS are changes in the kinematics and kinetics of the lower limbs during functional tasks, and reduced strength in the hip and knee. Because of the changes in the kinematics and kinetics of the lower limbs, there can be higher loads placed on the PF joint, and this could trigger PFPS. Lee et al., (1994) propose that weakness of the muscles of the hip and the thigh can change patella alignment within the PF joint, and this can lead to alterations in PF tracking. However, it may be noted that these factors have only been proposed, but are not unequivocally linked, to the onset of PFPS. Researchers such as Duffey et al., (2000), Fairbank et al., (1984) and Messier et al., (1991) propose additional risk factors for PFPS, which are higher rates of foot pronation and Q angle. However, these are again just proposals, and have not been proven to be linked to PFPS onset. Boling et al., (2009) investigated the biomechanical risk factors for PFPS, as well as the kinematics and kinetics of the lower limbs; Q-angle increases; strength of the muscles of the knee and hip; hip extensors; external / internal rotators, and the flexors and extensors of the knee. All of these potential risk factors were examined during jump – landing tasks and the association with PFPS was explored. The studies were based on the assumption that those suffering from PFPS also exhibit weak hip abduction and lower knee flexion / extensor strength. Moreover, those with PFPS have lower vertical reaction to ground forces moments of the extended knee, and those of external hip rotation. They also showed higher navicular drop during baseline assessments compared to those without PFPS. The study by Boling et al, (2009) proves that a reduction in knee flexion angle and reduced peak vertical ground reaction forces at the time of jump landing are indeed potential risk factors for the onset of PFPS. When related to strength, a decrease in knee flexing and extensor strength, as well as higher external rotation strength in the hip, are all risk factors for the onset of PFPS. It was also found that higher navicular drop is another significant risk factor for PFPS. It was therefore concluded that there is indeed a strong link between the onset of PFPS and changes in kinematics / kinetics, higher Q angle, navicular drop and reduced strength in the lower limbs. These findings corroborate those of Earl et al., (2005), Powers (2003) and Thomee et al., (1999) who discovered out that these biomechanical variables are indeed risk factors for PFPS. These include a decrease in the strength of the knee flexion and extension, higher navicular drop, a decrease in knee flexion angle, and higher rate of internal hip rotation angle.

Studies by Crossley et al., (2004), Nadeau et al., (1997) and Powers et al., (1999) found that individuals suffering from PFPS also show reduced angles of knee flexion during functional task performance. This was assumed to be a strategy of compensation to reduce the amount of loading on the patella, and in order to reduce pain. However, in the experiment by Boling et al., (2009), it was found that a decrease in the flexion angle of the knee was a risk factor for PFPS onset and not a compensation for PFPS. It was therefore proposed that those persons who develop PFPS also exhibit malalignment of the lateral patellar. This happens due to higher internal femoral rotation, which means the PF contact stresses may be higher for small angles of knee flexion because of the reduced contact area at lower knee flexion angles. Research by Schmitz et al., (2007) and Yu et al., (2005) also indicates that a lower angle of knee flexion during dynamic task performance results in higher vertical ground reaction forces. However, the experiment by Boling et al., (2009) found the opposite. During the assessment of peak vertical ground reach forces and the peak knee flexion angle, during the stance part of a jump landing activity, it was found that the reaction force for the peak vertical ground happens much earlier than the peak angle of knee flexion. Boling et al., (2009) evaluated the flexion of the knee and displacement of the hip during the stance stage to examine whether those who had PFPS exhibited a reduced amount of displacement, as this would have meant that they may not be able to absorb all the vertical ground reaction forces. However, there were no significant differences in the knee flexion angle and the displacement of the hip flexion between injured and healthy groups. It was found that those with PFPS also exhibited a much lower angle of knee flexion at initial contact in contrast to those who did not have PFPS. Based on this finding, it was proposed that less vertical ground reaction force is also a risk factor for PFPS. The overall inference that may be made from the research of Boling et al., (2009) is that a strong case exists to suggest that altered biomechanical changes after ACLR do lead to the onset of PFPS.

2.9 TF joint OA in ALCD and ALCR Knees

Research has been conducted by Zhang et al., (2003), Tashman (2004), Deneweth et al., (2010) and Gao et al., (2012) on the frontal plane motion analysis of knees in both ACLD and ACLR patients. The activities used for the studies include walking, running, hopping, and ascending and descending stairs. All of these studies have indicated higher levels of tibial adduction when compared to the knees of a healthy control group. Gao and Zheng (2010) studied the frontal movements of ACLD and ACLR knees and pointed out that the adduction / abduction movements do not correspond to those of healthy knees. Kozanek et al., (2011) stated that the kinematics of ACLR and ACLD knees are more similar to each other compared to healthy knees.

According to Shabani (2015), frontal plane mechanics are significantly impacted in both ACLD and ACLR knees. Shelbourne and Koltz (2006) indicate that the hamstring and quadriceps muscles provide stability by counteracting external knee adduction during walking. The hamstring muscles act as antagonists to the ACL, thereby restricting anterior – posterior and rotational displacements. The implication here is that restoration of the hamstring and quadriceps muscles during ACLR is very important. In both ACLD and ACLR knees, alterations in the strength of the hamstring / quadriceps (H/Q) ratio affect the stability of the knee. Hiemstra et al., (2004) found lower H/Q strength ratio values in ACLR knees. Schipplein and Andriacchi (1991) explain the dangers of this by pointing out that knee injuries increase the looseness of passive tissues within the knee joints, hence the need for high strength quadriceps and hamstring muscles for joint stability. A lower H/Q ratio in both ACLD and ALCR knees results in increased compression forces on articular cartilage, initiating its degeneration. Palmieri-Smith and Thomas (2009) also found that unbalanced contraction of the quadriceps and hamstring muscles post ACLR limits the ability of the knee to resist valgus loads, resulting in extra loading on the medial knee joint. Butler et al., (2009) found higher levels of external knee adduction moment in individuals five years after ACLR compared to those in a control group.

Hooper et al., (2002) point out that the mechanics of the hip and the ankle influence loading at the knee during weight bearing activities. Chang et al., (2005), Hooper et al., (2002) and Kowalk et al., (1997) all state that loading at the knee is heavily influenced by the hip and the ankle during weight bearing activities, and that impairment of the knee in the sagittal plane is compensated for by the hip and ankle. Change et al., (2005) indicate that the muscles of the hip play an important role in providing stability along the frontal plane. Wilson et al., (2005) points out that impairment in the abduction moments of the hip increases pelvic drop during the swing phase of walking. This in turn alters the position of the centre of gravity away from the centre of the knee joint, increasing the external knee adduction moment and compression in the medial knee joint.

According to Butler et al., (2009), external knee adduction moments and internal tibial rotation are found to be much higher in ACLR knees compared to healthy knees. In this regard, Tashman et al., (2004) point out that ACLR results in alterations in gait mechanics, which in turn causes more stress to occur in areas of the knee joint that do not normally absorb such forces. If such loading is repeatedly performed, the cartilage around the joint begins to deteriorate, particularly on the unloaded parts of the tibiofemoral joint. They conclude that ALCR does not restore normal rotational knee kinematics even after dynamic loading, and that these abnormal motions would result in the long-term degeneration of the joint.

Andriacchi et al. (2006) studied the thickness of the cartilage in ACL injured and healthy knees in an attempt to understand the alterations, if any, in the ACL injured knee. A modelling approach was used to analyse the walking patterns of the participants, as well as the associated load on the cartilage. It was found that where an offset rotation occurred in the ACL knee, this corresponded to a much higher rate of loss of cartilage when compared to healthy knees with no variations in loading patterns. It was also found that the portions of the knee joint where the cartilage started thinning were mainly in the central part of the medial compartment, gradually moving to the medial boundary, which is where the defects in the thickness of the cartilage first occur (Andriacchi et al., 2006). Thus, it was the middle portion of the TF joint, which showed the highest amount of cartilage degeneration. Moreover, an area of secondary wear and tear on the cartilage was discovered in the central part of the side area of the femur. The higher rate of cartilage loss in the middle portion of the knee resulted in a shift towards varus knee alignment, and that was the case throughout the experiment. It was also found that the increasing rate of middle compartment OA occurs due to the morphologic differences in cartilage thickness, and the variations in congruity between the medial and lateral compartments. This enhanced congruity in the medial compartment could be one of the reasons why the medial compartment is more susceptible to higher rates of cartilage loss. Small changes in tibiofemoral alignment can result in a shift in contact location, compared to the less congruent lateral compartment, potentially in areas that are not well adapted to loading, thereby leading to cartilage breakdown.

Andriacchi et al., (2004) developed a framework that shows the in vivo pathomechanics of OA in the knee and the origins of OA at the knee. This framework is shown in Figure 2.11.

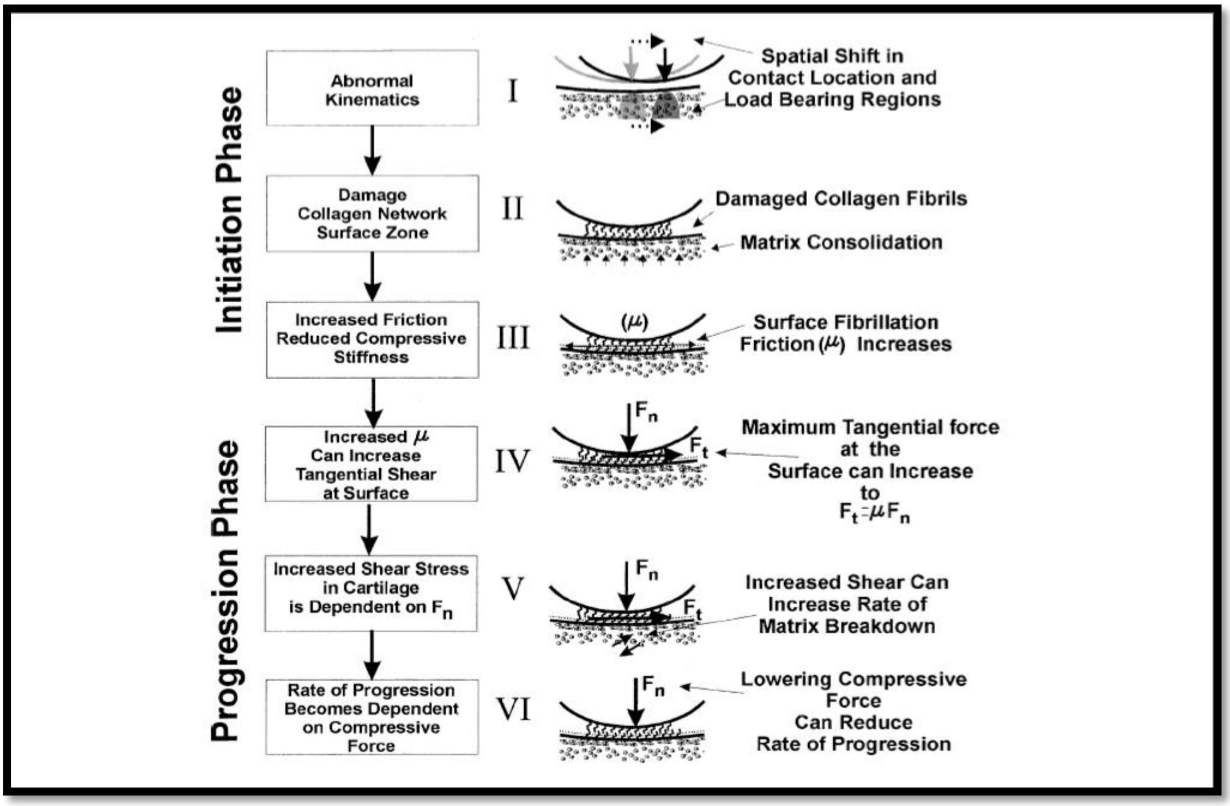


Figure 2-11 A Framework for the in vivo origins of OA at the Knee. This framework indicates how OA in the knee can potentially progress (Andriacchi et al., 2004)

From Figure 2.11 it can be seen that the framework includes biomarkers that indicate how cartilage degenerates, and how such degeneration is inherently linked to the kinematics and kinetics of the knee. Andriacchi et al., (2004) have broken down the framework to show the progression of OA at the knee according to two stages, that is, the start and progression stages. During the start phase, the healthy cartilage is subject to injury, or the healthy cartilage is exposed to alternative load bearing and stresses, which results in the cartilage becoming fibrillated. When the adaptive processes cannot cope with the biomechanical changes, the fibrillation of the cartilage is accelerated. Andriacchi et al., (2004) indicate that this would result in more friction, increasing the stress on the cartilage. During the second stage, or the progression stage, the already degenerated cartilage is further damaged by higher shear loading. As indicated in Figure 2.11, further compressive loading results in further progression of OA.

The framework proposed by Andriacchi et al., (2004) illustrates how OA can occur at the cellular level of the cartilage. However, the framework does not explore the relationships between any other factors that can potentially progress OA. An analysis of these factors is also necessary as this analysis, coupled with the framework in Figure 2.11, indicates how forces within ACL injured knees can be lowered, and how this would result in the enhanced long-term health of the knee, and better, more advanced rehabilitation and surgical procedures.

A summary of the research above indicates that in ACLD and ACLR knees there are higher levels of knee adduction, and that these adduction / abduction movements do not correspond to those of healthy knees. Frontal plane mechanics are significantly affected in both the ACLD and ACLR knee, as they exhibit a lower H/Q ratio in both types of knees, which results in increased compression forces on the articular cartilage, initiating its degeneration. In addition, impairments at the knee in the sagittal plane are compensated for by changes in the hip and ankle motions, which in turn differentially load the knee. It has also been found that alternations in knee adduction moments and internal tibial rotation are much higher in ACLD / ACLR knees compared to healthy knees. A comparison of these findings with those in sections 2.6.1 and 2.6.2 reveals that they correspond with the altered kinematics and kinetics associated with the onset of TF joint OA. A constantly higher adduction moment and increased internal rotation of the knee results in higher medial contact forces over time, resulting in degeneration of the cartilage and the development and progression of OA. The changes in biomechanical movements at the level of the hip result in hip adduction and internal rotation angles during loading, a condition that affects the kinematics of the knee. Such hip adduction and internal rotation angles, results in the centre of gravity of the knee joint moving medially relative to the feet (Powers, 2003). Because the feet are fixed to the ground, such inward knee movements in turn cause tibial abduction, which has been found to accelerate PF joint OA. The next section explores PF joint OA in ACLD and ACLR knees.

2.10 PFJOA in ALCD and ALCR Knees

Georgoulis et al., (2003) examined flexion – extension patterns in ACLD knees using gait analysis, and states that, as the main restraint in anteroposterior translation, the tearing of the ACL will cause knee instability. Because of this, ACLD patients use strong contractions of the hamstrings in order to pull the tibia from the back. Alternatively, they walk with lower quadriceps contraction in order to avoid pulling the tibia anteriorly. Chen et al., (2012) found that ACLD knees demonstrate higher flexion compared to normal knees during the stance phase. This means that patients use the strategy of higher flexion during gait. Beard et al.,

(1996) and Roberts et al., (1999) found higher flexing from the mid to stance phase. On the other hand, Fuentes et al., (2011) found that ACLD knees flex less during the mid and terminal stance phases. The same findings have been obtained with respect to ACLR knees. Knoll et al., (2004) examined ACLR knees and found that they flexed more during the stance phase and less during the swing phase. Devita et al., (1998) also examined the gait of ACLR patients and found lower knee joint flexing occurring six months after surgery, compared to three weeks from surgery. Delahunt (2012) also found much less keen joint flexion after ALCR four years post-surgery during studies conducted using diagonal jump landing, and during three single – legged, forward hop landing tasks.

One of the key phenomena associated with ACLR knees is weakness of the quadriceps. Drechsler et al., (2006) reported weak knee extensor musculature even three months after ACLR and extensive rehabilitation. Suter et al., (2001) and Lewek et al., (2002) observed that the quadriceps muscle of the injured limb was 80% weaker than the healthy limb. Petterson et al., (2008) define quadriceps weakness in terms of a decrease in voluntary isometric quadriceps knee extension moments, in comparison to the force created during the superimposition of electrical stimuli on maximum voluntary isometric contraction. Becker et al., (2004) point out that a reduction in voluntary quadriceps action occurs less in those individuals suffering from knee pain compared to those with healthy knees. According to Chmielewski et al., (2004) quadriceps weakness is associated with a change in gait pattern, with quadriceps-induced inhibition being observed in both the injured and the uninjured limbs of ACLR patients. Suter et al., (2001) state that knee extensor inhibition results in altered kinetic / kinematic variables during walking, and this results in premature degeneration of the articular cartilage. Cook et al., (1997) explain this by stating that reduced knee flexion angles result in degeneration of the articular cartilage, as it reduces the ability of the knee to absorb shock during loading activities. The ground reaction forces increase the impact at the TF joint, and consequently further stress the surfaces of the joint, leading to cartilage degeneration.

Lewek et al., (2002) and Bush-Joseph et al., (2001) also found that weakness of the quadriceps muscles is associated with lower angles of knee flexion and decreased internal knee extensor moments, even one year post ACLR. Hooper et. al., (2002) note the altered kinetics and kinematics that are exhibited in ALCR knees during activities, such as ascending stairs, with decreased knee flexion angles and reduced peak internal extensor moments observed; this also resulted in gait asymmetries in those with ACLR knees. Ernst et al., (2000) discovered reductions in knee extensor moments in patients with ACLR during functional activities such as single – leg vertical jumps and lateral step – up activities.

Berchuck et al., (1990) compared the gait analysis of patients with ACLD knees to those with normal knees during the activities of level walking, running, and ascending and descending stairs. Statistically significant differences were observed in the gait of case and control groups for all the activities. It was found that knee extensor moment was reduced the most during level walking, and the least during running. The reduced knee extensor moment was most likely due to the individual's effort to reduce, as much as possible, or avoid, contraction of the quadriceps. However, as reduced extensor moments reduce the quadriceps contraction, there needs to be a mechanical balance between the external moment and the weight of the limb for the knee to flex, which results in a change in gait that is referred to as the quadriceps avoidance gait. Chmielewski et al., (2001) found that athletes with ALCR and poor dynamic stability have alterations in movements in terms of reduced internal knee extensor moments during loading, and less knee flexion, which is not seen in those with good knee stability. Sagittal plane kinematics and kinetics during the stance phases of walking and running were analysed in subjects who had ACLR in comparison to those with uninjured knees. It was found that the ACLR subjects flexed their knees less than the uninjured subjects; they also displayed lower vertical ground reaction force during loading, and lower internal knee extensor moments during walking. During running, the ACLR knee extended more compared to the healthy knee, and the amount of knee flexion was less than in the uninvolved knee. The clinical significance of the study by Chmielewski et al., (2001) is that it shows that even though potential copers may develop some strategies to stabilise the consequences of ACLR knees, kinematic alterations in their gait suggests that they would benefit from training programs to dynamically increase the stability of their knees. One of the limitations of this study is the small sample of just 11 respondents, as a smaller sample size reduces the statistical relevance of the study. Another limitation of the study by Chmielewski et al., (2001) is the small number of cameras used to record the gait of the respondents. This means that the study was not powered, and there was a high chance of parallax and / or perspective errors occurring. Parallax error can result in an object, such as a leg, appearing to move even when it is stationary; while perspective errors can result in distortions in length and angle when the leg is observed from different positions. Such errors potentially increase the subjectivity of the results reported by Chmielewski et al., (2001).

However, their results are corroborated by Patel et al., (2003) who assessed ACLD knees during walking, running and ascending stairs. It was found that subjects had a significantly reduced peak internal extensor moment for all three activities when compared to the healthy knees of the control group participants. A reduction in knee extensor moments was also significantly correlated with a reduction in quadriceps muscle strength. Rudolph et al., (2001) examined the gait biomechanics of the ACLD knee amongst copers and non-copers during walking and running, and found that the non-copers were the ones who exhibited adaptations in gait, whereas the copers used motion patterns that were similar to those of the uninjured subjects. The non-copers exhibited reduced knee motion, knee extensor moments and lower quadriceps strength.

Bush-Joseph et al., (2001) examined dynamic knee function in patients who had undergone ACLR using the autologous patellar tendon technique. It was found that there were reductions in knee flexion angle and internal extensor moment between the ACLR and healthy control groups during activities of varying intensity, ranging from light walking, to moderate climbing and descending stairs, and the high demand activity of running. It was also found that there was decreased knee flexion angle and decreased knee extensor moment in the terminal stance, which is also related to reduced quadriceps and hamstring muscle strength.

Culvenor et al., (2014) investigated variations in transverse plane rotation between knees with varus and valgus alignment in the movements of respondents with and without PF joint OA after ACLR. It was found that in those individuals with PF joint OA and varus alignment, there was less knee internal rotation during walking and running. It was concluded that rotational shifts of this magnitude are enough to start, or even accelerate, the degeneration of the patellofemoral cartilage.

In section 2.7.1, it was observed that the most important function of the PF joint is to facilitate knee extension by improving the efficacy of the quadriceps muscle’s mechanical advantage by increasing the moment arm of the muscle force with respect to the centre of rotation of the knee. In addition, during weight bearing activities, the quadriceps muscles play an important role in distributing forces about the patella. The implication is that weaknesses in the quadriceps muscles in ALCR or in ACLD knees results in altered kinetics and differential loading, and an increase in contact pressures on individual facets of the joint.

This increased contact pressure leads to cartilage deformation and the onset of PF joint OA. Another key finding is that ACLD and ACLR results in reduced peak internal knee extensor moments. This is indicative of the reduced ability to dynamically absorb forces during functional movements, leading to alterations in PF joint loading patterns. The contact area of the patella that absorbs stress is reduced, resulting in higher loading at the joint. The reduction in the knee flexion angle occurs because of the decreased contact area, so there is an increase in contact stress, and this in turn results in decreased knee extensor moments, leading to reduced dynamic control of the patella. It may be inferred here that ACLD / ACLR results in altered kinematics and kinetics over time, with repetitive movements that predispose the PFPS to the onset of PF joint OA.

To evaluate the alteration in kinematics and kinetics after ACLR, functional performance tests are the most commonly used tools. Functional Performance tests (FPTs) simulate sporting activities and allow any functional problems with the knee to be evaluated and tested. FPTs measure issues related to the joint, the strength of muscles, flexibility, pain, and even confidence amongst patients, and they provide objective and measurable results (Barber et al., 1990). FPTs have been used extensively in recent years to evaluate athletic performance under conditions of injury to provide objective and measurable outcomes. FPTs mimic the joint loading forces that occur during sporting movements and assess the kinematics of functional movements (Lephart and Henry, 1995). Hence, FPTs can be used to examine alterations in the kinematics and kinetics of ACLR knees. It may also be inferred that since the variables measured by FPTs include knee movements and the strength of the muscles, they can also be used to study the onset of PFPS and PF joint OA.

2.11 Evaluation of Functional Performance Tests

The exercises involved in FPTs are closed chain and require the movement of the ankle, knee and hip joints to occur at the same time in order to control those segments of the body that are used during sporting activities (Lephart and Henry, 1995). FPTs allow for the quantified observation of lower limb functions in the absence of laboratory-based methods of 3D / force platform analysis. They require a minimum of space, time, cost and administrative complexity when compared to laboratory based measurements.

Tests for vertical jump and agility do not permit the injured and non–injured limbs to be compared (Barber et al., 1990). Hence, they will be eliminated for consideration in this research. However, the single limb tests of hopping, single leg vertical jumps, stair climbing, and the star excursion balance test (SEBT) allow for a comparison between injured and non–injured limbs (Goh and Boyle, 1997). Moreover, these FPTs can be used to compare the functionality of the injured limb after an ACL injury with the uninjured limb. However, Adams et al., (2012) point out that the requirement of stairs to perform the stairs hopple test is a clinical limitation, especially when there are larger numbers of respondents involved. Barber et al., (1990) points out that single leg vertical jump tests are not sensitive enough in detecting functional problems in injured limbs. The inference here is that the single leg vertical jump test cannot be used for the detection of functional problems in the lower limbs in injured populations. Hence, stair climbing and single leg vertical jump tests will be eliminated for consideration from this research.

According to Fitzgerald et al., (2000) hop for distance tests are used to test for functionality after an ACL injury. The advantage of hop tests is that they can detect differences in functionality between injured and non-injured limbs and between ACLR / ACLD and non-injured limbs (Barber et al., 1990; Goh & Boyle, 1997; Noyes et al., 1991; Petschnig et al., 1998, Eastlack et al., 1999; Rudolph et al., 2000). The SEBT test is similar to a single leg squat exercise, requiring strength, and greater muscular control and movements at the hip, knee and ankle (Olmsted et al., 2002; Robinson & Gribble, 2008). According to Aminaka and Gribble (2008), SEBT tests can detect functional deficits in patients, and can be used to examine knees after an ACL injury, as well as to predict the likelihood of future injury to the lower limbs. Both these tests may therefore be used in this research. However, researchers such as Sciascia and Kilber (2006), Stensrud et al., (2011), Padua et al., (2009), Overmoyer et al., (2012) and Onate et al., (2010) point out that hop tests and the SEBT use two-dimensional (2D) or three-dimensional (3D) video analysis and require multiple tasks, which makes them impractical for implementation within large group pre-participation examinations or in physicians' offices. Similarly, Ugalde (2015) points out that drop jump tests and the SEBT tests require multiple tasks, which makes them impractical for implementation with larger sets of respondents or in the office of a general physician. Moreover, although leg hop tests are more objective measurements compared to time postoperatively, they fail to address the multi-planar motion that is characteristic of cutting tasks (Cortes et al., 2013). Cutting involves frontal, sagittal, and transverse plane motion at the trunk, hip, knee, and ankle, whereas isokinetic tests and single leg hop tests generally only assess sagittal plane motion. Initiating cutting is a very important phase in rehabilitation.

For the purposes of this research, none of the motion and functional performance tests used above have been used in the context of ACLR / ACLD. This research therefore proposes to use the functional performance tests of running and squatting to examine the altered biomechanics of ACLR / ACLD knees, and how alterations result in PF joint OA.

2.11.1 Functional Performance Tests – Running

According to Karanikas et al., (2009), running is one of the most important FPTs as it places more demand on the knee, creates higher accelerations, and for every phase of running there is only the single leg transference of weight. It is an activity commonly performed by active people and athletes. In these individuals, a deficit in mechanics can cause excessive loading, or even indicate that rehabilitative processes are not complete, which would in fact increase the risk of re-injury or cause more trauma. The research by Rudolph et al., (2001) and Berchuck et al., (1990) all indicate that adaptations of the knee during running resemble the exaggerated adaptations of ACLR knees. This means that running can be used to examine the kinematics of ACLR knees. Nevertheless, the literature on the biomechanics of running with respect to ACLR respondents is scarce.

Running has been used as FPT in numerous studies, for example Rudolph et al., (2001), Takeda et al., (2014), Patel et al., (2003), Chielewski et al., (2001), Berchuk et al., (1990), Bush-Joseph (2001), Devita et al., (1998), Lewek et al., (2002), Sigward et al., (2015) all used running in their evaluation of functional performance for ACLD and ACLR knees. However, a critical shortcoming of all the aforementioned studies is that they did not investigate kinematics and kinetics before and after ACLR / ACLD during running. They also did not consider whether there was a similarity with the kinematics and kinetics of patellofemoral pain syndrome. However, the inference that can be made here is that running may be used as a clinically relevant tool to understand gait adaptations, kinematics and kinetics in ACLR and ACLD knees. Understanding how healthy people normally run is important, so that pathological running gait can be better understood, and this section provides background information, from a review of the literature, on normal running biomechanics, which should inform the interpretation of running by ACL injured subjects. Figure 2.12 shows that compared with walking, running patterns vary, but they still have phases in the gait cycle that are similar; for example, in comparison with walking, running involves increased stride rate and length (Luhtanen & Komi 1978).

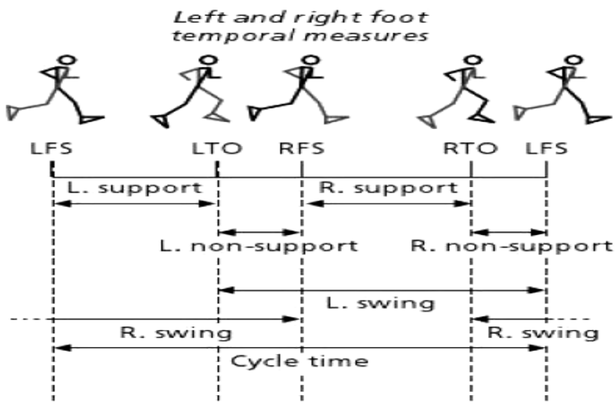


Figure 2.12 Human locomotion in running: LFS=left foot strike, LTO=left toe off, RFS=right foot strike, RTO=right toe off (Zatsiorsky 2000)

In running, there is decreased time spent in the support and cycle time, but when the foot makes contact with the surface, there is a short period of flexion, and then the hip movement quickly continues to extend (Nilsson et al. 1985). Also, there are two periods of flexion performed by the knee joint: in the support phase and in the swing phase, so that the first swing phase involves reduced leg moment inertia, as this assists the leg in swinging (Hamill & Knutzen 2009). At toe-off, hip peak extension occurs with increased peak hip flexion to help the leg in swing motion. Dicharry (2010) explains that there is approximately a 15º increase in abduction/adduction mobility, and around a 60º increase in hip flexion/extension mobility overall. During running, the knee is shown to produce 8º of valgus and 80º of flexion during the swing phase across the range of motion, and in the support phase the knee produces 8º varus angle, 11º external rotations, 8º internal rotations, and increased flexion angle up to 36º. The ankle remains in a neutral position during running, but before contact with a surface is made, it is dorsiflexed by 10º. The ankle produces pronation of between 8º and 15º and dorsiflexion of 50º during mid-stance, but the ankle changes at toe-off to 25º of plantar flexion from 50º of dorsiflexion (Hamill & Knutzen 2009). Vertical GRF has one clear peak for running. Vertical GRF spikes sharply during running to produce an impact peak at contact, then depending on the style of contact, it can decrease slightly, and it returns to an active peak of between 2.2 and 2.6 times the body weight for a normal distance runner (Dicharry 2010). Noehren et al., (2013) also used a treadmill to simulate running tasks in participants after ACLR. It was found that during running, the ACLR case group had higher initial vertical impact force and loading forces, and smaller knee extensor moments and hip angles. They displayed smaller knee moments and hip angle as well as increased extensor moments of the hip when compared to the control group participants.

A similar conclusion was arrived at by Nigg et al., (1994) following a kinematic comparison of overground and treadmill running. The goal of this research was to ascertain the suitability of treadmills for simulating overground locomotion, given the widely held assumption that locomotion on a treadmill is similar to that of overground running. The respondents were made to run alternatively on a treadmill and overground, and the kinematics of the right leg and foot were examined. Several differences in kinematics between motion on the treadmill and overground were found. It was found that most of the respondents systematically used a flatter position for running on the treadmill compared to overground running. Statistically significant differences were observed in the respondents' running styles, running speed, and the placement of their shoes on the ground and on the treadmill. The research concludes that assessing kinematics on a treadmill for lower limb assessments can lead to erroneous conclusions about overground running. It may be inferred, therefore, that use of a treadmill cannot mimic running overground, and that for the purposes of examining kinetics under conditions of running, actual overground running may be recommended.

2.11.1.1 Running After an ACL Injury

Assessing running performance in ACLD groups has been undertaken in a limited number of studies. Berchuk et al., (1990) used running as one of the FPTs, in addition to level walking and ascending stairs. Their study examined gait adaptations by patients with ACLD, with the control group composed of persons with normal, healthy knees. It was found that there was an increase in the moments that flexed the knees during running when compared to walking. It was possible to ascertain that the pattern of internal knee extensor moments of the PF joint in both groups was the same, although at mid-stance the peak extensor moments about the knees of the ACLD group was significantly smaller than that of the control group. The study also indicated gait adaptations performed by those in the case group that were not performed by those in the control group, nor indicated in the non-injured limbs of the case group. However, a weakness of this study is that the differences in the kinetics and gait adaptations between the case and control groups were not statistically significant. Moreover, the study used only a small sample of 16 respondents, which limits the statistical relevance of the results. The clinical relevance of the study by Berchuk et al., (1990) is that it indicates that when a person suffers from ACLD knees, even low stress activities, such as walking on flat surfaces, will be performed in an abnormal manner. This study concludes that an abnormal gait can have long-term implications related to changes occurring in the knees of individuals whose ACLs were ruptured but never reconstructed. However, Berchuk et al., (1990) did not go on to specify what those implications could be. In addition, the study by Berchuk et al., (1990) was based on gait analysis involving patients and a control group for walking, jogging, ascending stairs and descending stairs. Data was collected during the middle stride, with measurements starting just before the foot struck the force-plate. However, the only instrumentation that was used to observe the gait of 16 patients was two cameras. This seems to be rather inadequate for the observation and gathering of data on alterations in gait, and some inaccuracies in the data collection and analysis may have occurred due to this, therefore affecting the validity of the results.

Chmielewski et al., (2001) examined biomechanical motion in patients with ACLD during running. From the stance phase of running, sagittal plane kinematics and kinetics were examined. It was found that the knee flexion angle at initial contact was greater for those in the ACLD group compared to the control group, with less knee flexion angle than for the uninjured limb during running. The same study also revealed no differences in kinetics whilst running between the ACLD and control groups. However, this research is also limited by its small sample size of 11 respondents, which makes generalisation of the findings to larger groups of patients with ACLR difficult.

Patel et al., (2003) used gait analysis, including walking, running and stair climbing, to assess dynamic functionality in patients suffering from ACLD. It was found that during running, the internal knee extensor moment for those in the ACLD group was significantly lower than for the control group. It is also possible to ascertain that the decreasing strength of the quadriceps muscle significantly correlated with a reduction in the peak external extensor moment. A critical weakness of this research, however, is the ambiguity in the measuring instruments. More of the participants in the ACLD group were tested using the Kin-Com dynamometer, whereas the Cybex dynamometer was used more with the control group; therefore, the quadriceps muscle weakness in the ACLD group cannot be taken as being conclusive. Moreover, research conducted by other investigators, such as Kannus (1988), Lephart et al., (1992), McNair et al., (1989), have indicated a substantial deficit in the strength of the quadriceps muscles in persons with ACLD knees. However, the deficit in strength in this study was very significant, with $p = 0.001$. Hence, it may be inferred that the deficit in the strength of the quadriceps muscle amongst the ACLD groups was not the outcome of the testing protocol. It is also uncertain whether the similarity in the strength of the hamstring muscle of the two groups was on account of the testing protocol. This inference has been made, as in general the knee flexors of ACLD groups are weaker compared to those of healthy groups, even though this difference is not very significant.

In addition, it may be noted that Patel et al., (2003) recruited patients from the clinics of two surgeons who had diagnosed all the respondents as having ACLD knees. This indicates that the findings of Patel et al., (2003) are only applicable to symptomatic patients seeking care. Moreover, again, just two cameras were used to record the movements of 44 respondents. This could potentially have led to loss of data or inaccuracies in the data collection, which may have had an effect on the validity of the data. Because of the exclusion and inclusion criteria, the study group therefore represents those patients typically seen in an orthopaedic practice. In patients with effusion, the quadriceps muscle strength and its relationship to dynamic function may be different. Additional adaptations may be present in patients with effusion or more extensive meniscal injuries, which were not explored in this research.

Running performance in ACLD copers and non-copers versus control group participants has been assessed by Rudolph et al. (2001), with participants running at a self-selected speed. The participants were divided into three groups: copers, non-copers and adapters. Copers are individuals who have resumed normal sporting activities with no occurrences of the knee giving way. Non-copers are those who experience frequent episodes of giving way at the knee, including whilst performing activities of daily living. Adapters are individuals who have not faced episodes of giving way in activities of daily living, but have adapted their sporting activities due to experiencing knee instability when performing more difficult cutting and pivoting activities (Rudolph et al., 2001).

According to Rudolph et al. (2001), at peak knee flexion angle, there is a lower internal knee extensor moment presented by non-copers in the associated leg, but for both legs of those in the control group and ACLD copers, this was significantly lower. These findings support those of Patel et al. (2003) and Bush-Joseph et al. (2001) who report significantly lower internal knee extensor moments for those in the ACLD group compared to the control group (Patel et al. 2003). Berchuck et al. (1990) also claim that internal knee extensor moments were significantly reduced for the ACLD injured legs of subjects when comparisons were made with both legs of the control group and ACLD contralateral legs. If there are adaptations in other activities performed by the person with ACLD knees, it may be inferred that the internal extensor moment is being reduced in order to protect the ACL graft from being excessively loaded. However, it may be noted that this affects the ability of the person to resume normal levels of activity, and it has long-term implications for the overall stability of the knee. Overall, the methodological quality suffers from the disadvantages of a small sample size; lack of adequate assessor blinding; lack of randomised controlled trials (RCT's), and the inadequacies of clinical, subjective and functional tests for assessing ACLD copers and non-copers. Moreover, as in the case of Patel et al., (2003), the low numbers of cameras used to record the movements means that the chance of inadequate / inaccurate data collection would have been high, resulting in the statistical relevance of the results being limited.

Therefore, it has been shown that non-copers tend to shift knee loads to the hip from the knee, and limit their knee flexion, but when compared with those in a control group, ACLD copers tend to move with similar biomechanics. Rudolph et al. (2001) suggest that the transference from the knee to the hip mechanism during running could be due to hamstring co-contraction, as gait level indicates this. However, long-term knee problems present an increased risk for non-copers due to greater compression and shear forces from the reduced shock absorption capability of the knee joint as a result of reduced knee motion. Rudolph et al. (2001), also conclude that muscle activation is a factor of adaption in ACLD in terms of magnitude and timing, and other factors could play a key role in knee stabilisation, indicated by an insufficient relationship between internal knee extensor moment and quadriceps strength for ACLD copers, which for non-copers is significant.

Rudolph et al., (2001) claim that knee joint integrity in the long-term could be positive for copers' normal muscle timing and biomechanics; however, to date, there is insufficient evidence from other studies on ACL subjects' long-term knee health to support this claim. In addition, a weakness of the study by Rudolph et al., (2001) is that it does not clearly specify the relationship between quadriceps' strength and functional ability, or how the severity of the injury affects the pattern of reduced knee extensor moment in non-copers.

2.11.1.2 Running After ACLR

Karanikas et al. (2009) have reported a smaller range of motion at the knee during running on a treadmill for the injured versus contralateral limb, throughout a three to six month post ACL reconstruction period. This corresponds with a significantly lower knee flexion angle during the running stance phase. These deficiencies in the injured versus contralateral limb continued to be present in the period six to twelve months post injury. There were no differences evident in the kinematics between limbs at the end of a twelve-month period. Such studies provide evidence of functional adaptations in activities that are more demanding. In ACL injured groups, the discussion continues concerning whether normal knee biomechanics are good or bad for long-term knee health, and especially for those who return to activities that are more demanding.

The research conducted by Karanikas et al., (2009) highlights the importance of time in understanding injuries caused to the knees amongst ACL injured groups. This is because all the respondents in their group regained normal strength in the knee, as well as normal knee biomechanics, within a year following ACLR surgery. This changed motor strategy during running was time dependent, first reducing knee angles and ROM. Karanikas et al., (2009) report that there is a connection between muscle strength and knee angle, and ROM could be a strategy used by subjects to protect themselves from pain. This is because during running tasks, knee joint angles and ROM usually increase, which suggests that to regain normal knee function, the recovery period in the early stages relies on muscle strength, although more complex mechanisms than muscle strength alone are required for successful adaptation. An issue that has limited the results of this study is that the longitudinal analysis of muscle strength and the motor tasks for the same group at different postoperative stages is missing. Due to organisational and ethical reasons, and the duration of the examinations, it was not possible to persuade respondents to take part in repeat analysis; also, using a treadmill to assess ALCR knees has its limitations, as explained in section 2.11.1. The small sample size and the results not being powered, as well as using two genlocked video cameras, may have affected the significance of the results. This is because a smaller sample may lead to a result that is not sufficiently powered to detect a difference between the groups. In addition, the use of two cameras to record the movements means that the chance of inadequate, and even inaccurate, data collection would have been high; it would have reduced the size of the capture volume, which contributes towards the system resolution and the accuracy of the position of the data. Therefore, the reliability and validity of the results are questionable. In addition, the use of a treadmill for running is problematic, as a treadmill cannot mimic running over-ground. These factors could have influenced some of the results reported by Karanikas et al., (2009).

Downhill running was studied by Tashman et al. (2007), in 16 ACLR subjects of mixed gender, with patella (n=7) and hamstring reconstructions (n=9), during a five and 12 month post-surgery period. This study reports increased knee adduction and external rotation angles, and no measurement time changes for ACLR knees. In addition, flexion and extension angles showed no variation, with no significant differences found between limbs. These findings are in contrast to the findings of Karanikas et al. (2009) during a post-surgery period of between five to six months; although there were similar findings for flexion and extension angles after a one-year time period. Adduction or rotation angles at the knee were not investigated by Karanikas et al. (2009), which shows all knee motions need to be assessed and not just the sagittal plane, since in ACL injured groups, there may be differences in other planes of motion.

There is a gap in the research according to Tashman et al., (2007) concerning the clinical significance of differences in rotational motion between ACLR and normal knees. Several limitations were also noted, including the heterogeneity of the respondents regarding age, gender, surgeon, graft type (hamstring or patellar tendon), graft fixation, rate of rehabilitation, and level of athletic activity. It was also observed that there were no effects from age, graft type, meniscal injury, surgery and/or testing timing, or functional scoring for knee kinematics. Moreover, the sample size and statistical power were not sufficient for reliably detecting such relationships. Graft positioning was also not tightly controlled (although a coronal angle of 30° from vertical was targeted). These factors could have influenced some of the results reported by Tashman et al., (2007). There is also another limitation due to using a treadmill for ACLR knee analysis, as previously mentioned. Several limitations can be noted, including the heterogeneity of the subject population in regards to rate of rehabilitation and level of athletic activity. However, the sample size and statistical power were insufficient for detecting such relationships reliably. These factors could have influenced some of the results reported here. This is an ongoing study, and data on all of the confounding variables listed above have been collected (excluding graft tension, which was not measured, but including graft positioning as determined from the CT). As data from more subjects becomes available, it should be possible to investigate the influence of these confounding factors in a more robust manner. This analysis was limited to a single activity.

Bush-Joseph et al. (2001) studied the knee kinetics during running for ACLR individuals only, and found that the internal knee extensor moment of the ACLR group significantly decreased, more so during running tasks than in walking and gait tasks. This study reports that for peak internal knee extensor moments, there were no differences between the ACLR and control groups, but subjects with weak quadriceps presented an internal knee extensor moment that was reduced the most. Solely assessing gait may provide insufficient evidence for measuring whether kinematics and kinetics are regained following surgical interventions, because many ACLR subjects that have been studied are likely to take up sporting activities again, and possibly activities at higher levels; although the protective strategy of decreased internal knee extensor moment may help to reduce excessive loading. These factors influence long-term knee health and levels of performance in activities. In a study by Bush-Joseph et al., (2001), significant decreases were observed in the peak internal knee extensor moment during higher-demand activities, including running and running followed by a cut. When patients were running, the decrease in the peak knee extensor moment was correlated with strength. Those patients with relatively weaker quadriceps muscles displayed more dramatic reductions in the knee internal extensor moments than did those with normal strength. This data reinforces the value of ensuring complete quadriceps muscle rehabilitation after ACL reconstruction. The data also indicates that greater isokinetic quadriceps muscle strength is predictive of normal dynamic function during higher-demand activities. However, the same effect was not noted with run and cut manoeuvres for the ACL-reconstructed subjects.

Although the peak knee extensor moment was significantly reduced from normal, it did not correlate with quadriceps muscle strength (Bush-Joseph et al., 2001). Other factors exclusive of strength, including patient apprehension, decreased proprioception, or micro instability, may in part be the cause of the patients' decreased ability to generate a normal peak knee extensor moment. It may be inferred here that a running activity mimics most of the functional movements that an individual performs after ACLR, and hence may be used to examine those biomechanical changes associated with onset of PFPS and PF joint OA. A key shortcoming of this research is the small sample size of 22 respondents, which limits the statistical relevance of the results. In addition, the research makes no mention of any of the data collection instruments used, or how the observations were carried out. Also, whether cameras were used to examine changes in gait is not indicated. Therefore, the reliability and validity of the results is questionable.

It is the opinion of Boling et al., (2009) that a reduction in the knee flexion angle results in a decrease in the surface contact area of the patella. This in turn increases the contact stress forces at the patella, and decreases the knee extensor moments as well. In addition, Boling et al., (2009) point out that as the femoral internal rotation increases, there is higher malalignment of the lateral patella, resulting in an increase in patellofemoral contact stresses caused by small knee flexion angles. A small angle of knee flexion in the region of the TF joint reduces the contact area of the PF joint, and this in turn increases the contact stress area, resulting in a decrease in knee extensor moment, which reduces overall patella control. This study corroborates with the results of Schnmitz et al., (2007), and Yu and Garrett (2006), who also found that a reduction in knee flexion angle around the TF joint causes the ground reaction forces to stress the various surfaces of the TF joint even more.

Research conducted by Powers (2003) reveals that altered kinematics and kinetics lead to differential loading in the patellofemoral joint, which ultimately causes PFPS. Studies by Crossley et al., (2004), Nadeau et al., (1997) and Lee et al., (1994) all indicate that individuals exhibiting PFPS syndrome also show a reduction in knee flexion angles whilst performing functional tasks. Crossley et al., (2004) and Nadeau et al., (1997) point out that this is a compensatory strategy used to reduce the amount of stress and pressure on the patella, and so reduce pain. These studies have therefore established a decrease in the flexion angle of the knee as a potential risk factor that precedes PFPS. It may be inferred that those persons who develop PFPS also suffer from malalignment of the side patella. The patellofemoral contact stress may increase further at the smaller knee flexion angles if the individual ultimately develops PFPS and has lateral patellar malalignment. This is due to the increased femoral internal rotation as a result of the decreased contact area at these lower knee flexion angles.

During a dynamic task, individuals with low quadriceps strength may have decreased knee flexion angles, as such tasks require the quadriceps musculature to exert a great amount of eccentric force, causing difficulties for these individuals. While increased or decreased knee extension moments are not a predictor of the development of PFPS, the descriptive analysis reveals that individuals that developed PFPS had significantly lower knee extensor moments during the jump-landing task. This decrease in knee extensor moment, along with decreased quadriceps strength, may lead to the dynamic control of the patella also decreasing. In addition, an increase in hip internal rotation, perhaps due to the increased navicular drop, could lead to the patella being laterally aligned. Increased hip internal rotation angle, along with decreased knee flexion angle, is likely to increase the patellofemoral contact pressures, and with time, repetitive movements could lead to PFPS developing. Therefore, it is suggested that increased hip external rotation strength is caused by individuals continuously having to control the increased hip internal rotation angles whilst performing dynamic tasks. However, the discovery of the increased hip internal rotation angle, as well as the increased hip external rotation strength, suggests that this may be a neuromuscular control issue. This means that individuals are not aware of when to engage the hip external rotators during dynamic tasks, which causes the increased hip internal rotation angle.

These alterations match those induced by ACLR, and so it may be inferred that as ACLR induces changes in kinematics, kinetics and the biomechanical movement of the PF joint, it is the cause of PF joint OA. This inference is supported by Altman et al., (1986) who claim that PF joint OA of the knee can be determined in terms of PFPS and radiographic abnormalities. Of these, PFPS is the most important symptom of PFOA.

The above analyses suggest that running can be used as a functional performance test to better understand the kinematics, kinetics and biomechanics of ACLD and ACLR knees. However, in the studies described above, there are certain limitations, particularly as some studies have used a treadmill for running, and due to methodological errors. For example, an insufficient number of cameras used to record the movements may increase the likelihood of inadequate or inaccurate data collection due to the resultant reduction in the size of the capture volume, as this contributes towards the system resolution and the accuracy (or inaccuracy) of the position data. In addition, the small sample size in some studies limits the statistical relevance of the results, and may not adequately represent a typical cross section of the population under examination. The most of the studies 12 months or more post ACLR. In addition, no studies have measured the hip joint frontal and transverse kinematics and kinetics during running. PFPS as well as the onset of PF joint OA after ACLR can be evaluated through a running task, however, in general clinical settings, the option to actually ask the individual to run and assess this is limited; therefore a more space-optimised clinical assessment is needed. Thus, the single leg squat (SLS) has been chosen as the measure for assessing such individuals.

2.11.2 Functional Performance Tests – Squatting

According to Whatman et al (2011), the SLS is correlated to running, as they found strong correlations ($r \geq 0.70$) between running and SLS. Despite the small sample size and control velocity, this study demonstrates the potential of using small knee bending tasks when assessing dynamic lower-extremity alignment. The implication here is that SLS can be used to identify those persons who are at risk from PFPS. Whatman (2011), claims that SLS mimics most of the movements of running, and this is especially useful in a clinical setting, where it may not be possible to examine running conditions due to the paucity of space. This is another reason why SLS can be used to examine the biomechanical movements of ACLR and ACLD knees.

According to Clairborne et al., (2006), Willson et al., (2006) and Zeller et al., (2003), squatting tests can help to identify core strength related to landing, running and cutting tasks. As such, they mimic most of the movements that athletes have to perform during the course of their sporting activities. The single leg squat (SLS) has been used as a valid and reliable assessment tool for the analysis of faulty movement patterns, especially in regard to preventing injury at the trunk, hip and knee (Myer et al., 2012). The inference here is that FPTs in the form of squatting tests allow for the examination of the biomechanics of ACLR and ACLD knees in the real-life settings of athletes.

Myer et al., (2012) explain that SLS is an exercise used for developing leg weight-bearing capabilities, and improving the angle of knee flexion. Wilk et al., (2012) used SLS to examine single leg stance / balance. Alentorn et al., (2009) used the SLS for the purposes of evaluating the risk of leg injury. SLS exercises strengthen the hip and are used to activate the gluteus medius, which provides primary hip stability for both frontal and transverse planes. Kristianslund and Krosshaug (2013) state that whenever the gluteus medius is inactive or becomes weak, femoral adduction, internal rotation and an increase in the displacement of the medial knees occur, all of which are risk factors for ACL injuries. Krause et al., (2009) point out that SLS exercises provide high peak gluteus maximus and medius activation in comparison to double limb standing exercises. The inference may be made here that the single leg squat is a closed kinetic chain exercise that can be started during the initial phases of ACL rehabilitation, and it can be used to assess functional dynamic positioning in athletes.

The SLS has also been used as an injury assessment tool for the knees. They have been used in biomechanical analysis and clinical screening assessments to examine injuries that have already occurred in the lower limbs, and to screen athletes to assess the potential for future injury. The SLS has also been used for the evaluation, screening and assessment of movement dysfunction. Chmielewski et al. (2007) evaluated the intra and interrater reliability of the single leg squat as a movement assessment. Poulsen and James (2011) used an ordinal scale to grade single leg squat motion, and Crossley et al. (2011) used an ordinal scale to grade single leg squat performance. This research indicates that functional evaluation of the SLS is an indicator of hip muscle function. It was found that those respondents with poor functional performance in SLS also display delays in hip abduction activation. This study suggests that the single leg squat can be used as a clinical screening tool. The research is significant given the finding that hip abduction also affects the kinematics and kinetics of the knee, with abnormal hip movements being one of the reasons for the onset of PFPS. Okada et al., (2011) point out that SLS are simple and can be used in smaller settings such as a physician's office or clinic. Claiborne et al., (2006), state that SLS can be used to identify core strengths, and used instead of landing, cutting and running tasks. This implies that SLS tests can also mimic the same functional movements of running. Willson and Davis (2008) used SLS to study injury to the lower limbs and lower limb angles. This indicates that the SLS can be used to evaluate the kinematics of injured limbs. In the SLS tests used by Willson and Davies (2008), the SLS was used to distinguish respondents with and without PFPS.

Yamazaki et al., (2009) examined the angles of the lower limbs in ACLD during SLS. This was because most injuries occur during a single leg landing and due to the valgus position of the lower limb. A comparison between injured and non-injured legs revealed that the uninjured legs showed much less external rotation of the knee and the hip, less flexing of the knee, and enhanced levels of knee adduction. This was true of both the male and the female respondents. It may be noted here that all the respondents investigated by Yamazaki et al., (2009) suffered from ACLD knees and required an ALCR intervention. Most of the respondents could not finish a proper SLS. However, the research exhibits a limitation, as Yamazaki et al., (2009) point out that most of the ACLD respondents could not do a proper full depth SLS and still maintain their balance; therefore, only SLS of half depth could be used. Moreover, this research did not examine the kinetics variables of the ACLD knees, which could have given some indication of variations between injured and non-injured limbs.

There has been limited research into using the SLS, even though it is commonly used as an assessment tool in physiotherapy and rehabilitation to monitor patients' performance and recovery following a knee injury (Weeks et al., 2012). Although it is widely used, not much is known about the validity and reliability of the SLS, or its appropriateness for carrying out a comparison between people with knee pathology and those with healthy knees. Crossley et al. (2011) found the SLS to be a reliable tool for assessing hip dysfunction when it is used by trained physiotherapists; however, this research was carried out with an asymptomatic respondent. According to DiMattia et al. (2005), although the use of the SLS is widespread, no standardised method exists for the prescription of this exercise, and the kinematic outcome measures in relation to pathology and recovery have, to date, not been supported by motion analysis evidence.

According to Lewis (2015), when selecting assessment exercises for patients with knee pathologies, it is necessary to examine the kinematic differences between various tasks. It has been noted that tasks involving SLS are usually performed using greater abduction of the knee. However, walking down the stairs (step down tasks) involves greater hip adduction. The goal of such exercises is to rehabilitate the patient in terms of reducing knee abduction and adduction of the hip. Therefore, as Lewis (2015) points out, both these exercises involve different levels of difficulty and have to be used progressively. For example, SLS is more proper for a person suffering femeroacetabular impingement (FAI), which occurs due to the combined action of hip adduction and flexion. In the same way, for a person suffering from PFPS, walking down the stairs may not be appropriate, as this involves enhanced levels of hip adduction, which is linked to PFPS. Conversely, step down tasks may be more appropriate for individuals suffering from an ACL injury that occurred a long time ago, as increased abduction of the knee can contribute towards this injury.

In conclusion, little is known about SLS and its relationship to angles and moments in pathological motion at the knee. As a tool that is commonly used by rehabilitation professionals, it may be suggested that a thorough investigation of the SLS with healthy populations would be useful to obtain a suitable reference for normal knee function; followed by under pathological conditions, in order to map deficits in lower limb function. This should be carried out in relation to a wide range of pathologies in order to provide evidence for the efficacy of the SLS as a tool used in physiotherapy to assess knee function. Specialists involved in helping ACL subjects to regain their kinematics and kinetics of the knee often use

SLS as a tool for measurement and assessment, but more research is needed to make this commonly used tool more effective and better understood for knee function assessment. The use of the SLS as an FPT for the analysis of the biomechanics of ACLR and ACLD knees would bridge this gap in the literature on the use of SLS as an FPT.

2.12 Summary of the Literature

From the literature, it has been identified that anterior cruciate ligament (ACL) and patellofemoral (PF) joint injuries are common and cause significant time loss within sports participation, as well as the reported increased risk of osteoarthritis (OA) from ACL and PF joint injuries. There is little data on why individuals with an ACL-deficient (ACLD) or ACL-reconstructed (ACLR) knee subsequently develop OA, but altered kinematics and kinetics are one of the hypothesised reasons. Numerous studies that have investigated walking gait have found significant reductions in knee extensor moment, and small reductions in peak knee flexion angle, with individuals adopting a quadriceps avoidance gait pattern during walking. One of the most common activities pre- and post-surgery is running, and it is not known whether individuals, before and after ACLR knee, have different knee kinematic and kinetic patterns to the healthy population. As well as significant reductions in knee extensor moment, reductions in knee flexion angle have been found in the previously mentioned studies that investigated running gait (Karanikas et al 2009). In addition, a reduction in knee abduction angle was found (Tashman et al. 2004). Therefore, the purpose of this research is to compare the hip and knee kinematics and kinetics before and after for ACLR individuals and healthy non-injured subjects during running.

Several theories on the development and progression of patellofemoral joint OA after ACLR have been explored in the literature. Five to ten years' post-surgery, patellofemoral joint OA is common following ACLR, with prevalence rates of 47% (Culvenor et al. 2013). The lower peak knee extensor moments observed in those with PF joint OA is consistent, with lower peak knee flexion angles a feature of those with PF joint OA (Culvenor et al 2016); the kinematics and kinetics of ACL injury and patellofemoral pain may be similar, but were not investigated in this study. Other studies have investigated excessive hip internal rotation (Boling et al. 2009), hip and knee kinematics, and increased external knee abduction moment (Myer et al. 2010), and they suggest that these are connected to patellofemoral pain, and high knee abduction loads, when landing from a jump (Powers 2010). Hewett (2005) reports that ACL injury risks could be predicted from dynamic knee valgus (hip adduction and internal rotation and knee abduction angle and moments) measures, and they found evidence of a connection between ACL injury and patellofemoral pain, which could reduce injuries, ACLR, and PF joint OA. No studies have investigated kinematics and kinetics before and after ACLR during running, and whether there are similarities between kinematics and kinetics and patellofemoral pain syndrome. There is also a need to determine whether the kinematics and kinetics that drive an ACL rupture are also the kinematics and kinetics involved in the development of PFPS after an ACL injury.

The literature indicates that in ACLD and ACLR knees, there are higher levels of tibial adduction, and that these adduction / abduction movements do not correspond to those of healthy knees. Frontal plane mechanics are significantly affected in both ACLD and ACLR knees, where they exhibit a lower H/Q ratio in both types of knees, which results in increased compression forces on the articular cartilage, initiating its degeneration. In addition, impairments at the knee in the sagittal plane are compensated for by changes in hip and ankle motions, which in turn differentially load the knee. It has also been found that alternations in knee adduction moments and internal tibial rotation are much higher in ACLD/ACLR knees compared to healthy knees. A constantly higher adduction and increased internal rotation results in higher medial contact forces over time, resulting in degeneration of the cartilage, and the development and progression of TF joint OA.

The literature also reveals that the PF joint plays a very important role in knee extensor movements. The PF joint increases the efficacy of the quadriceps muscle by increasing the moments arm of the muscle force with respect to the centre of rotation of the knee. During weight bearing activities, the quadriceps distributes forces around the patella. The inference that may be made here is that if the quadriceps muscles are weak in ACLR or ACLD, this will lead to altered kinetics and loading, which in turn increases the contact pressure on different portions of the joint. This increased contact pressure ultimately leads to deformation of the cartilage and the onset of PF joint OA. Another finding from the literature is that ACLD and ACLR reduce peak knee extensor moments. This reduced ability of the knee to absorb forces in real time during functional movements, leads to changes in PF joint loading. The contact area of the patella that can absorb stress becomes reduced, and this in turn increases loading at the joint. The decrease in contact area leads to increased contact stress, decreased knee extensor moments, and a reduction in the dynamic control of the patella. It may be inferred here that ACLD or ACLR results in altered kinematics and kinetics that predispose the PFPS to the onset of PF joint OA.

2.12.1 Gaps in the current literature

- ACL patients may have different kinematics and kinetics before and after ACL reconstruction, which may result in different performance abilities during running. This may have consequences for long-term rehabilitation and other injuries (PFPS and OA), and return to sport for the individual. Currently, there is no research detailing kinematics and kinetics before and after ACL reconstruction during running and SLS.

- The majority of studies have investigated only single tasks, although some researchers have analysed two different tasks (Cowley et al., 2006; Chappell et al., 2007; Pollard et al., 2004; Houck et al., 2003; Earl et al., 2007), while others have investigated three different tasks (Munro et al., 2012; McLean et al., 2005; Dwyer et al., 2010; Willson & Davis, 2009). However, no investigation to date has linked the relationship between kinetic and kinematic variables during running and single-leg squats before and after ACL reconstruction.

- There is a gap in the literature on the biomechanical differences between ALCR and control group participants during running. The existing studies suffer from the limitation of small sample sizes and the use of instruments such as treadmills, which do not mimic the actual conditions of running. Because of these limitations, it is difficult to generalise the findings from such studies to all cases of ACLR knees.

- There is a gap in the literature concerning measuring hip joint kinematics and kinetics after ACLR to show how altered kinematics and kinetics lead to PFPS as a risk factor. Due to this, it is difficult to draw generic and concrete conclusions about alterations in ACLR groups. In addition, there is a gap in the literature on kinematics and kinetics six to eight months after ACLR during running and SLS tasks. This is important, as existing studies show that biomechanical alterations that occur due to ACLR are time dependent and can lead to the onset of PFPS and OA.

2.13 Research Question

The overriding research question is to determine whether there is an alteration in kinematics and kinetics of hip and knee joints related risk factors for patellofemoral pain syndrome and OA before and after ACLR during running and SLS. This will be answered through the following objectives and null hypotheses:

1. To investigate the hip joint frontal and transverse plane movements six to eight months after ACL reconstruction during running between limbs (injured and non-injured).

   Null Hypothesis: There will be no statistically significant differences in the hip joint frontal and transverse planes movements six to eight months after ACL reconstruction during running between limbs (injured and non-injured).

2. To investigate the hip joint frontal and transverse plane movements six to eight months after ACL reconstruction during SLS between limbs (injured and non-injured).

   Null Hypothesis: There will be no statistically significant differences in the hip joint frontal and transverse planes movements six to eight months after ACL reconstruction during SLS between limbs (injured and non-injured).

3. To compare the hip joint frontal and transverse plane movements six to eight months after ACL reconstruction with a healthy control group during running.

   Null Hypothesis: There will be no statistically significant differences in the hip joint frontal and transverse plane movements between six to eight months after ACL reconstruction individuals and a healthy control group during running.

4. To compare the hip joint frontal and transverse plane movements six to eight months after ACL reconstruction with a healthy control group during SLS.

   Null Hypothesis: There will be no statistically significant differences in the hip joint frontal and transverse plane movements between six to eight months after ACL reconstruction individuals and a healthy control group during SLS.

5. To investigate the knee joint sagittal and frontal plane movements six to eight months after ACL reconstruction during running between limbs (injured and non-injured).

   Null Hypothesis: There will be no statistically significant differences in the knee joint sagittal and frontal plane movements six to eight months after ACL reconstruction during running between limbs (injured and non-injured).

6. To investigate the knee joint sagittal and frontal plane movements six to eight months after ACL reconstruction during SLS between limbs (injured and non-injured).

   Null Hypothesis: There will be no statistically significant differences in the knee joint sagittal and frontal plane movements six to eight months after ACL reconstruction during SLS between limbs (injured and non-injured).

7. To compare the knee joint sagittal and frontal plane movements six to eight months after ACL reconstruction with a healthy control group during running.

   Null Hypothesis: There will be no statistically significant differences in the knee joint sagittal and frontal plane movements between six to eight months after ACL reconstruction individuals and a healthy control group during running.

8. To compare the knee joint sagittal and frontal plane movements six to eight months after ACL reconstruction with a healthy control group during SLS.

   Null Hypothesis: There will be no statistically significant differences in the knee joint sagittal and frontal plane movements between six to eight months after ACL reconstruction individuals and a healthy control group during SLS.

9. To investigate the hip joint frontal and transverse plane movements before and after ACL reconstruction during running between limbs (injured and non-injured).

   Null Hypothesis: There will be no statistically significant differences in the hip joint frontal and transverse plane movements before and after ACL reconstruction during running between limbs (injured and non-injured).

10. To investigate the hip joint frontal and transverse plane movements before and after ACL reconstruction during SLS between limbs (injured and non-injured).

    Null Hypothesis: There will be no statistically significant differences in the hip joint frontal and transverse plane movements before and after ACL reconstruction during SLS between limbs (injured and non-injured).

11. To compare the hip joint frontal and transverse plane movements before and after ACL reconstruction with a healthy control group during running.

    Null Hypothesis: There will be no statistically significant difference in the hip joint frontal and transverse plane movements between before and after ACL reconstruction individuals and a healthy control group during running.

12. To compare the hip joint frontal and transverse plane movements before and after ACL reconstruction with a healthy control group during SLS.

    Null Hypothesis: There will be no statistically significant difference in the hip joint frontal and transverse plane movements between before and after ACL reconstruction individuals and a healthy control group during SLS.

13. To investigate the knee joint sagittal and frontal plane movements before and after ACL reconstruction during running between limbs (injured and non-injured).

    Null Hypothesis: There will be no statistically significant differences in the knee joint sagittal and frontal plane movements before and after ACL reconstruction during running between limbs (injured and non-injured).

14. To investigate the knee joint sagittal and frontal plane movements before and after ACL reconstruction during SLS between limbs (injured and non-injured).

    Null Hypothesis: There will be no statistically significant differences in the knee joint sagittal and frontal plane movements before and after ACL reconstruction during SLS between limbs (injured and non-injured).

15. To compare the knee joint sagittal and frontal plane movements before and after ACL reconstruction with a healthy control group during running.

    Null Hypothesis: There will be no statistically significant difference in the knee joint sagittal and frontal plane movements between before and after ACL reconstruction individuals and healthy control group during running.

16. To compare the knee joint sagittal and frontal plane movements before and after ACL reconstruction with a healthy control group during SLS.

    Null Hypothesis: There will be no statistically significant difference in the knee joint sagittal and frontal plane movements between before and after ACL reconstruction individuals and healthy control group during SLS.

Hypothesis rule:

The null hypothesis will be rejected if there is a significant difference in both planes, and partially rejected if there is one significant difference but in the other there is not, and if there is no significant difference in both, it will be accepted.

CHAPTER THREE

BIOMECHANICS OF FUNCTIONAL TASKS WITH ANTERIOR CRUCIATE LIGAMENT DEFICIENT (ACLD), ANTERIOR CRUCIATE LIGAMENT RECONSTRUCTION (ACLR) AND PATELLOFEMORAL PAIN SYNDROME: A SYSTEMATIC REVIEW

3.1 Introduction

Anterior cruciate ligament (ACL) injuries are common and cause significant time loss from sports participation (Webb and Corry 2000), as well as the reported increased risk of osteoarthritis (OA) from ACL injuries (Lohmander et al., 2007). There is little data available on why individuals with an ACL-deficient (ACLD) or ACL-reconstructed (ACLR) knee subsequently develop OA, but altered kinematics and kinetics are some of the hypothesised reasons. Numerous studies that have investigated running gait have found significant reductions in knee flexion moment and small reductions in peak knee flexion angle, with individuals adopting a quadriceps avoidance gait pattern during running (Sigward et al 2015; Karanikas et al., 2009; Bush joseph et al., 2001; Rudolph et al., 2001). One of the most common activities pre- and post-surgery is running, and it is not known whether individuals have different knee kinematic and kinetic patterns to healthy individuals before and/or after ACLR knee.

Running is one of the most common activities before and after ACLR, and possibly increases the risk of OA (Tashman et al., 2007). Patellofemoral (PF) OA, also combined with tibiofemoral OA, is prevalent and has been highlighted in a recent review, which reveals the impact on a median of 47% (range 12–76%) of people 10 years after ACLR (Culvenor et al., 2013). Evidence is emerging that ACL injuries share some similarities with kinematics and kinetics, such as patellofemoral pain, even though this review by Culvenor et al., (2013) did not specifically investigate kinematics and kinetics for ACL injury. Excessive hip and internal rotation within knee and hip kinematics (Boling et al., 2009), together with increased external knee valgus moment (Myer et al., 2010), are likely to contribute towards high knee valgus loads when landing from a jump (Powers 2010). In addition, ACL injury risks have been predicted from measures of dynamic knee valgus during landing (Hewett 2005).

Preventative strategies that aid in targeting both problems simultaneously to reduce ACL injuries, ACLR, as well as post-traumatic PFJ OA, may be discovered as a result of establishing a link between patellofemoral pain and ACL injury. To date, few studies have investigated both kinematics and kinetics post ACLR during running, and kinematics and kinetics has been shown to be similar to patellofemoral pain syndrome. However, as mentioned previously, in general clinical space, the option of asking an individual to run so that they can be assessed is limited; therefore a more space-optimised clinical assessment is needed. Thus, the single leg squat (SLS) has been chosen as the measure for assessing these individuals.

Only one study has been carried out on SLS with ACL patients, and this was by Yamazaki et al., (2009). Yamazaki et al., (2009) examined the angles of the lower limbs in ACLD during SLS. This is because most injuries occur on account of a single leg landing and due to the valgus position of the lower limb. A comparison between injured and uninjured legs reveals that the uninjured legs showed much less external rotation of the knee and the hip, less flexing of the knee, and enhanced levels of knee adduction. The exclusion of SLS tasks in this systematic review is because there is only one study that has utilised a SLS task with ACL patients.

Therefore, the aim of this review is to systematically assess the findings from the literature concerning 3D gait deviations in individuals with ACLD, ACLR and PFPS knees, compared to healthy knees, during running, to identify the current level of knowledge in this area.

3.2 Materials and Methods

3.2.1 Literature search strategy

This systematic review has utilised PRISMA to select, identify and appraise appropriate literature sources (Moher et al., 2009). A comprehensive search strategy was devised using the following electronic databases with no date restrictions: (i) MEDLINE via OVID; (ii) AMED via OVID; (iii) CINAHL via EBSCO; (iv) SPORTDiscus via EBSCO. The MEDLINE search strategy was adopted for the other databases. Keywords were included for ACL injury groups (‘anterior cruciate ligament’, ‘ACL’, ‘biomechanics’, ‘moment’, ‘angle*’, ‘rotation’, ‘kinetic*’, ‘kinematic*’, ‘joint load’, ‘gait’ or ‘walking’, ‘running’, ‘jogging’, ‘patellofemoral pain syndrome’, ‘PFPS’, ‘PFJ OA’), and for the patellofemoral group (‘patellofemoral pain syndrome’, ‘PFPS’, ‘PFJ OA’ ‘biomechanics’, ‘moment’, ‘angle*’, ‘rotation’, ‘kinetic*’, ‘kinematic*’, ‘joint load’, ‘running’, ‘jogging’,). The search was limited to the English language and full-text articles. The results of database searches were reviewed, together with appropriate abstracts. In cases where the abstracts suggested suitable papers, full-text versions were retrieved and included in the review. Lists of all research publications considered for inclusion were carefully examined for their suitability for inclusion.

3.2.2 Inclusion and Exclusion Criteria

Only studies that included adults, and were carried out from 1990 to 2015, have been included. The abstract and titles were used to identify relevant studies. The inclusion criteria meant that only studies that deal with knee and hip kinetic or kinematic information on ACLR and ACL injured knees and uninjured control groups were eligible for review. Only studies using three-dimensional biomechanical studies, live human studies, and those that reported kinetic and/or kinematic data, including for all planes of motion, have been included. Studies carried out before 1990 have been excluded, as well as any replication of the data presented. In addition, systematic reviews and meta-analysis studies have been excluded. However, the analysis includes studies that have reported on gait evaluation in individuals with ACL injury before and after reconstruction or patellofemoral disorders during running. Animal studies, case studies and surveys whose subjects fall outside the 18-45 age group, have also been excluded. In addition, studies that used 2D motion analysis, fluoroscopy, and those that investigated other tasks apart from running (e.g. landing from a jump, cutting) have been excluded.

3.2.2.1 Assessment of methodological quality and risk of bias

A modified version of the Downs and Black checklist (Downs and Black 1998) has been deployed to ensure the high quality methodological rating of the studies included. A valid and reliable checklist with the following features has been used: appropriate for assessing both randomised and non-randomised and observational studies. This version allowed for 15 as a maximum score, with a score of ≥12 indicating high methodological quality; 10 or 11 indicating moderate quality, and ≤9 depicting low quality (Munn et al., 2010). Each study has been rated using the 15 item criteria.

3.2.3 Data extraction

Demographic data was extracted for each study, along with data on the kinematic and kinetic variables of the knee and hip joints. The data was independently extracted by one reviewer (Saud Alarifi), and this was checked by Dr Lee Herrington.

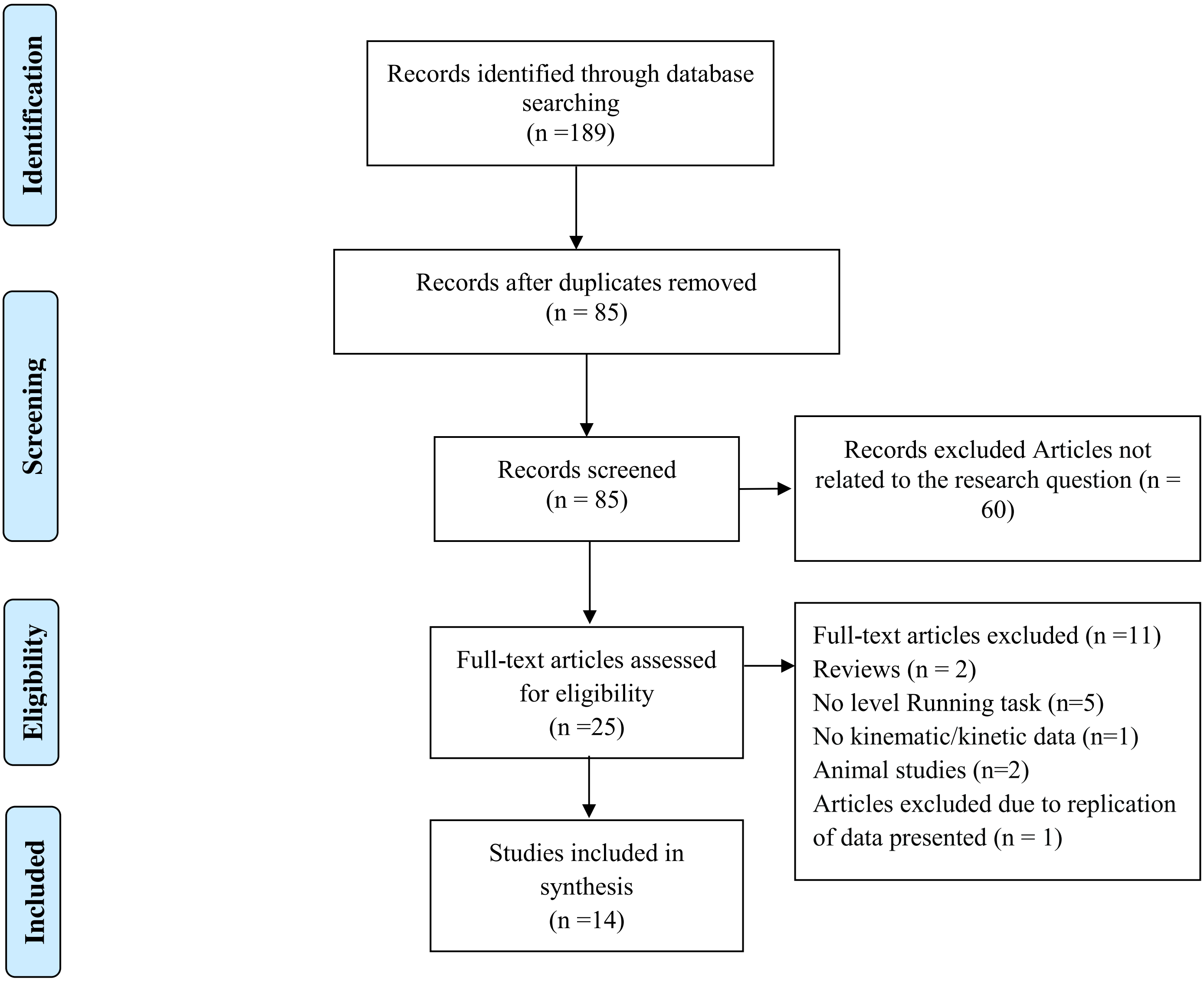


Figure 3.1: PRISMA flow diagram of search strategy for ACL Studies

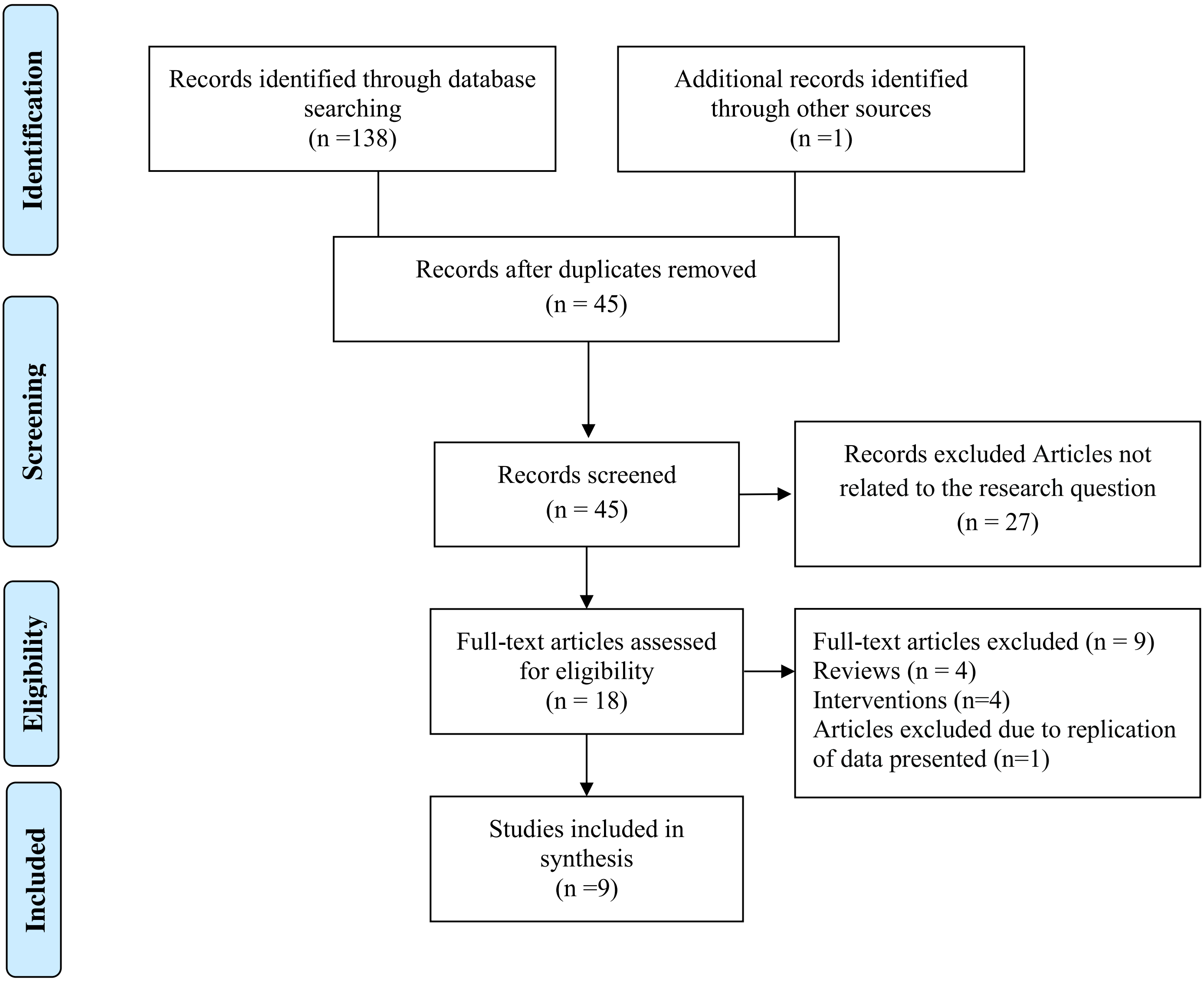


Figure 3.2: PRISMA flow diagram of search strategy for PFPS Studies

Table 3.1: Methodological Quality Rating Score using Modified Downs and Black Scale for ACL Studies.

ACL Studies																
	Item number															
Study	1	2	3	4	5	6	7	8	9	10	11	12	13	14	15	**Total Quality**
Lewek (2002)	1	1	1	0	1	1	1	1	1	1	1	0	0	0	0	10-Moderate
Chemieliwiski (2001)	1	1	1	0	1	1	1	1	1	1	1	1	0	0	0	11-Moderate
Bush-Joseph (2001)	1	1	1	0	1	1	1	1	1	1	1	1	0	0	0	11-Moderate
Kuenze (2014)	1	1	1	0	1	1	1	1	1	1	1	1	0	0	0	11-Moderate
Sigward (2015)	1	1	1	0	1	1	1	1	1	1	1	0	1	0	0	11-Moderate
Karanikas (2009)	1	1	1	0	1	1	1	1	1	1	1	0	0	0	0	10-Moderate
Takeda (2014)	1	1	1	0	1	1	1	1	1	1	1	0	1	0	0	11-Moderate
Tashman (2004)	1	1	1	0	1	1	1	1	1	1	1	0	0	1	0	11-Moderate
Noehren (2013)	1	1	1	0	1	1	1	1	1	1	1	0	0	0	0	10-Moderate
Culvenor (2014)	1	1	1	1	1	1	1	1	1	1	1	0	1	0	0	12-High
Rudolph (2001)	1	1	1	0	1	1	1	1	1	1	1	1	0	0	0	11-Moderate
Tachman (2007)	1	1	1	0	1	1	1	1	1	1	1	0	1	0	0	11-Moderate
Berchuck (1990)	1	1	1	0	1	1	1	1	1	1	1	1	0	0	0	11-Moderate
Patel (2013)	1	1	1	1	1	1	1	1	1	1	1	1	0	0	0	12-High

Table 3-2: Methodological Quality Rating Score using Modified Downs and Black Scale for Patellofemoral Studies.

Patellofemoral Studies																
	Item number															
Study	1	2	3	4	5	6	7	8	9	10	11	12	13	14	15	**Total Quality**
Souza & Powers (2009a)	1	1	1	0	1	1	1	1	1	1	1	1	0	0	0	11-Moderate
Souza & Powers (2009b)	1	1	1	1	1	1	1	1	1	1	1	1	0	0	0	12- High
Dierks (2008)	1	1	1	0	1	1	1	1	1	1	1	1	0	1	1	13- High
Dierks (2011)	1	1	1	0	1	1	1	1	1	1	1	1	0	1	1	13- High
Wirtz (2011)	1	1	1	0	1	1	1	1	1	1	1	1	1	1	1	14- High
Noehren (2011)	1	1	1	0	1	1	1	1	1	1	1	1	1	0	0	12-Moderate
Stefanyshyn (2006)	1	1	1	0	1	1	1	1	1	1	1	1	0	0	0	11-Moderate
Noehren (2012)	1	1	1	0	1	1	1	1	1	1	1	1	0	0	0	11-Moderate
Willson & Davis (2007)	1	1	1	0	1	1	1	1	1	1	1	1	1	1	1	14- High

N.B: A Score of more or equal to 12 defines high methodological quality; a score of 10 to 11 defines moderate methodological quality, and a score that is equal to or less than 9 indicates low quality

3.2.4 Search and Quality of Results

A comprehensive search strategy for ACL studies identified 189 titles, with the last search conducted on the 10th of November 2015 (see Figure 3.1). Following the removal of duplicate publications and conference proceedings, the titles of 85 publications were evaluated beyond the title (i.e., abstract). The full texts of 25 articles were also retrieved, with

14 articles meeting the selection criteria. Tables 3.3 and 3.4 present the characteristics of the studies included. Table 3.1 presents the methodological quality scores ranging from 10 to 12 (out of 15), with an average score of 11. There were two studies of high quality and 12 were of moderate quality. All of the 14 studies scored positively on item five (comparison group) and negatively on item 13 (adjustment for confounders), except four studies. Only one study reported sample size calculations.

The comprehensive search strategy for patellofemoral studies identified 139 titles, with the last search conducted on the 10th of November 2015 (see Figure 3-2). Following the removal of duplicate publications and conference proceedings, the titles of 45 publications were evaluated beyond the title (i.e., Abstract). The full texts of 45 articles were retrieved, with nine articles meeting the selection criteria. Table 3.5 presents the characteristics of the studies included. Table 3.2 presents the methodological quality scores ranged from 10 to 14 (out of 15), with an average score of 11. There were five studies of high quality and four were moderate quality. All nine studies scored positively on item five (comparison group) and negatively on item 13 (adjustment for confounders), except for three studies. Four studies reported sample size calculations.

3.2.5 Patient Selection Bias

The Downs and Black Systematic Evaluation Tool consists of 15 provisions used to measure the quality of the methods that researchers have deployed in their studies. These provisions are grouped into six principal categories whose accumulative scores contribute towards the overall score for the research. The first (patient selection and bias) has been satisfied by all studies. For instance, each of the literature pieces included had their hypothesis, aims or objectives adequately stated, which contributed one point to the overall quality of the literature review. This category provided an ample scenario for examining whether respective researchers prioritised the need to describe their samples adequately. In addition to the clarity of the studies' objectives and aims, each premise managed to score the full four points in the selection/patient bias category of the Downs and Black Systematic Measurement tools. These tools were established by adequately describing the characteristics of the patients included in the project, including a patient sample that was representative of the patients treated in a regular clinical practice, and providing information on how selection bias was prevented. For instance, the research objective of Lewek et al., (2002) was to explore the impact of joint stability and quadriceps strength on gait sequences after ACL injury and reconstruction. Chmielewski et al., (2001) mention the inclusion of 11 injured subjects and ten uninjured subjects (both males and females) who had passed the criteria of the screening evaluation, as the core strategy for preventing selection bias. The analysis by Chmielewski et al., (2001) includes a control variable of uninjured individuals. Nonetheless, only Culvenor et al., (2014); Patel et al., (2003), and Souza and Powers (2009), provide information on how selection bias was prevented.

3.2.5.1 Comparison

The second category (element 5) of the Downs and Black Systematic Evaluation tool addresses whether the inclusion and description of the comparison group (control group) adds to the overall quality of the study. All studies have identified and described the control group (mostly regarding their number and the ACL injury status) meaning the category is a point contributor for the overall methodological quality. Worth noting, however, is that while most studies, including Bush-Joseph et al., (2001), Kuenze et al., (2014), and Karanikas et al., (2009), have described their comparison groups and healthy individuals without an ACL injury background, Sigward et al., (2015) have described the comparison of groups based on the duration after undergoing ACL reconstruction. In another study, Takeda et al., (2014) have described their comparison group based on a specific time following injury.

3.2.5.2 Outcomes

All of the studies have fulfilled the fourth category of the Downs and Black Systematic Evaluation tool, which focuses on the outcomes to be measured in the study. Each study has managed to attain this score by mentioning the primary outcomes to be measured either in the methodology or introduction sections. Among the studies that mention the results to be measured in the introductory part are Lewek et al., (2002), and this includes where the authors evaluated kinematics after ACL reconstruction, as well as the principle outcomes. Meanwhile, Karanikas et al., (2009), as well as Souza and Powers (2009), are among the researchers who have mentioned their primary result measures in the methodology section. The main outcomes of each study have been used correctly, and adequate efforts have been channelled into blinding those measuring the primary results in each investigation.

3.2.6 Reported Findings and Statistical Analysis

The quality and clarity of the description of the results are two of the key measures when considering the studies to be included in a literature review, and most of the studies fulfilled the ninth, tenth, and eleventh criteria. For instance, Noehren et al., (2013) ensured that all findings were reported with clarity and conciseness by including impact loading and joint deterioration. The studies further provide approximations of probability distribution of random differences in the data for the primary outcomes. Nonetheless, while most studies have used statistical analysis to evaluate the principal results, the studies have provided no evidence of bias.

3.2.6.1 Confounders

The distribution of fundamental confounders (e.g. age, sex, height, weight, activity level, sporting activity, player position, dominance, duration symptoms) in all groups of subjects to be compared has not been clearly described in the studies. Only the studies by Chmielewski et al., (2001), Bush-Joseph et al., (2001), Kuenze et al., (2014), Rudolph et al., (2001), Berchuck et al., (1990), and Patel et al,. (2003), for ACL studies, describe this outcome. Even so, most studies have not included adequate adjustments for confounding in the evaluation from which the primary results were drawn, except Sigward et al., (2015), Culvenor et al., (2014), Tashman et al., (2007) and Takeda et al., (2014). The distribution of fundamental confounders in all groups of subjects to be compared is clearly described in all patellofemoral studies. Most patellofemoral studies have not included adequate adjustments for confounding in the evaluation from which the primary results were drawn, except Wirtz et al., (2012), Noehren et al., (2012a) and Willson and Davis (2008).

3.2.6.2 Power

None of the ACL studies mention power and sample size calculations, which is the main limitation of the relatively small sample sizes in most of these studies. Statistical power is difficult to achieve with small sample sizes, and variations in the performance of individuals while running would require higher sample sizes in order to reveal meaningful differences when deploying adaptation strategies used by ACL injured groups. Most of the patellofemoral studies have not mentioned power and sample size calculations, except for Dierks et al., (2008); Dierks et al., (2010); Wirtz et al., (2012), and Willson and Davis (2008).

Table 3.3: Summary of ACLR studies during Running.

Study	Population	Tasks	Findings
Noehren 2013	• 20 females ACLR 5±3 Y, (25±6 Y, 1.6 ±0.1 m, 64±8 kg), • 20F. control., (26±5 Y, 1.6±0.1 m, 61±5 kg)	Treadmill Running	• Reduction in internal knee extensor moment and hip flexion angle (P = 0.08 and P = 0.06, respectively) • No significant differences between-limb for other variables
Tashman 2004	• 6 ACLR after 4-12M, Age 16-50years	Treadmill Running	• Significantly higher knee external rotation angle of 3.8 ± 2.3° compared to control group and during all time points (*P* = .0011) • Significantly higher in knee adduction angle, by 2.8 ± 1.6° (*P* = .0091)
Lewek 2002	• 28 patients, 10 ACLD (6 female, 4 males) • 18 ACLR at 1-2.2Y (9 F/9M) • Age average 25 years	Running	• Significant reduction in knee internal knee extensor moments of ACLR (P = 0.003), and ACL-D (P =0.014) groups compared control group at peak knee flexion
Bush-Joseph2001	• 22ACLR at 22±12 M, 13M/9F, 27±11 Y • 22 Controls, (13M/9F, 29±8 Y)	Running	• Significant reduction in knee peak extensor moment in the ACL-R group compared to the control group (P 0.005)
Sigward et al. 2015	• 12 ACLR after 8-12W. • Age (14-55Y)	Running	• Significant reduction in knee flexion angle in the contralateral limb • Significant reduction in knee extensor moment and impulse compared to the contralateral limb
Kuenze 2014	• 20 ACLR (9 females and 11 males) and 23 healthy controls (11 females and 12 males), (18 and 40Y) • ACLR after 33.9 ± 23.4 months	Treadmill Running	• Significant higher in hip flexion angle and external hip flexion moments • Significant reduction in knee extensor moments (14%–16%, 0.24± 0. 02N.m[kg.m]) compared with healthy controls
Karanikas (2009)	• 35ACLR 3-24M (12 females, 23 males)	Treadmill Running	• Significant reduction in knee extension and flexion angles between limbs (P<0.05)
Tashman 2007	• 16 ACLR after 5-12M. (6 women and 10 men), of mean age 35 years (24–48 years)	Treadmill Running	• Significantly higher knee external rotation angle (*P* = .0011) compared to uninjured limb • Significant increase in knee adduction angle (*P* = .007) compared to uninjured limb
Culvenor 2014	• 36 ACLR after 2±9Y, (18 PFOA&18 no knee OA) • Age 47±10/40±9Y • Sex 10M/11M	Running	• Significant reduction in knee internal rotation to 6.1° (*P* 0.002) for ACLR group with PFJOA and valgus-aligned knees

Table 3.4: Summary of ACLD studies during Running

Berchuck 1990	• 16 ACLD (26±9.5Y, 1.75±0.15m, 80±10.1Kg). • 10 Healthy (5M/ 5F), (26±5Y, 1.67±0.20m, 62±12Kg)	Running	• Significant reduction in knee extensor moments (P<0.05) compared to the control group
Chmielewski et al. 2001	• 11ACLD (9M,2F), (23.8;17±3Y) • 10 Healthy (8M, 2F), (32.2Y)	Running	• Significant reduction in peak knee flexion angle (P > 0.035) between limbs • No difference in kinetics • Significant reduction in peak knee flexion angle compared to control group (P > 0.033)
Patel 2003	• 44 ACLD. (25M, 19F, 29 ±9Y, 1.72 ±0.09m, 727 ±145 N) • 44 Healthy (25 M, 19 F, 30 ±9Y, 1.72 ±0.11m, 718± 162N)	Running	• Significant reduction in peak extensor moment compared to uninjured group (P > 0.024)
Rudolph 2001	• 11 ACLD as copers (2F,9M,30.7Y) • 10 ACLD as non-copers (4F, 6M, 28.1Y) • 10 Healthy (32.2Y)	Running	• Significant reduction in knee flexion angle (39°±2.4 compared with 46.1°±2.4) • Significant reduction in knee extensor moments compared to uninjured subjects
Takeda 2014	• 22 ACLD (11M and 11F, 22Y,19.6 months after injury)	Running	• No significant differences between-limb for all variables during running

Table 3.5: Summary of Patellofemoral studies during Running

Study	Population	Tasks	Findings
Souza & Powers 2009a	• 19 PFPS (27±6Y,1.69±.08m, 64.7±10.4Kg) • 19 controls (26±4Y,1.69±.06m,63.9±6.8kg)	Running	• Significantly higher in hip internal rotation angle compared to the uninured subjects
Souza & Powers 2009b	• 21PFPS Females (27±6Y,1.7±8.1m,64.7±10.4kg) • 20 C Females (26±5Y,1.7±6m,62.9±6.6kg)	Running	• Significantly higher in hip internal rotation compared to the uninjured subjects
Willson& Davis 2008	• 20 PFPS Females (23.3±3.1Y,1.66±.08m,61.7±10.6k) • 20 C Females (23.7±2.6Y, 1.66±.06m,61.1±5.4kg)	Running	• Significantly higher in knee external rotation compared to the uninured subjects (P = 0.06) by 4.3° • Significantly higher in hip adduction compared to the uninured subjects (P = 0.012) by 3.5° • Significant reduction in hip internal rotation compared to the uninured subjects (P = 0.01) by 3.9°
Dierks 2008	• 20 PFPS 5M/15F (24.1±7.4Y,1.71±0.10m,65.75±12.56k) • 20 C 5M/15F (22.7±5.6Y, 1.70±.08m,63.02±9.15kg)	Treadmill Running	• Significant reduction in peak hip adduction angle compared to the control group (P = .044)
Dierks 2010	• 20 PFPS 5M/15F (24.1±7.4Y, 1.71±0.10m,65.75±12.56k) • 20 C 5M/15F (22.7±5.6Y, 1.70±.08m,63.02±9.15kg)	Treadmill Running	• Significant reduction in peak hip adduction angle compared to the control group (P = 0.044) • Significant reduction in peak knee flexion angle compared to the control group (P = 0.034) • Significant reduction in peak knee internal rotation angle compared to the control group (P = 0.001) • Significant reduction in peak knee adduction angle compared to the control group (P = 0.02)
Wirtz 2012	• 20 PFPS F (21.3±2.6Y, 1.7±0.1m,62.9±7.7kg) • 20 C F (21.6±4.4Y, 1.7±0.1m,61.8±9.2kg)	Running	• Significantly higher hip internal rotation angle of 6° (P=0.04) in females with PFPS when ground reaction forces were greatest
Noehren 2012a	• 16 PFPS F (27±6Y, 1.64±0.05m,57.4±4.6kg) • 16 C F (25±4Y, 1.65±0.07m,58.7±6.5kg)	Running	• Significantly higher in peak hip internal rotation and adduction angle for PFPS group (P=0.002, P=0.046 respectively) at 4.6° and 2.2° respectively • Significantly higher in peak knee internal rotation angle for PFPS group at 4.5° compared to the control group (P=0.03)
Stefanyshyn 2006	• 20 PFPS M/F (34.6±9.8Y, 1.7±0.94m,66.8±1.2kg) • 20 C M/F (34.4±10.3Y, 1.76±0.94m,70.8±13.4kg)	Running	• Significantly higher in knee abduction impulses (P = 0.02) for PFPS group impulses (17.0±8.5 Nm*s) compared to healthy subjects (12.5 ± 5.5 Nm*s) • Significantly higher in knee abduction moment (P = 0.04) for PFPS group moments (9.2±3.7 Nm*s) compared to healthy subjects (4.7 ± 3.5 Nm*s)
Noehren 2012b	• 15 PFPS F (27±6Y,1.64±0.05m, 57.4±4.6kg) • 15 C F (25±4Y, 1.64±.07m,57.9±6.5kg)	Running	• Significantly higher in hip internal rotation and adduction (p=0.006, p=0.001 respectively) for the PFPS who did not develop pain group compared to those who did develop pain

3.3 Biomechanics

3.3.1 ACLD Kinematics

In biomechanics, the term ACLD kinematics refers to the nature of the movement of the anterior cruciate ligament deficient knee joint system from a geometrical perspective. This section compares the results from different 3D kinematic studies of ACLD patients. Rudolph et al., (2001) argue that certain individuals can regain their knee functionality after an ACL rupture when pivoting and cutting (copers), while they report instability for non-copers (injured athletes who require surgery to restore functionality) in daily activities. This study investigated movement patterns in 11 copers, 10 healthy uninjured individuals, and 10 non-copers while running. The findings shown in Table 3.4 reveal that a particular gait adaptation was presented, mostly by non-copers, and this has been explained by the reduction in knee flexion. Rudolph et al., (2001) cite results showing that non-copers had a tendency towards reduced knee flexion at heel strike, and reduced first stance peak knee flexion angle; while copers demonstrated identical knee kinematics to the control group participants.

The range of literature on this activity makes it simple to compare the kinematic variations and the strategies utilised in responding to an ACL injury. The reduced knee angle flexion may be a stiffness strategy used to stabilise the knee joint during dynamic exercise. Reduced knee-flexion angles of between 0 to 30° protect the ACL from high impulse forces, thereby reducing joint motion. Similarly, a study by Karanikas et al. (2009) shows a significant reduction in knee extension and flexion angles between limbs ($P<0.05$) among injured patients. Even so, additional studies are required to adequately comprehend the kinematic adaptation of the knee during running for more accurate diagnosis of ACL injuries (Bacchini et al., 2009).

The study by Chmielewski et al., (2001) compared the knee flexion angle between healthy subjects and copers. The study shows that potential copers lowered their knee flexion angle more so than healthy subjects did on their injured side while running. When running, the injured knee, during its preliminary contact, was more extended than in the healthy subjects, and the level of knee flexion angle was lower than that of the uninjured side. Chmielewski et al., (2001), report that at the initial contact during running, the potential copers' knees were more extended than those of the non-injured group ($P= 0.033$), as shown in Table 3.4. Similarly, the study showed lower knee flexion angles among injured patients.

Takeda et al., (2014) proposed that abnormalities in three-dimensional knee kinematics occur because ACLD happens more frequently during high demand exercises. ACLD knees have the propensity to demonstrate greater tibial external rotation and lower knee valgus angles during straight-line running, and a greater difference has been noted during cutting activities. Side cutting exercises are said to apply more torque to the knee than level running. The lowering of the knee valgus angle during more demanding tasks suggests that ACLD patients tend to sustain knee injuries in a less abducted location to prevent greater valgus angle, because of the increased mechanical loading on the knee in side cutting. This view is held due to the researchers observing that the injured ACL becomes unable to provide stability to the knee joint (Takeda et al., 2014), as shown in Table 3.4. Additionally, the patients had increased hip internal rotation angle.

In conclusion, the ACLD knees displayed lower knee flexion angles, while the uninjured group showed higher knee flexion angles and knee valgus angle. During injury, knee flexion angle is reduced. These three sets of groups can be used to compare the extent of injury among athletes.

3.3.2 ACLD Kinetics

ACLD kinetics refers to the reasons for the gait adaptations made due to a deficiency in the ACL. This section compares different kinetic studies for healthy and injured participants. Rudolph et al., (2001) explain that compared to those in the control group, during running activities, ACLD subjects present lower internal knee extensor moment at peak knee flexion angle due to the loss of function of the ACL. Bush-Joseph et al., (2001) (see Table 3.3) and Patel et al., (2003) (see Table 3.4) support these findings. Berchuck et al., (1990) also confirm these findings in their study and report that ACLD subjects showed significantly reduced internal knee extensor moment. Moreover, Rudolph et al., (2001) explain that reduced knee extensor moment is common in non-copers, and functional ability is important when considering ACLD gait adaptations.

When compared with the control group, ACLD copers show biomechanics that are similar, but non-copers compensate by shifting loads at the knee to the hip joint, by increasing the hip flexion angle and decreasing the knee extensor moment (Berchuck et al., 1990). Control over running is transferred away from the knee to the hip, and co-contraction of the hamstrings may be a mechanism that allows this. However, there may be a risk of long-term knee problems, such as greater compression and shear forces when shock absorption is reduced (Hurd and Snyder-Mackler 2007). Other factors may also be at play in knee stabilisation, for instance, the lack of relationship between quadriceps strength and internal knee extensor moment in ACLD copers, which was found to be significantly related in non-copers (Rudolph et al., 2001). Another important factor concerning adaptations in ACLD is muscle activation timing and magnitude (Chmielewski et al., 2001). Kinetic variables hardly showed a notable difference between groups. This study reports that potential copers showed a reduced vertical ground reaction force in loading response, a reduced knee extensor moment, and a greater ankle movement.

Table 3.4 illustrates the study by Berchuck et al., (1990) and shows that the quadriceps avoidance gait, which links hip flexion movement, increases with internal extensor moment reduction as a protection strategy, so that patients avoid quadriceps contraction in mid-stance, and improve hamstring co-contraction. This approach affects both legs. Similarly, Patel et al., (2003) observed a reduction in knee extensor moment after reconstruction, as reported in Table 3.4. Compared with low demand activities, knee stability control requires increased quadriceps strength, therefore there is a connection between internal peak knee extensor moment and quadriceps strength in running tasks, but this is not shown in walking activities. Normal biomechanics and muscle timing in the copers may lead to greater integrity of the knee joint, but there is no evidence concerning ACL and long-term knee health in the literature (Rudolph et al., 2001). While the research discussed above has considered ACLD or ACLR groups, other studies have investigated spatiotemporal, kinematic and kinetic outcomes.

In conclusion, the ACLD research studies show that gait depends on the level of knee stability. The studies have found that the hip flexion movement increases when knee internal extensor moment reduces.

3.3.3 ACLR Kinematics

After ACL reconstruction, the nature of movement is such that it does not return to normalcy. ACLR kinematics refers to the type of movement after ACL reconstruction. Different scientific studies have illustrated various post-reconstruction 3D kinematics in participants, as discussed in this section. In their study, Kuenze et al., (2014) found that ACLR patients experienced a larger hip flexion angle during running when compared with non-injured individuals. While running, Karanikas et al., (2009) report significantly lower knee flexion and extension angles during the stance stage for the injured limb ($P< 0.05$). According to Table 3.3, the observations made were 3-6, 6-12 and 12-24 months post-surgery during running. In addition, Sigward et al., (2015) explored knee loading asymmetries in running gait in early rehabilitation subsequent to ACL reconstruction and note the low flexion angle ($p=0.008$; effect size 0.9).

Karanikas et al., (2009) also report changes in motion in the involved limb verses the contra-lateral limb in a range of patient groups examined 3-6, 6-12 and 12-24 months after ACLR. They showed significantly reduced knee extension and flexion angles in the stance phase of running. These deficiencies in the injured, versus the contralateral limb, persisted after the twelfth month post-surgery. The study further demonstrates a significant correlation between demanding activities and functional adaptation. Regardless, it has not been established whether targeting the recovery towards normal knee biomechanics is a negative or positive aim for people with ACL injuries, and more so for those who go back to more demanding exercises. Karanikas et al., (2009) have further demonstrated the significance of time in the functioning of ACL injured knees during running, and they claim that the ACL injured group showed an alteration in motor strategy that was time-dependent, initially reducing ROM and knee angle, but then improving.

Culvenor et al., (2014) have reported no considerable differences between limbs concerning extension and flexion angles, and this too did not change with time in the study by Tashman et al., (2007). The findings by Tashman et al., (2007) contravene those of Karanikas et al., (2009) at about six months after ACLR; even so, findings concerning extension and flexion angles tallied one year after surgery. This difference in results is attributed to differences in the participant recovery rate between Karanikas et al.'s (2009) group and Tashman et al.'s (2007) group. The tests performed six months after surgery differed between the groups because the participants displayed differences in gait as they were at different stages of recovery.

Tashman et al., (2007) report similar trends with a mean peak adduction angle of 1.9° for individuals with ACLR compared to 0.9° for those in the control group. Tashman et al., (2007) studied downhill running for 2.5m/s in 10 male and six female ACLR patients with a combination of both hamstring (n=9) and patella (n=7) reconstruction at 12 months post-surgery.

Tashman et al., (2007), report that ACLR knees showed a considerable increase in external knee rotation compared to healthy limbs. Meanwhile, Culvenor et al., (2014), claim that notable differences existed between subjects with and without injury for knee internal-external rotation angle. An evaluation of simple effects reveals that patients with an injury and valgus configuration showed a mean of 3.9° (95% confidence range (95% CI) 0.7, 7.1) lower knee internal rotation compared to varus configuration, and no substantial effects were noted. Thus, it is suggested that certain kinematic characteristics, such as valgus-aligned knees, could contribute towards a significantly lower knee internal rotation angle (Culvenor et al., 2014).

A survey by Lewek et al., (2002) shows that the angles of non-injured subjects were similar to injured subjects in the preliminary stance of running. Recession evaluation has a notable effect on the initial stance stage for knee movement and angle in running. Similarly, Noehren et al., (2013) have reported a trend towards lower hip angle while running ($p=0.06$) in the ACL group. No significant differences were noted in knee angle while running

In conclusion, the ACLR kinematics differ from patient to patient depending on their stage of recovery. The research studies show that participants' gait improves with reconstruction; however, the knee's external rotation increases, while knee flexion angle reduces significantly, following reconstruction.

3.3.4 ACLR Kinetics

ACLR Kinetics refers to the reasons for the gait adaptations concerned with ACL reconstruction after injury. Sigward et al., (2015), report that knee moment impulse was (-0.15 Nm*s/kg) in surgical knees while running for some time. Concerning asymmetry, Sigward et al., (2015) state that the reduced knee extensor moment impulse was 1.7 times greater ($p=0.004$; effect size 1.82) in the non-surgical knee when running. Kuenze et al., (2014) also observed that ACLR patients experienced a greater magnitude of external hip flexion moment and reduced magnitude in internal knee extensor moment during running compared to non-injured individuals. In their study, Kuenze et al., (2014) explain that the gait illustrated in sagittal plane knee moments showed probable functionality adaptations due to quadriceps weakness after reconstruction.

Lewek et al., (2002) and Bush-Joseph et al., (2001) also found that weakness in the quadriceps muscles is associated with decreased internal knee extensor moments in an ACLR group compared to a control group, even one year post-surgery. In the study by Bush-Joseph et al., (2001), significant decreases were observed in the peak internal knee extensor moment during higher-demand activities, including running and running followed by a cut. During running, patients' peak knee extensor moment was found to decrease, and this correlated with a decrease in strength. Where patients had relatively weaker quadriceps muscles, they showed even greater reductions in knee internal extensor moments in comparison to individuals with normal quadriceps strength. These results highlight the importance of focusing on quadriceps muscle strength and effective rehabilitation following ACL reconstruction. The data provides evidence that good isokinetic quadriceps muscle strength is suggestive of normal dynamic functioning when performing activities that are physically demanding. However, for individuals that had undergone reconstructive ACL, different results were produced, and while the peak knee extensor moment reduced significantly, there was little correlation with quadriceps muscle strength. Apart from strength, another factor is the patient's apprehension, in addition to reduced proprioception, or micro instability, which could add to the difficulties patients face when required to perform a normal peak knee extensor moment.

In conclusion, the research studies analysed in this section reveal a reduction in the magnitude of the internal knee extensor moment, which is attributed to post-reconstruction adaptations.

3.3.5 Patellofemoral Pain Syndrome studies

Anterior knee pain is a significant symptom of patellofemoral pain syndrome occurring as a result of athletic activity. Souza and Powers (2009a), and Souza and Powers (2009b), used both PFPS and non-PFPS females in their study. Therefore, it was possible to carry out a comparison for the differences in hip kinematics and kinetics between the two groups. Both groups participated in a "running manoeuvre" to determine these differences (Souza and Powers 2009).

3.3.6 Kinematics

This section outlines the nature of movement in patients with patellofemoral pain syndrome. The findings show that when considering some of the activities, those with PFPS had higher levels of hip internal rotation angle and gluteus muscle recruitment, yet lower levels of hip torque production in comparison to those in the control group (Souza and Powers 2009a; Souza and Powers 2009b) as shown in Table 4-5.

Dierks et al., (2008) considered mainly females in their study; the specific ratio was 15 females to five males, but was equal between injured and uninjured runners. This study focused on the relationship between hip strength and kinematics during running for subjects both with and without PFPS. Following an extended run, the study showed that both groups experienced similar decreases in “hip abductor and external rotator strengths,” with injured subjects having “significantly lower hip abduction strength” (Dierks et al., 2008). Thus, the weaker hip abductors were seen to be associated with increased hip adduction angle both while running and after running (Dierks et al., 2008). In a related study, Dierks et al., (2010) found that individuals with PFPS had less overall motion than the control group participants. Therefore, through the measurement of different subgroups in this study, it was suggested that PFPS could result in unique kinematic mechanisms (Dierks et al., 2010) as shown in Table 4.5.

Since females are more likely to develop PFPS, commonly indicated through abnormal mechanics, Noehren et al., (2012a) studied the differences in “hip, trunk, and foot kinematics” in females with and without PFPS. The study found that those with PFPS had higher rates of hip abduction and internal rotation angles due to weakness in the external rotator muscles (Salsich et al., 2012). An increase in pain causes adaptive changes in gait. The study conducted by Dierks et al., (2008) indicates that low motion in PFPS participants could be due to a strategy that focuses on pain reduction through limiting movement. At the same time, it shows that joint motion increased, and the highest pain level was reported at the end of the run due to increased movement.

The study by Dierks et al., (2008) confirms the previous studies by Souza and Powers (2009 a,b). Moreover, the study by Dierks et al., (2008) considered subjects both with PFPS and those without PFPS. In conclusion, the studies show that patients with PFPS had higher levels of hip internal rotation and adduction angles, and patients with PFPS exhibited more motion than the healthy subjects.

3.3.7 Kinetics

PFPS Kinetics refers to the reasons for the gait adaptations concerned with PFPS. As described by Stefanyshyn et al., (2006), PFPS and excessive knee abduction moment are linked, as after six months of running tasks, patients that developed PFPS were observed to have greater knee abduction moment and impulse in baseline measurements when compared to a similar age group that did not develop PFPS. Three-dimensional (3D) kinetic analysis was used in this study, which reports that during the loading of the stance leg, patients with PFPS showed significantly greater knee abduction moments and impulse than those in the control group. Research by Stefanyshyn et al., (2006) indicates that during a running task, the presence of high knee abduction loads can predict PFPS.

3.3.8 Cause of Patellofemoral Pain

According to Stefanyshyn et al., (2006), patellofemoral pain and joint diseases are possibly linked to lateral ligament stress due to knee internal adduction moment increasing from the active hip adduction angle. Stefanyshyn et al., (2006) claim that during running tasks, the presence of high knee valgus loads could predict PFPS as increased hip-adduction, and external rotation angles may likewise be linked to amplified knee valgus.

During knee valgus postures, Noehren et al., (2012a) found that increased lateral patellar displacement could be related to PFPS. In addition, the possibility of lateral PFJ OA may be increased by greater PFJ stress because of knee valgus alignment (Shultz et al., 2010). Patellofemoral joint osteoarthritis can occur due to unusual stress being placed on the patella by the patella alta, and the enlarged Q-angle, along with other soft tissue malfunctioning (Kim and Joo 2012). An increase in Q angle results in the development of knee valgus.

A comparison between PFPS patients and a healthy control group in retrospective studies shows that the findings do not consistently support the importance of increased knee valgus in PFPS patients (Dierks et al., 2008). According to Almeida et al., (2016), incredibly dynamic knee valgus is a neuromuscular control anomaly of the lower appendage. The condition develops a force on the side of the patella, amplifying a compressive burden between the side of the femoral condyle and the patella. However, pain could deter PFPS patients from adopting knee valgus positions.

In a study of 20 PFPS patients, Stefanyshyn et al., (2006) focused on subjects both with and without PFPS "during the stance phase of running." According to the study, PFPS pain is believed to develop from frontal plane loading. The objective of the survey was to measure knee abduction moment and impulse. Data was collected at the beginning of the running season to eliminate bias. Moreover, the researchers avoided data collection at the end of the running session, as it was possible that the subjects would develop muscles throughout the running season. The results show that subjects with PFPS had higher "knee valgus impulse" at P=0.026 than the control group participants. During their research, Stefanyshyn et al., (2006) found that patients with PFPS demonstrated increased knee valgus moment when running. According to the findings of these studies, different knee conditions may cause patellofemoral pain.

In summary, ACL injuries, or other forms of knee injury, may cause patellofemoral pain syndrome. The condition also affects knee stability and results in functionality adaptations exhibited by a difference in gait between patients with the disease and healthy subjects. The kinetic and kinematic tests performed with these participants show that increasing the hip internal rotation and adduction angle are related to this condition.

3.3.9 Similarities between PFPS and ACL Studies during Running

Although the kinematics and kinetics of ACL injuries have not specifically been investigated, there is developing evidence that subjects with ACL injuries share some similar kinematics and kinetics to subjects with patellofemoral pain (Souza and Powers 2009). For instance, the mechanism of ACL damage usually entails irregular motion of the anterior and transverse planes. This damage positions the knee at valgus and internal rotation angles. Similarly, patients with PFPS exhibit increased internal knee abduction moment (Aminaka et al., 2011).

Several patellofemoral studies have reported hip and knee kinematics, specifically, excessive increase in hip internal rotation and adduction angles. These include studies by Willson and Davis (2008); Noehren et al., (2012a), and Noehren et al., (2012b), who report greater hip adduction angles. Also, Souza and Powers (2009a); Souza and Powers (2009b); Noehren et al., (2012a); Noehren et al., (2012b), and Wirtz et al., (2012) indicate greater hip internal rotation angles. Meanwhile, Dierks et al., (2010) report increased external knee abduction and reduction in knee flexion angles. Also, Stefanyshyn et al., (2006) report increased external knee abduction moments. All such studies indicate that kinematics and kinetics for patellofemoral pain are likely to contribute towards high knee valgus loads when running (Powers 2010).

ACLD studies report reduced knee flexion angle (Rudolph et al., 2001; Chmielewski et al., 2001) and reduced internal knee extensor moment (Berchuck et al., 1990; Patel et al., 2003; Rudolph et al., 2001; Lewek et al., 2002). Furthermore, ACLR studies have reported a reduction in flexion angle (Sigward et al., 2015; Karanikas et al., 2009). Noehren et al., (2013); Bush-Joseph et al., (2001); Lewek et al., (2002); Sigward et al., (2015), and Kuenze et al., (2014) have all reported decreased internal knee extensor moment. Other studies have reported greater internal rotation angle (Tashman et al., 2007; Tashman et al., 2004; Culvenor et al., 2014) and knee adduction angle (Tashman et al., 2007; Tashman et al., 2004).

PFPS and ACL studies show similarities in the reduction in knee flexion angle and lower knee internal rotation angle. This finding is relevant because these two conditions occur because of similar risk factors. Therefore, similar therapeutic approaches can be applied to both conditions. Notably, no studies have been found so far that have explored hip adduction and internal rotation angle and moment, which may increase knee valgus angle and moment during running. For instance, a study by Imwalle et al., (2009) observed that hip rotation may contribute towards the hazardous kinematics of the coronal plane, leading to ACL damage. For this reason, reduced hip abduction strength is associated with a knee injury. The studies cited above have addressed relevant matters such as knee valgus, rotation, and flexion of the knee in ACLD patients and after ACLR. Some studies (Kuenze et al., 2014; Noehren et al., 2013; Tashman et al., 2007; Tashman et al., 2004; Karanikas et al., 2009) have explored lower knee flexion angle, internal knee extensor moment, greater knee internal rotation and adduction angle. However, these studies used a treadmill for running, and it was concluded that the mechanics of treadmill running cannot be generalised to over-ground running (Sinclair et al., 2013). The studies demonstrate a limitation through their use of relatively small sample sizes since none showed thorough power and sample size calculations. For this reason, their results cannot be generalised to the wider population. Consequently, the researcher needs to take an overall view when evaluating the matters of hip adduction and internal rotation angle and moment, knee valgus angle and moment, knee flexion angle and internal knee extensor moment, during running.

According to Souza and Powers (2009), brutal knee valgus is frequently exhibited by females practicing athletics during landing from a jump due to extreme hip internal rotation and hip adduction. The relationship between the patella and the hip occurs as a result of the distal femur articulation with the patella. Preventative strategies to target both problems simultaneously and reduce ACL injuries, and ACLR and post-traumatic PFJ OA, may be possible through the link between patellofemoral pain and ACL injury. The literature suggests some theories for the development and progression of patellofemoral joint OA after ACLR. This condition is common after ACLR, with 47% of those treated affected 5-10 years following surgery (Culvenor et al., 2013). Currently, there is a shortage of research that has investigated hip and knee kinematics and kinetics after ACLR during running, to determine the similarity between kinematics and kinetics with patellofemoral pain syndrome. Future research should consider whether the kinematics that drive an ACL rupture are also kinetics for the development of PFPS.

3.4 Conclusion

Numerous factors may contribute towards the functional variations between the control and ACL-injured groups. As per the aim of the systematic review, this chapter has analysed the scientific reports on the biomechanics of running tasks with anterior cruciate ligament (ACL) injury and patellofemoral pain syndrome. Attempting to generalise the adoption of mechanisms in ACL-injured patients is challenging because of the small sample sizes of participant groups employed in the relevant studies. Additionally, some studies have concluded that treadmill-running mechanics cannot be generalised to over-ground running. Through evaluating studies that have assessed knee injuries during running, it has been possible to examine the biomechanics that appear to be associated with ACLD; ACLR, and PFPS injuries. With regard to the knee, decreased internal knee extensor moment along with shallow flexion angles, are the typical biomechanics apparent in both ACL and PFPS individuals. Increased hip adduction and internal rotation angle at the hip is the most common issue throughout the PFPS studies.

Previous studies have discovered that major risk factors for the development of PFPS are decreased knee flexion angles and internal knee extensor moment (Boling et al., 2009). This is because decreased internal knee extensor moment can lead to a decrease in the dynamic control of the patella; therefore, patellofemoral contact stress could increase even further for the smaller knee flexion angles due to the decreased contact area that results. Repetitive movements in this same position over time could result in PFPS presenting. Furthermore, increased hip adduction and internal rotation angle have been noted following a PFPS injury. However, no ACL study has considered hip adduction and internal rotation angles and moments, and their effect on increasing knee valgus angle and moment during running, even though this is a risk factor for PFPS.

Regardless of the different methods used, all of the studies mentioned above state that people experience gait pattern alterations after ACLR, which persist for about five years after surgery. Most ACLR studies have been conducted around 12 months after reconstruction. Duration after injury and ACLR has a bearing on disfiguration, and misalignments in neuromuscular control may be life-long after undergoing a reconstruction procedure. Some studies suggest different theories for the development and progression of patellofemoral joint OA after ACLR, due to the link between patellofemoral pain and ACL injury. While the careful movement of the involved limb shortly after ACLR may be a useful strategy for avoiding pain and preventing the alteration of kinematics and kinetics, the findings confirm the underlying premise of the existence of gait changes long after ACLR. Scholars presume that patients may never resume normal gait after ACLR, and may show patellofemoral pain syndrome symptoms or run like patellofemoral OA patients long after the corrective surgery. Hip joint kinematic studies carried out after ACLR indicate an increase in hip flexion angle, while kinetic studies show higher external hip flexion moments. PFPS studies reveal that there is a considerably higher hip adduction and increased hip internal rotation among female athletes with PFPS syndrome. Even so, few studies have explored patellofemoral pain syndrome or patellofemoral OA as a result of ACLR, which is a crucial gap in the research in this area.

CHAPTER FOUR

METHODOLOGY AND INSTRUMENTATION

4.1 Instrumentation

The collection of kinematic and kinetic data on the lower limbs in a laboratory with a running track (10m) that utilises a twelve-camera Qualisys Proreflex motion analysis system (Qualisys, Gothenburg, Sweden) sampling at 240Hz, and three force platforms (AMTI BP600900, USA) sampling at 1200Hz, has been used. Passive retro-reflective markers were identified by the infrared (IR) cameras, where Qualisys Track Manager (QTM) provided the visualisation of the output of the cameras and force platforms. An infra-red motion capture system was used, which emits infra-red light that returned from the markers back to the camera. This enabled the two-dimensional position of each marker to be captured. The two dimensional position of each marker, along with the position of the cameras relative to the system, enabled the calculation of the three-dimensional position of each marker (Kaufman and Sutherland 2006). Two cameras are required as a minimum in order to identify each marker when determining its three-dimensional position at any one time (Cappozzo et al., 2005). In the current study, three non-collinear markers were placed on each required body segment, which could be viewed by the cameras in order to record the location and their alignment. The angle between the two segments was calculated once both the alignment and location of the nearby body segment were determined (Kaufman and Sutherland 2006). Inverse dynamics calculated the hip, knee and ankle moments from the force platform and kinematic data (Winter et al., 1990). QTM has been used to provide a connection to the cameras, and to enable calibration of the capture volume, data collection and 3D reconstruction of the retro-reflective markers for the three stages adopted to collect the coordinate data.

The consideration of the size of the capture volume is important, as this contributes towards the system resolution and the accuracy of the position data. In addition, to minimise blind space around the selected capture volume in the field of view of the camera, appropriate camera position is important (Richards et al., 2008; Payton and Bartlett 2007). This research collected variables of interest in the stance phase of running and SLS tasks; therefore, to include all selected movements, an umbrella configuration of twelve cameras was placed around the three force platforms, as shown in Figure 4.1. Running speeds were monitored using Brower Timing Gate Systems (TC-Timing System, USA).

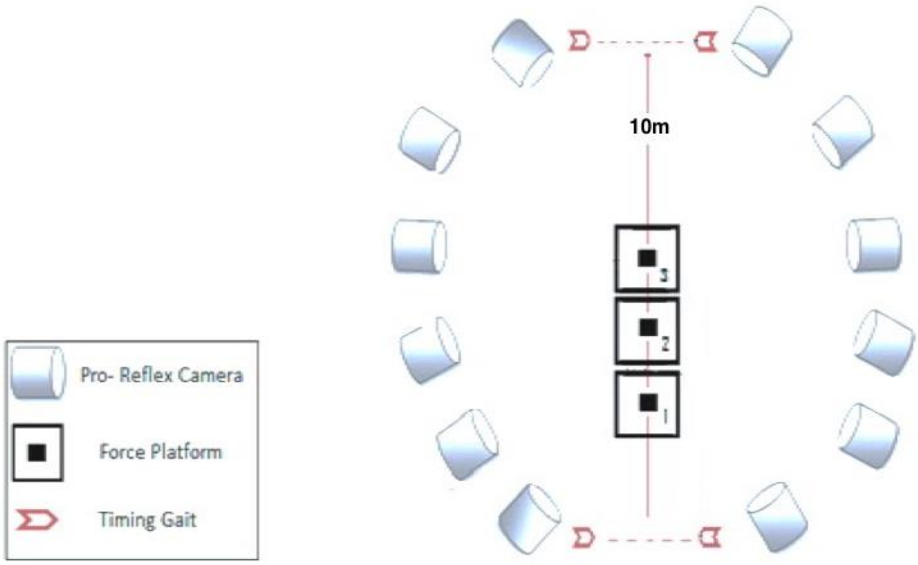
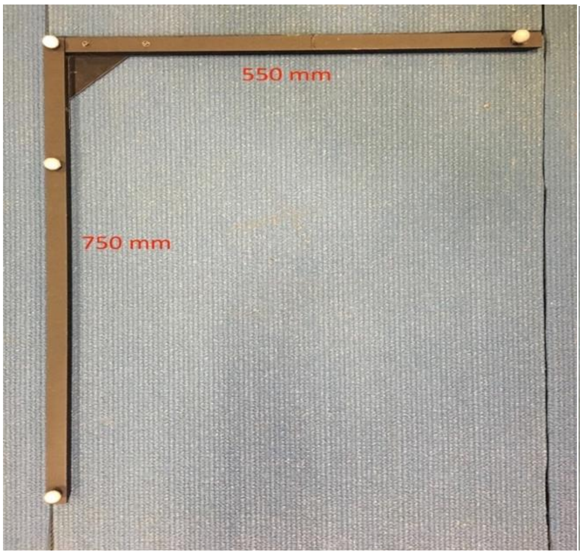


Figure 4.1: Data collection set up.

4.1.1 System calibration

The camera system needed to be calibrated to collect the kinematic and kinetic data. First, a metal frame in the shape of an L with four markers attached to it, which acted as the reference object, was placed along the corner of the force platform. Its position needed to be parallel to its Y and X axes, and the specific distance between the markers and the initial force platform coordinates was measured, that is, the corner of the platform; these measurements were calculated automatically before being typed into the software package (Winter 2009). A reference object is necessary to clarify the coordinates, along with the X axis (medial/lateral), Y axis (anterior/posterior), and Z axis (vertical).

Figure 4.2 shows the static calibration of the motion capture system, which used a rigid L frame and is associated with the laboratory reference frame. Figure 4.3 shows the position of the hand-held wand with reflective markers at a distance of 750.43 mm, which was fixed to calibrate the volume for the dynamic trials that would be carried out. A 45 second capture time was selected to calibrate the volume between the designated height and floor (Richards et al., 2008).



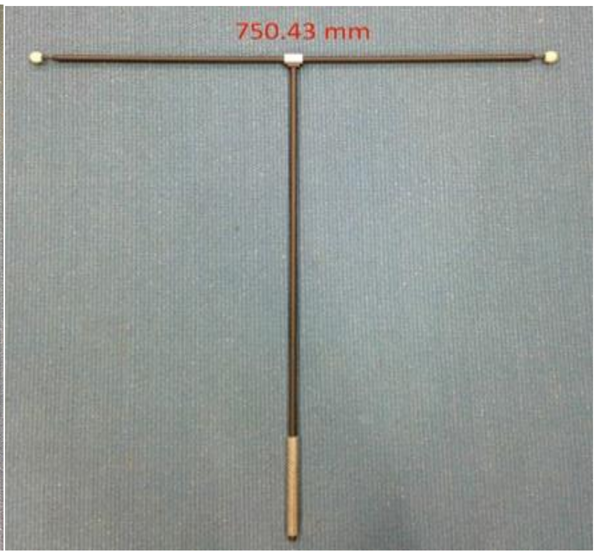


Figure 4.2: L-frame.

Figure 4.3: Handheld Wand.

The calibration of the system determines the accuracy of the marker position in 3D space (Payton and Bartlett 2007), and accurate 3D marker coordinates from the measurements can be achieved with a lower residual measurement. This study adopted the position of a marker in the space to be within 1.00 mm of the true position, because measurement residuals are normally accepted below 1.00 mm.

4.1.2 Marker Placement:

Prior to testing, reflective markers 14.5 mm in diameter were used in all trials for data collection, and the flat-based markers were attached to the skin using hypoallergenic adhesive tape. Cappozzo et al., (1995) demonstrated a method that used groups of markers fixed on a rigid plate that overcame artefact movement created by skin-mounted clusters, as intra-cluster marker movement was disabled. Four markers were suggested by Cappozzo et al., (1997), as well as attaching the clusters to the segment using an overwrap technique. These guidelines were followed, and polypropylene plates were used to arrange clusters of four markers attached to segments with double-sided adhesive tape, before covering with non-migratory tape (Fabio foam Super wrap) that is self-attaching. This prevented the cluster plates from moving during the trials.

Each participant was asked to stand on the force plates so that the cameras could view all markers in a static standing trial to link the anatomical and tracking marker signals to the Qualisys software before transferring to the post-processing software. The position of the anatomical markers enabled a reference point to be noted in order to identify bone movement using only the tracking markers used in the trials. The local coordinate system (LCS) frames and centre of rotation of joints were defined by using 20 anatomical markers on the participant, and markers were positioned at medial and lateral aspects of the joints at anatomical landmarks at distal and proximal ends of the segment, as shown in Table 4.1.

Table 4.1: Visual 3D model segments.

Segment	Proximal radius/joint	Distal radius/joint	Tracking markers
Pelvis	Right anterior superior iliac spine Left anterior superior iliac spine	Right posterior superior iliac spine Left posterior superior iliac spine	
Thigh	Hip joint centre* Greater trochanter	Medial femoral condyle Lateral femoral condyle	Thigh cluster pad (4 tracking markers)
Shank	Medial femoral condyle Lateral femoral condyle	Medial malleolus Lateral malleolus	Shank cluster pad (4 tracking markers)
Foot	Medial malleolus Lateral malleolus	1^{st} metatarsal head 5^{th} metatarsal head	Superior/inferior calcaneus, medial/lateral calcaneus
Virtual foot	Medial malleolus floor Lateral malleolus floor	1^{st} metatarsal head floor 5^{th}metatarsal head floor	
*Hip joint centre is automatically calculated by using anterior and posterior superior iliac spine markers using the regression equation by Bell et al., (1990)			

Anatomical markers were removed when the static markers were successfully captured, so that four markers remained on PSIS and ASIS, eight markers remained fixed to the standard shoes, and 16 markers covered four cluster plates, leaving 28 in total, as shown in Figure 4.4.

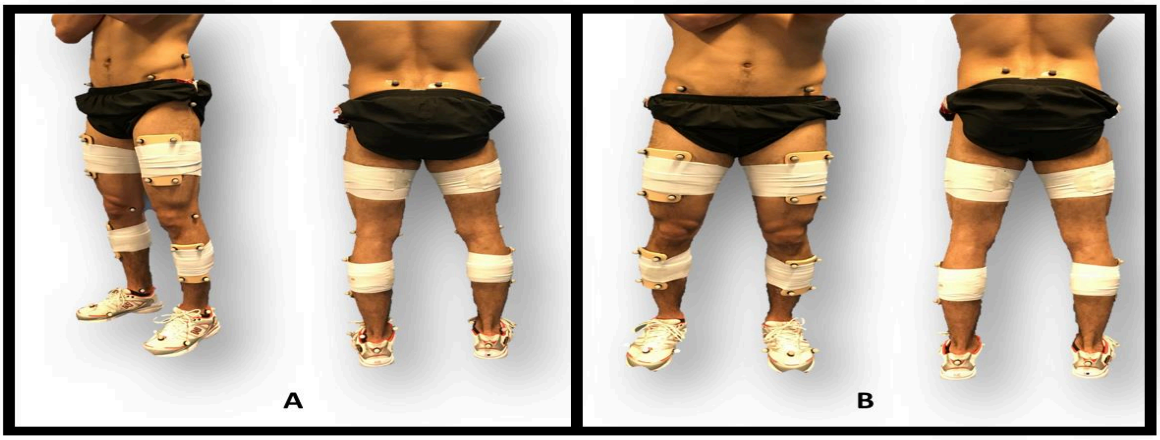


Figure 4-4: (A) Static Trial Markers and (B) Tracking Markers

4.1.3 Biomechanical model

The most common variety of marker sets in clinical use are variations of the Helen Hayes (HH) set (Kadaba et al., 1990). This is useful, as few markers are needed, low resolution imaging is available, and less advanced measurement systems; however, there are disadvantages if they are used for three degrees of freedom (DOF) for the ankle, knee and hip (Della Croce et al. ,2005). Furthermore, the anatomical markers in the HH model are used not only for tracking movement, but also to check inaccuracies in the results on the movement of proximal segments, and errors with the distal segments (Cereatti et al., 2007; Schwartz et al., 2004). During the movement trials, in order to determine each segment's movement and its anatomical significance, the calibrated anatomical system technique (CAST) was used (Cappozzo et al., 1995). CAST offers the advantage of enhanced anatomical relevance, and the markers were a rigid plate with four markers attached to the segment (thigh and shank) close to the middle of it rather than close to joints, to reduce skin movement artefact. It is possible to calculate the relationship between the joint and segment mounted markers in a static calibration, and after removal of the joint mounted markers in a dynamic trial. Their original position can be determined from the segment-mounted markers according to this information. The anatomical markers were used to determine the LCS of each segment. The CAST was used with 6 DOF marker sets, and these were developed as technical markers to track the movement using three rotational and three translational sets; at the joints, segment movement was tracked independently (Cappozzo et al., 2005; Cereatti et al., 2007). The 6 DOF (CAST) model has shown a reduction in some of the errors found in earlier models (Cereatti et al., 2007). 6 DOF is regarded as preferable, since its performance is comparable, as well as it overcoming many of the theoretical limitations of HH (Collins et al., 2009).

For the lower limbs, a six-degree freedom model was constructed, which contained rigid segments attached to the joints (see Figure 4.5). Six variables made up each joint to describe its pose, with three variables describing the position of origin, and three variables describing the 3D rotation. The segment translation has been described by three variables within the vertical, medial-lateral and anterior-posterior axes, with three variables within the sagittal, frontal and transverse segments' rotation axes. To calculate kinetics, the subject's height and weight were entered into the software. The individual segments of the pelvis, thigh, shank, and foot were modelled in order to determine the proximal and distal joint/radius, and the tracking markers, as shown in Table 4.1. The hip joint centre was calculated automatically using anterior and posterior superior iliac spine markers, and the regression equation put forward by Bell et al., (1990).

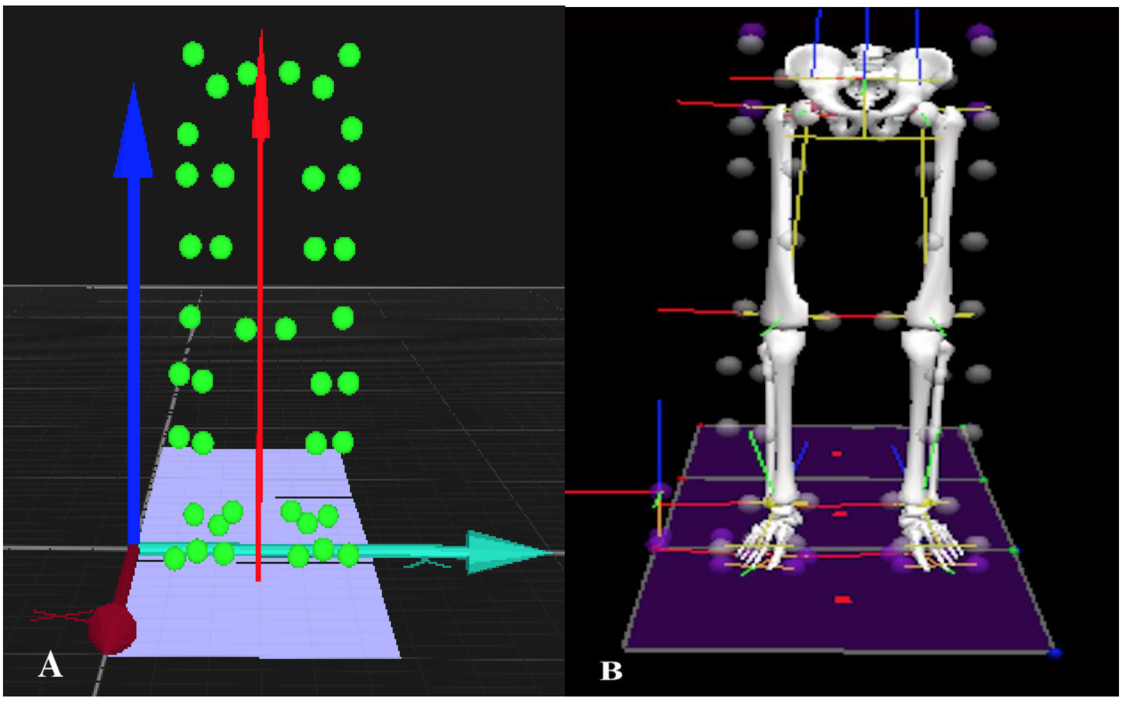


Figure 4.5: (A) Static subject model in QTM. (B) Bone embedded model in Visual 3D (Anterior view)

4.2 Conducting the tests

Participants also completed the Knee Injury and Osteoarthritis Outcome Score (KOOS) questionnaire (Appendix 2), which measures the subjective outcomes after an ACL injury. The International Knee Documentation Committee Scale (IKDC) evaluation scales (Appendix 3) were also used. Participants wore compression shorts and standard training shoes (New Balance, UK) to control the shoe-surface interface before testing. In two or three trial situations, the participants began with three minutes of low intensity warm-up on a cycle ergometer, and they were familiarised with the procedure for testing by practising each task until they felt comfortable. The principal researcher then attached a total of 40 markers to the participant's lower limbs, as described in section 4.1.2. Each participant was asked to stand in a stationary position on the force plate in order to conduct static standing trials. The arms of the participant were held clear of the markers to ensure that any detection was not compromised. The participants were then asked to undertake various tasks, commencing with running and then the SLS task, followed by removal of the anatomical markers.

4.2.1 Functional Scores

Patient-reported measures, and other subjective knee evaluation methods pertaining to knee function, are of vital importance, as they facilitate a comprehensive assessment of the knee after ACL reconstruction (Collins et al., 2011). Subjective knee evaluation is used in assessing the various aspects that are important for diagnosis, treatment, and rehabilitation of adult patients with knee problems, inclusive of sports injuries, PFPS, asymptomatic pain, or osteoarthritis (OA) (Collins et al., 2011). It is essential to gauge the various dimensions that are important to patients, such as quality of life, pain, function, and activity level. Subjective knee evaluation is of key importance in assessing the outcomes of ACLR with reference to the various dimensions described above (Collins et al., 2011). As detailed in the aforementioned sections, this study has delved deep into utilising subjective knee evaluation, as the significant changes in pain or another subscale, may affect biomechanics, and could be quantified by the same. The IKDC has been used for outcome assessment in contemporary studies based on ACL reconstruction, and is one of the most frequently employed patient-reported outcomes (PRO) measures for patients undergoing ACL or those with ACL deficiency (Hambly and Griva, 2010). Research studies have reported that IKDC items outperformed KOOS items in most of the established criteria; although KOOS outperformed the IKDC in outcomes reporting in patients who were in the post-op phase (after 12 months) of ACLR (Hambly and Griva, 2010). This research study has employed IKDC and KOOS, as there is a need for standardisation of the outcome measurement in orthopaedics, in addition to a growing need for detailed assessment of present PRO measures (Hambly and Griva, 2010). The items used to report in IKDC and KOOS collectively facilitate an accurate interpretation of ACL-based clinical trials (Hambly and Griva, 2010).

4.2.2 IKDC subjective knee form score (IKDC 2000)

In 2001, a new IKDC evaluation form was developed (Irrgang et al., 2001). The IKDC 2000 Knee Form includes a demographic form, a current health assessment form, a subjective knee evaluation form, a knee history form, a surgical documentation form and a knee examination form (Irrgang et al 2001). The subjective knee evaluation form used in Chapter Five and Chapter Six has been validated and found to be reliable (Irrgang et al., 2001). It comprises an aggregated score with 10 items. Item 10 relates to knee function prior to injury and current knee function, and is not included in the score. The item scores for items 1–9 are summed up and transformed into a scale from zero to 100. A score of 100 means there is no limitation in ADL or sport activities and no symptoms.

4.2.3 Knee Osteoarthritis Outcome Score (KOOS)

The KOOS is an instrument that is knee-specific. It was developed and introduced for assessing patients' views concerning their knees and associated problems. The KOOS is made up of five subscales, which are pain; other symptoms (symptoms); functions of daily living (ADL); function in sport and recreation (Sp/Rec), and knee related quality of life (QoL) (Roos et al., 1998). Patients must answer nine questions to assess pain; seven questions to assess symptoms; 17 questions regarding ADL; five questions regarding Sp/Rec, and four questions regarding QoL. All questions are graded from 0 to 4 points. A normalised score for each subscale is then calculated, with a maximum of 100 points indicating no symptoms, and 0 points indicating extreme symptoms. KOOS has been used in a number of studies to assess health related QoL after ACL injuries (Barenius et al., 2010; Gerhard et al., 2013; Kostogiannis et al., 2007; Lohmander et al., 2007; Mansson et al., 2013). KOOS provides an outcome measure for all the Scandinavian Knee Ligament Registers (Granan et al., 2009).

4.3 Tasks

4.3.1 Running

Subjects were asked to perform five successful trials for each movement on the over-ground running track in the Human Performance Laboratory. A successful trial required making contact with the third force platform while running along a ten-metre runway, as shown in Figure 4.6.

Figure 4-6: Running Task

The speed was measured using timing gates. Brower timing lights (Draper, UT) were used to ensure the running task was carried out at consistent speeds, and to ensure that only one body part broke the beam, such as the lower torso. These were set at hip height for each participant. To monitor a participant's performance between each test, the time taken to complete the running task was used, with running speeds beyond +/- 5% of average speed excluded from the final results. Five successful trials for each task were completed by the participants, and to overcome the effect of fatigue, they had a break of between one and 1.5 minutes between trials.

4.3.2 Single-Leg Squat

Subjects were asked to stand on one leg on the third force platform. They were asked to squat as low as possible to a minimum of 45° knee flexion, and hold the position for five seconds. During the practice trials, knee-flexion angle was checked using a standard goniometer, and the same examiner carried out the observations throughout the trials. In accordance with the work carried out by Dwyer et al., (2010) and Zeller et al., (2003), the depth of the squat was not controlled, as this ensures a clinical setting in which normal inter-participant variability can arise. A counter was used for each participant during this five second period, with number one initiating the movement, number three indicating the lowest point of the squat, and number five indicating the end. This standardised the tasks for all of the participants, and reduced the impact of velocity on knee angles. Trials were only acceptable if the subject squatted to the minimum required degree of knee flexion, along with maintaining their balance throughout, as shown in Figure 4.7.

Figure 4.7: SLS Task.

4.4 Data processing

Visual 3D motion capture software was used to process the kinematic and kinetic time series data (Version 4.21, C-Motion Inc., USA). A Butterworth 4th order bi-directional low-pass filter, with cut-off frequencies of 25Hz for kinetic data, and 12Hz for kinematics data, filtered the motion and force plate data. Digital filters were used to smooth the data without affecting the signal, and to reduce random noise. Removing random noise in human body movement kinematic and kinetic data is effective when using the Butterworth filter, and is widely used in biomechanical research (Winter et al., 1974). The findings of Yu et al., (1999) were used as the basis of the cut-off frequencies selected. All lower extremity segments were modelled as conical frustum, with inertial parameters estimated from anthropometric data (Hanavan 1964). Figure 4-8 explains how an X-Y-Z Euler rotation sequence was used to calculate joint kinematics, where Z equals internal-external rotation, Y equals abduction-adduction/ varus valgus and X equals flexion-extension. Joint moment information was presented as external moments and normalised to body mass, and three-dimensional inverse dynamic was used to calculate joint kinetic information. External moments have been described in this study; for example, an external knee valgus load will lead to abduction of the the knee (valgus position), and an external knee flexion load will tend to flex the knee (Malfait and Schnitzer 2013).

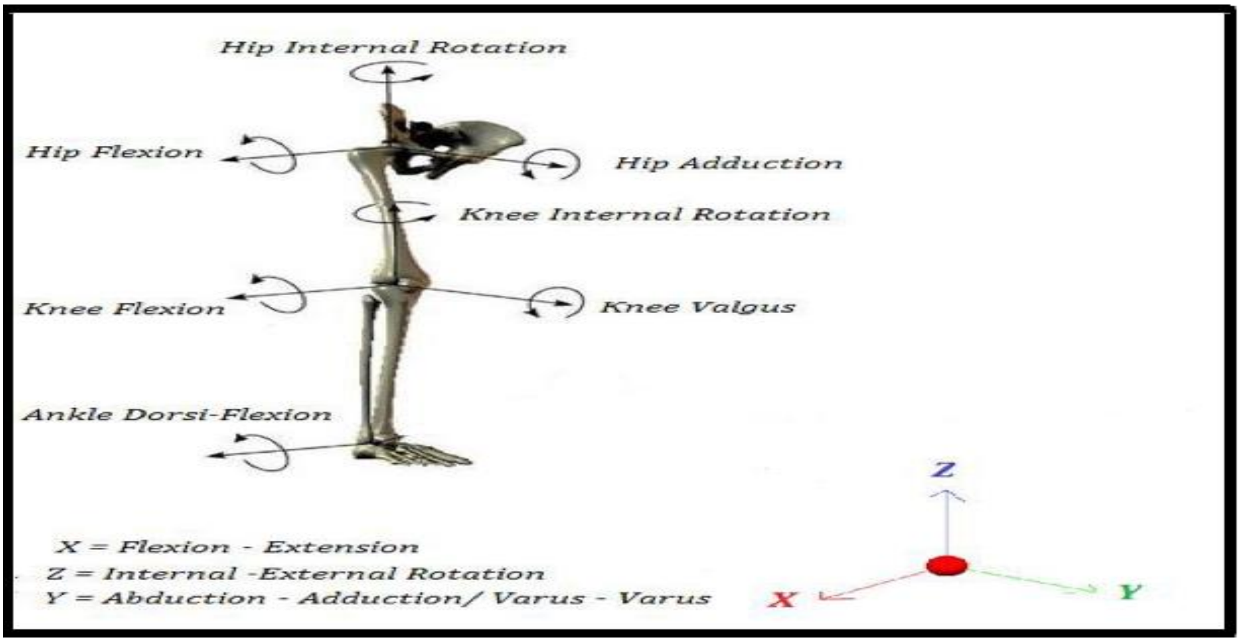


Figure 4.8: Lower extremity segment and joint rotation denotations (adapted from Mclean et al., 2004)

The leg's initial contact (IC) to toe-off (TO) defined the stance phase when kinematics and kinetic data were normalised to 100% for the running task. When VGRF first exceeded 10 N, the IC was defined, and when VGRF fell below 10 N, the TO was defined. For the single leg squat, kinematic and kinetic variables were normalised to 100% of the descending phase when in the single leg squat. When the knee exceeded 10º flexion, this indicated the start of the movement until reaching maximum knee flexion, which ended the defined descend phase. The convention of denoting internal rotation hip adduction, knee valgus and internal rotation as positive was adopted.

4.5 Outcome measures

These discrete variables were selected based on their frequent use in relation to possible ACL, PFPS and OA injury studies, as discussed in Chapter Two. This thesis will address the internal knee extensor moments in sagittal plane, and the external knee adduction moment in frontal plane. Alongside the clinical questionnaire results, the following discrete variables were calculated during the early stance phase (0-50) for each trial, as also shown in Figure 4.9:

a) Peaks of hip adduction angle and moment
b) Peaks of hip internal rotation angle and moment
c) Peak of knee flexion angle
d) Peak of internal knee extensor moment
e) Knee extensor impulse
f) Peak of knee valgus angle
g) Peak of external knee adduction moment
h) Peak of vertical ground-reaction force (VGRF)

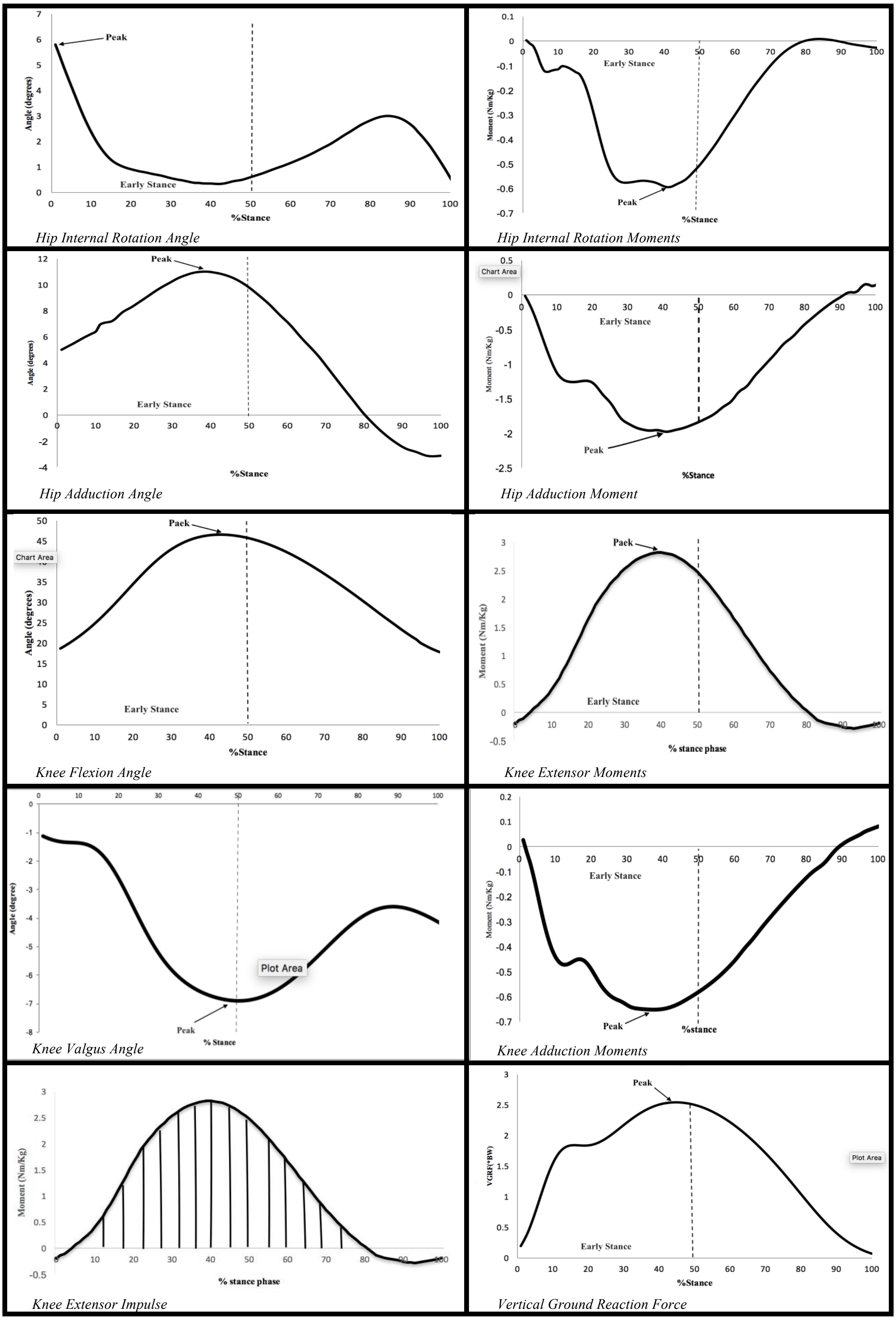


Figure 4.9: Outcome measures

4.6 The Repeatability of Lower Limb Biomechanical Variables Collected During Running and Single Leg Squat Tasks

4.6.1 Introduction

The study's main aims are to establish the differences before and after ACL reconstruction for lower limb kinematics and kinetics during running and SLS, and to discover the differences between those variables. Therefore, it is important to carry out the study using valuable tools that provide stable and reproducible values with small errors of measurement.

Errors in measurement can be reduced by ensuring the accurate placing of markers, checking equipment is not faulty, and making sure no data processing errors occur (Schwartz et al., 2004). It is essential for markers to be placed accurately in order to calculate and determine the exact position of joint centres, as errors in calculating joint kinematic and kinetic data can result from incorrectly identifying bony prominences (Cappello et al., 1997; Stagni et al., 2000; Baker 2006). However, there may be difficulties in accurately positioning the markers on bony prominences, which can cause variability and increase measurement error (Cappozzo et al., 1996). This is because it is sometimes difficult to palpate bony prominences because of being covered with adipose tissue along with muscle layers (Baker 2006).

For an outcome measurement to be valuable, it must provide stable or reproducible values with small measurement errors (Rankin & Stokes, 1998). Knowledge of the reliability and measurement errors associated with each of these screening tools is important (Batterham and George 2003). Random and systematic bias are potential measurement errors, and systematic bias could indicate fatigue or learned effect factors. In addition, random measurement errors cannot be avoided, such as mechanical, psychological and biological factors, which are unpredictable and cannot be anticipated, even if the potential source is known (Portney and Watkins 2000). The performance of participants could fluctuate, and they could lack motivation and fail to follow instructions properly, which refers to psychological and biological factors. Problems with equipment or instrumentation involves mechanical factors, and noise in measurements could be due to uncontrolled confounding variables (Batterham and George 2003).

A key factor in 3D motion analysis is the ability to measure kinematic and kinetic variables reliably, both within and between days. Several authors have reported that measuring kinematic and kinetic variables during the same session is usually more reliable than across different sessions (e.g. Ferber et al., 2002; Ford et al., 2007; Milner et al., 2011; Queen et al., 2006). Marker-placement error influences between-days reliability more so than other factors (e.g. Ferber et al., 2002; Ford et al., 2007; Queen et al., 2006).

Sagittal-plane variables have the greatest reliability compared to those for frontal and transverse planes during running (Ferber et al., 2002; Queen et al., 2006), drop vertical jumps (Ford et al., 2007; Malfait et al., 2014) or single-leg squats (Nakagawa et al., 2014). Motion in the frontal and transverse planes, especially dynamic-knee valgus, presents the solution to the high-risk movements associated with ACL and PFJ injuries (Hewett 2005; Myer et al., 2010). This means that measurement errors in these planes can have a major impact on the identification of athletes at high-risk when using 3D motion analysis.

It is possible to describe reliability either in relative terms (i.e., consistency of rank of score), or in absolute terms (i.e., consistency of actual score) (Weir 2005). Reliability in relative terms can be accurately judged using reliability coefficients, which are calculated using a ratio of true score variance to discover the score variance. Intraclass correlation (ICC; Shrout amd Fleiss 1979) is seen as more important than interclass correlation (i.e., Pearson) in determining relative consistency due to ICC comparing two or more repeated measures of the same variable, and because it can provide estimates of the different sources of variance (Thomas et al., 2005). In addition, measuring the precision of the scores, calculated as the variation in the measurement error or standard error of measurement (SEM), can be done to assess absolute consistency. Both relative consistency (i.e., ICC) and absolute consistency (i.e., SEM) should be taken into account when assessing reliability (Harvill 1991; McGinley et al., 2009; Weir 2005); moreover, absolute consistency is very useful when clinicians are attempting to differentiate real change from changes caused in error (Eliasziw et al., 1994). An estimation of the minimal detectable change (MDC) can be determined by using SEM. According to Hollman et al., (2008), this may be interpreted as the smallest amount of change needed to determine whether a change is real and beyond measurement error. Researchers and clinicians regard MDC values as important in the interpretation of change scores when evaluating the effectiveness of therapeutic interventions (Goldberg et al., 2011).

4.6.2 Aim

The aim is to investigate the within-day and between-days reliability of using 3D movement analysis to measure lower limb kinematic and kinetic variables during over-ground running and single leg squat.

4.6.3 Methods:

4.6.3.1 Participants

12 university students (3 females and 9 males, mean +/- standard deviation age, height and mass are 25 ± 2.00 years, 1.71 ± 0.06 m, 58.7 ± 4.25 kg; 31.4 ± 3.71 years, 1.69 ± 0.09 m, 69.2 ± 11.16 kg respectively) volunteered for the study and were defined as recreationally active. These participants had no history of lower extremity surgery, and for the previous six months had suffered no lower extremity injuries. The definition of injury is all musculoskeletal complaints that have prevented the participants from engaging in their normal exercise routines. All participants were required to read and sign a University of Salford Research, Innovation and Academic Engagement Ethical Approval Panel (HSCR12/64) informed consent statement before testing.

4.6.3.2 Procedure

The participants were asked to perform two sessions of tasks (see section 2.11.2) with a break of one hour on the same day, and then these were repeated one week later. The participants were fitted with standard training shoes (New Balance, UK) to ensure consistent shoe surface interface, and their height and mass were measured before testing commenced. Once individuals fully understood the requirements of the tasks, they were invited to practice each task five times as practice trials before undertaking the measurement tests. The running and SLS tasks are described in section 4.3.

4.6.3.3 Main outcome measures

During the stance phase for each test, the discrete variables of peaks joint angles for the hip, and knee joints, (Hip adduction, Hip internal rotation, Knee flexion and Knee valgus), peaks of hip, and knee joints moment (Hip adduction, Hip internal rotation, Internal knee extensor and External knee valgus) and peak VGRF were calculated. These discrete variables were selected based on their frequent use in relation to possible ACL and PFPS injuries studies, as discussed in Chapter Two.

4.6.3.4 Statistical Analysis

SPSS for Windows (version 13.0) was used for all statistical analysis. Session repeatability of discrete variables used intra-class correlation coefficients (ICC) was used to analyse within (ICC (3, k)) and between (ICC (3, 1)). The ICC classifications of Fleiss (1986) were used, where greater than 0.75 was excellent, between 0.4 and 0.75 was fair to good, and less than 0.4 was poor. ICC alone cannot provide complete reliability, and it requires confidence intervals (CI), despite appearing to be simple to interpret, and the level of disagreement between measurements is not indicated by ICC. Therefore, ICC and CI were used in conjunction with standard error of measurement (SEM), with a significance value of $P < 0.05$ (Rankin and Stokes 1998). SEM refers to the amount of variation in the results produced, and it can be calculated to determine absolute reliability. A lower SEM demonstrates good reliability (Baumgartner 2006), which allows clinicians and researchers to estimate of the extent of actual change in an outcome measure, as opposed to measurement error. SEM was obtained by using this equation: $SEM = SD_p*(\sqrt{1\text{-}ICC})$ (Harvill 1991). The SEM is then divided by the mean of all measurement and multiplied by 100 to give a percentage value of SEM (Lexell and Downham 2005). The minimum detectable change (MDC), the minimum change in a variable over time that is meaningful, is defined as ($1.96 *\sqrt{2} * SEM$), where SEM is the standard error of measurement; MDC is also expressed as a percentage: (minimum detectable change/mean of all observations) * 100 (Webber and Porter 2010; Weir 2005).

4.6.4 Results

Based on the use of three running trials, all variables were normally distributed (Kolmogorov-Smirnov $\geq$ 0.05), except hip-adduction moment in the second session ($p= 0.007$) and internal knee extensor moment in the first and second sessions (0.008 and 0.007 respectively). Although based on the use of three trials of SLS, all variables were normally distributed (Kolmogorov-Smirnov $\geq$ 0.05), except VGRF, in all sessions ($p < 0.05$).

Data collected from the running and SLS trials are shown in Tables 4.2 and 4.3. The within day ICC values show combined averages of (ICC_{SLS}=0.95& ICC_{run} = 0.97), and are shown to be higher when comparisons are made with between days (ICC_{SLS} = 0.88; & ICC_{run} = 0.90). Running tasks maintained average speeds of in running (3.02±0.29m/s). In all tasks, greater reliability was shown by VGRF data ($ICC_{SLS} \geq 0.92$ & $ICC_{run} \geq 0.97$)

when this was compared with angles ($ICC_{SLS} \geq 0.88$ & $ICC_{run} \geq 0.80$), and moments ($ICC_{SLS} \geq 0.83$ & $ICC_{run} \geq 0.88$) data. SEM values for joint angles within and between days showed a range of (0.75°-3.5°) during SLS, and (0.3°-2.3°) during running. MDC values for joint angles within and between days showed a range of (0.03°-6.45°) during SLS, and (0.10°-6.48°) during running All tasks revealed the highest SEM and MDC values for kinematic measures in running hip internal rotation angle (SEM = 2.34, MDC= 6.48).

Table 4.2: Within & between day ICC, Mean, and SEM values for 3D variables during running task

Variables		***Within-day***				***Between-days***			
		ICC (95% CI)	*Mean (°)*	*SEM(°)*	*MDC(°)*	*ICC (95% CI)*	*Mean (°)*	*SEM(°)*	*MDC(°)*
Joint Angles (°)	Hip Adduction	0.98 (0.96-0.99)	8.9	0.5	1.6	0.94 (0.88-0.97)	8.1	1.0	2.7
	Hip I. Rotation	0.97 (0.94-0.98)	5.2	1.0	2.9	0.88 (0.78-0.94)	5.9	2.3	6.48
	Knee Valgus	0.99 (0.98-0.99)	-2.0	0.3	0.9	0.80 (0.64-0.89)	-1.6	0.9	4.4
	Knee Flexion	0.99 (0.98-0.99)	44.1	0.7	2	0.91 (0.83-0.95)	44.2	1.8	5.1
Moments (Nm/Kg)	Hip Adduction	0.96 (0.92-0.98)	-1.7	0.1	0.2	0.97 (0.94-0.98)	-1.7	0.08	0.2
	Hip I. Rotation	0.97 (0.94-0.98)	-0.60	0.03	0.1	0.90 (0.81-0.95)	-0.62	0.08	0.2
	Knee valgus	0.96 (0.92-0.98)	0.23	0.05	0.1	0.88 (0.78-0.94)	0.20	0.08	0.2
	Knee Extensor	0.95 (0.90-0.97)	2.7	0.1	0.3	0.89 (0.79-0.94)	2.64	0.15	0.4
	VGRF (*Bw)	0.98 (0.98 -0.99)	2.56	0.05	0.1	0.97 (0.94-0.98)	2.57	0.06	0.1

(-) Knee valgus, Hip adduction moment, Hip internal rotation moment, Dorsiflexion moment. (SEM)=standard error of measurement. (Mean)= Mean Value for 2 sessions

Table 4.3:Within & between days ICC, Mean, SEM, & SEM% values for 3D variables during SLS task

Variables		***Within-day***				***Between-days***			
		ICC (95% CI)	*Mean (°)*	*SEM(°)*	*MDC(°)*	*ICC (95% CI)*	*Mean (°)*	*SEM (°)*	*MDC(°)*
Joint Angles (°)	Hip Adduction	0.98 (0.96-0.99)	13.09	0.71	1.9	0.89 (0.79-0.94)	13.07	1.50	4.2
	Hip Int. Rotation	0.98 (0.96-0.98)	10.16	1.14	3.1	0.92 (0.85-0.96)	10.80	2.32	6.45
	Knee Valgus	0.98 (0.96-0.99)	2.48	0.65	1.8	0.90 (0.81-0.95)	3.01	1.10	4.2
	Knee Flexion	0.94 (0.88-0.97)	86.65	1.78	4.9	0.88 (0.88-0.94)	88.34	3.30	9.1
Moments (Nm/Kg)	Hip Adduction	0.95 (0.90-0.97)	-0.94	0.04	0.1	0.90 (0.81-0.95)	- 0.97	0.07	0.1
	Hip Int. Rotation	0.86 (0.74-0.93)	-0.44	0.02	0.08	0.84 (0.70-0.92)	-0.44	0.06	0.1
	Knee valgus	0.99 (0.98-0.99)	0.11	0.03	0.08	0.83 (0.69-0.91)	0.09	0.03	0.3
	Knee Extensor	0.96 (0.92-0.98)	1.78	0.06	0.1	0.89 (0.79-0.94)	1.80	0.08	0.2
	VGRF (*Bw)	0.94 (0.88-0.94)	1.13	0.01	0.03	0.92(0.85-0.96)	1.13	0.01	0.04

(-) Knee valgus, Hip adduction moment, Hip internal rotation moment, Dorsiflexion moment. (SEM)=standard error of measurement. (Mean)= Mean Value for 2 sessions

4.7 Discussion

The objective of the study was to assess the within and between day reliability of biomechanical variables during running and SLS tasks in recreational athletes. The findings of this study show that for kinematic, kinetic and vertical GRF, the between day ICC values for both tasks were lower than for those of within day values. However, the findings in previous studies report similar values for landing (Ford et al., 2007) and for running (Ferber et al., 2002).

There was greater consistency of vertical GRF data, as this showed more consistency than joint angles and moments, which was expected, as gathering GRF data required no markers and was not influenced by marker placement error, and GRF data represents all segmental masses and accelerations (Winter 2009). Running knee valgus angle and SLS knee valgus moment showed the lowest ICC values for the nine study variables for between day, and a high SEM and MDC score; therefore, between day measures suggested significant subject differences. It appears that differences across the subjects were equally and randomly distributed, and the mean data was similar (2.7° vs. 2.67° and 0.11 Nm/kg vs. 0.09 Nm/kg respectively). These findings indicate that rather than making individual comparisons, comparing mean group data for repeated measures is more reliable.

Task difficulty, referenced static alignment and skin marker movement are factors that could influence between day and within day reliability (Ford et al., 2007; Ferber et al., 2002), and marker reapplication could contribute towards the variability of between day measures (Kadaba et al., 1989). One investigator applied the markers in this study during all of the trials, which has reduced between day values, although marker replacement differences could have affected the consistency in the measurements. This study attempted to overcome this reported variability and used the CAST marker based protocol (Cappozzo et al., 1995). This is because when compared with the modified Helen Hayes marker set, CAST provides improved anatomical relevance (Schwartz et al., 2004; Cereatti et al., 2007), as explained previously. Thus, markers were attached to the centre of segments instead of near joints, in an attempt to avoid skin movement artefact.

Both SEM and MDC reference values for running and SLS tasks have been provided in this study, which may be helpful for the evaluation of outcomes (Tables 4.2–4.3). Clinicians find SEM to be a useful tool when determining the improvement of individuals

(Munro et al., 2012; Domholdt 2005). It allows the clinician to be 68% confident that the true value lies within ±1 SEM of an observed value, since SEM calculations are dependent upon the standard deviation of measurements (Portney and Watkins 2009). Whilst MDC is based on SEM calculations, it is more conservative (2.7 SEMs), because when changes in the score are larger than the MDC, the difference is not caused by patient variability or error in measurement, offering a probability of 95% (Ries et al., 2009; Wilken et al., 2012).

During the SLS task, SEM values for peak knee-valgus angle ranged between 0.65° and 1.10° for within-day and between-days measures. This result indicates a 68% confidence as the true measures of participants were within a range of 3.01°, based on a one-week gap between repeated measures. When two measurements were taken on the same day, the range reduced to 2.48°. There was a 95% chance that the true value lay within 1.8° when both measures were taken on the same day, and 4.2° if the gap between measures was seven days. Lower values than the ones reported in the current study for knee-valgus angle during the same task (SEM=0.5-1.5°; MDC=1.3-3.7°, within-day and between-days, respectively) have been reported by Nakagawa et al., (2014). It may be that the participants in the study by Nakagawa et al., (2014) were younger than the participants in the current study (21±1.1 vs. 25±2.0 years), and that the interval between days was shorter than in the current study (3 vs. 7 days), which resulted in improved ICC values, together with lower SEM values.

The transverse-plane angles (hip-internal rotation), which is the third hypothesis of this study, revealed variability at high levels when compared to other planes. This was particularly relevant for between-days measurements of hip-internal rotation angles during the running task (ICC=0.88; SEM=2.3°; MDC=6.4°). Based on the interpretations used in this study, the ICC value was good (Coppieters et al., 2002), yet the highest MDC value occurred during running. The study by Noehren et al., (2010), which used a marker placement device, is the only attempt to improve between-days reliability, determining that the largest reduction in SEM values was in the transverse plane during the running task (reducing SEM to 57% and improving ICC by 7%). Possible areas for future research should concentrate on this issue, together with improving the reliability of hip-rotation measurements during the running task.

However, there are limitations to these findings, as they only apply to models and a laboratory setting, despite being consistent with those previously reported. The laboratory results could have been affected by an individual's ability to place markers. In addition, the depth of the squat needs to be considered, because while reflecting standard practice, there was insufficient control for each subject when squatting. Subjects were instructed to squat down on their right extremity as far as possible, and return to a single-legged standing position without losing their balance.

A further limitation identified within the current study is that participants wore standard trainers on a Mondo running surface. This limitation fails to replicate the interaction of a typical shoe-surface in real games using studded boots on grass, as well as trainers on AstroTurf. An uninjured population was examined, which is a further limitation to the study, but useful as screening tasks. Further investigation is required into the reliability of these functional tests in a population representing ACL injury, since this injury has been linked to excessive hip internal rotation and adduction, knee valgus and external rotation during different functional tasks (Hewett et al., 2004; Willson and Davis 2008).

4.8 Conclusion

This study concludes that for between and within day sessions, specific variables demonstrate good and excellent levels of consistency, and exhibit standard errors of measurement that have relatively low values. This is the first research study of its kind that has shown repeatability with the cluster model used. The data has also shown the repeatability of marker set and kinematic and kinetic data for three different sessions, which gives confidence in the data collection at multiple time points for individuals before and after ACL reconstruction, and can be taken forward into further studies.

CHAPTER FIVE

BIOMECHANICAL ANALYSIS OF FUNCTIONAL TASKS AFTER ANTERIOR CRUCIATE LIGAMENT RECONSTRUCTION (ACLR)

5.1 Introduction

After anterior cruciate ligament reconstruction (ACLR), the athlete is often anxious to return to their normal high-level activities as soon as possible. Concerning being ready to return to activity after ACLR, the literature has revealed that 60% of research reports use time from surgery to determine readiness, and six months post-surgery is the most common timeframe (Barber-Westin and Noyes 2011). With regard to a clinical perspective, the problem with only using time from reconstruction is that it does not take into account the patient's condition, even though this can vary greatly following ACLR (Daniel et al 1994). Alterations in movement patterns are likely to be the main factor in the development of PFJ and OA in the ACL-injured knee (Butler et al., 2009; Paterno et al., 2010; Webster and Feller, 2012; Culvenor et al., 2015). There is little data on why individuals with ACL-reconstructed (ACLR) knee consequently develop OA (Lohmander et al., 2007; Neuman et al., 2008), but altered kinematics and kinetics are one of the hypothesised reasons. Numerous studies investigating walking gait have found significant reductions in internal knee extensor moment and reductions in peak knee flexion angle, with individuals adopting a quadriceps avoidance gait pattern during walking (DeVita et al. 1998; Berchuck et al., 1990; Birac et al., 1991; Wexler et al., 1998). One of the most common activities pre- and post-surgery is running and it is not known if individuals with an ACLR knee have different knee kinematic and kinetic patterns between limbs.

One of the most important adaptions found in the ACLR group is a reduction in internal knee extensor moment during running. In studies conducted by Lewek et al., (2002) and Bush-Joseph et al., (2002), it was found that during the early stages of ACL reconstruction, an efficient stress prevention strategy for the new ligament might be a reduction in internal knee extensor moment during running. This strategy may also be useful in providing protection to the ligament from incurring further injuries, reducing/avoiding pain, and allowing for compensation that will occur following surgery. Other studies in the literature show that alternative strategies may have been used for these same purposes by the ACLR groups in those studies. These strategies involve reduction of flexion angles with a reduction in internal knee extensor moment (Berchuck et al., 1990). Therefore, this reduction in flexion angles is indicative of co-contraction of hamstrings with quadriceps muscles (Elias et al., 2015). The goal of strategies like these is to provide knee stabilisation (Andriacchi & Dyrby 2005; Alkjaer et al., 2003; Von Porat et al., 2006; Knoll, et al., 2004). A clear understanding of differences is of utmost importance, since the strategy may increase of the development of degenerative diseases.

There is minimal literature with methodological limitations on biomechanical differences between ACLR groups and control groups during running. As well as using small sample sizes and using a treadmill to run, no studies have been found that have measured hip kinematics and kinetics in frontal and transverse plane after ACLR. However, in general clinical practice, the option to actually ask the individual to run and assess this is limited, and therefore a more space-optimised clinical assessment is needed. Therefore, the single leg squat (SLS) has been chosen as the measure to assess these individuals.

Whatman et al., (2011) found correlations between running angles and SLS angles at the hip and knee joints. Yamazaki et al., (2009) compared the kinematics of SLS between ACL injured individuals and control group individuals. When comparing the injured and uninjured legs within participants, the injured leg of both male and female individuals showed greater knee adduction angle than the uninjured leg. The injured leg of the male individuals showed less knee and hip external rotation angles, less knee flexion and more knee adduction angle than those of their uninjured leg. However, no studies have been found that have examined kinematics and kinetics after ACLR during a SLS task. Therefore, the purpose of this study was to compare the hip and knee kinematics and kinetics of ACLR individuals six to eight months post-surgery with a control group, during running and SLS.

5.2 Objectives and Null Hypotheses

1. To investigate the hip joint frontal and transverse planes movements six to eight months after ACL reconstruction during running between limbs (injured and non-injured).

   Null Hypothesis: There will be no differences in hip joint frontal and transverse planes movements six to eight months after ACL reconstruction during running between limbs (injured and non-injured).
2. To investigate the hip joint frontal and transverse planes movements six to eight months after ACL reconstruction during SLS between limbs (injured and non-injured).

   Null Hypothesis: There will be no differences in hip joint frontal and transverse planes movements six to eight months after ACL reconstruction during SLS between limbs (injured and non-injured).

3. To compare the hip joint frontal and transverse plane movements six to eight months after ACL reconstruction with a healthy control group during running.

   Null Hypothesis: There will be no differences in hip joint frontal and transverse plane movements between individuals six to eight months after ACL reconstruction and healthy control group during running.

4. To compare the hip joint frontal and transverse plane movements six to eight months after ACL reconstruction with a healthy control group during SLS.

   Null Hypothesis: There will be no differences in hip joint frontal and transverse plane movements between individuals six to eight months after ACL reconstruction and healthy control group during SLS.

5. To investigate the knee joint sagittal and frontal planes movements six to eight months after ACL reconstruction during running between limbs (injured and non-injured).

   Null Hypothesis: There will be no differences in knee joint sagittal and frontal planes movements six to eight months after ACL reconstruction during running between limbs (injured and non-injured).

6. To investigate the knee joint sagittal and frontal planes movements six to eight months after ACL reconstruction during SLS between limbs (injured and non-injured).

   Null Hypothesis: There will be no differences in knee joint sagittal and frontal planes movements six to eight months after ACL reconstruction during SLS between limbs (injured and non-injured).

7. To compare the knee joint sagittal and frontal planes movements six to eight months after ACL reconstruction with a healthy control group during running.

   Null Hypothesis: There will be no differences in knee joint sagittal and frontal planes movements between individuals six to eight months after ACL reconstruction individuals and healthy control group during running.

8. To compare the knee joint sagittal and frontal planes movements six to eight months after ACL reconstruction with healthy control group during SLS.

   Null Hypothesis: There will be no differences in knee joint sagittal and frontal planes movements between individuals six to eight months after ACL reconstruction individuals and healthy control group during SLS.

5.3 Methods
5.3.1 Research Environment

The running gait and SLS analysis work was all completed at the Human Performance Laboratory (situated in the Mary Seacole Building, University of Salford), which has a strong record in musculoskeletal research and clinical gait analysis.

5.3.2 Participants

Thirty-four ACLR elite athletes (footballers, rugby plyers and taekwondo competitors) who had all undergone ACL reconstructive surgery (6-8 months since surgery; type of reconstruction: 20 hamstring grafts, 14 bone-patellar tendon bone grafts). The College of Health Sciences Research Governance and Ethical Committee approved the study (HSCR13/74). Additionally, this study received a favourable opinion from the NHS ethical committee (NHS rec number 14/LO/0255) (see Appendix 1). The recruitment was initially from orthopaedic consultants at Stepping Hill Hospital, Stockport, or private hospitals. Participant Identification Centres (PICs) were arranged at this trust, and invitation letters (see Appendix 6) were made available to give to patients, along with the participant information sheet (see Appendix 4). Individuals were requested to return the data access forms in the stamped addressed envelope provided to indicate they were interested in the study. Once the form was received, the principal investigator would contact the individual to check their eligibility. If the individual was eligible and still interested in the study, an appointment was made for them at the Human Performance laboratory at the University of Salford, as illustrated in Figure 5.1. All participants completed similar post-operative rehabilitation, including early weight-bearing, range of movement and neuromuscular retraining, and a graduated return to sport.

Thirty-four subjects that made up the control group were highly active; all of them are students recruited from the university, and they volunteered for the study. These participants had no history of lower extremity surgery, and had not suffered any lower extremity injuries during the previous six months. The definition of injury was all musculoskeletal complaints that had prevented engagement in normal exercise routines for participants. Ethical approval was obtained (HSCR12/64) and all participants were required to read and sign the informed consent statement before testing.

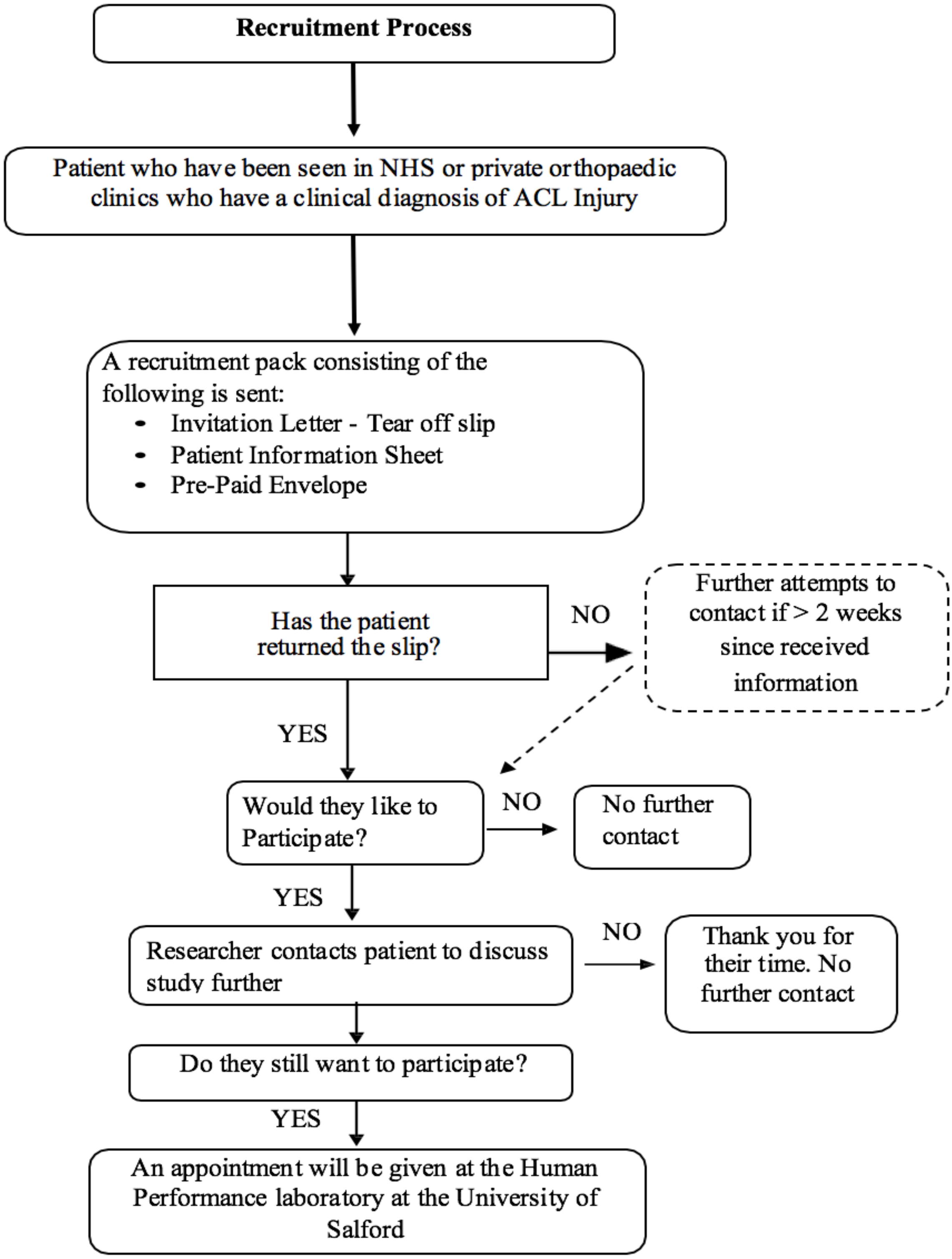


Figure 5.1: Flow chart of Recruitment Process.

5.3.3 Inclusion criteria:

To be eligible for the study, participants were required to meet the following criteria:

- Aged between 18 to 40 years old
- Adult six to eight months post ACLR with a hamstring graft and patellar tendon graft, and with or without an accompanying meniscal tear

5.3.4 Exclusion criteria:

Individuals who have any of the following conditions were excluded from the study:

- Those who were unable to give informed consent or comply with the study procedures
- Subjects with cardiovascular, pulmonary or neurological conditions that limit physical activity
- Subjects with any lower limb, pelvic or spinal pathology that limits the ability to run comfortably for five minutes
- Those who do not run regularly

5.3.5 Sample size:

G*Power software was used to calculate sample size. This software was used to estimate the sample size by using the difference between the means and standard deviation for six ACLR and six control group participants ((mean± SD)2.60±0.82 and 3.24±0.69 respectively) during running to calculate the effect size. Statistical significance was deemed to be α=0.05, power of 90% and effect size =0.84 for internal knee extensor moment. The calculation results indicated that 31 subjects were required for the ACLR group, and another 31 subjects for the control group, in the sample in order to measure the differences.

5.3.6 Procedures

Once subjects demonstrated that they were interested in the study, they were given an appointment at the Human Performance Laboratory at the University of Salford. When participants attended the Human Performance laboratory for their initial visit, they were briefed on the study and had all the equipment and procedures explained to them. Any questions were answered in full, and if happy, they were asked to sign the informed consent form (see Appendix 5). Participants also completed the Knee Injury and Osteoarthritis Outcome Score (KOOS) questionnaire (Appendix 2), which measures subjective outcomes (symptoms, stiffness, pain, daily living, sports and recreational activities function, and quality of life) to determine the outcomes after ACL injury. The International Knee Documentation Committee Scale (IKDC) evaluation scales (Appendix 3) were performed six to eight months after ACL reconstruction for the ACLR group; they were also used with the control group one time before the session. A demographics form was also completed (age, height and weight) for each of the participants. A twelve-camera motion analysis system (Pro-Reflex, Qualisys, Sweden), with sampling at 240 Hz, was used to collect kinematic data. A force platform was embedded into the floor (AMTI, USA), and sampling at 1200 Hz was used to collect kinetic data. As previously outlined in the methodology chapter (Chapter 4, Sections 4.1-3), the same instrumentation, calibration, training shoes, filtration, biomechanical model and marker list were deployed. Sections 4.3.1-2 have already described the running and SLS tasks.

5.3.7 Outcome measures

These discrete variables were selected based on their frequency in relation to possible ACL, PFPS and OA injury studies, as discussed in Chapter One, alongside the clinical questionnaire results. The following discrete variables were calculated during the early stance (0-50) phase for each trial:

a) Peaks of hip adduction angle and moment
b) Peaks of hip internal rotation angle and moment
c) Peak knee flexion angle
d) Peak internal knee extensor moment
e) Knee extensor impulse
f) Peak knee valgus angle
g) Peak external knee adduction moment
h) Peak vertical ground-reaction force (VGRF)

5.3.8 Data analysis

The descriptive analysis (mean and standard deviation) covers each dependent variable in the target tasks (running and SLS). A Kolmogorov-Smirnov test was used to check whether data was normally distributed or not (parametric or non-parametric). A comparison between the injured limb and non-injured limb for ACLR was performed using a paired t-test for parametric variables and a Wilcoxon Rank Test for non-parametric variables. A comparison between the injured limb and the non-injured limb for ACLR was made against the control group using an independent t-test for parametric variables, and a Mann-Whitney U test for non-parametric variables, as shown in Figure 5.2. The p-value was set at p=0.05 or less to be statistically significant. All statistical analyses were performed using SPSS (v. 20, SPSS Inc., USA).

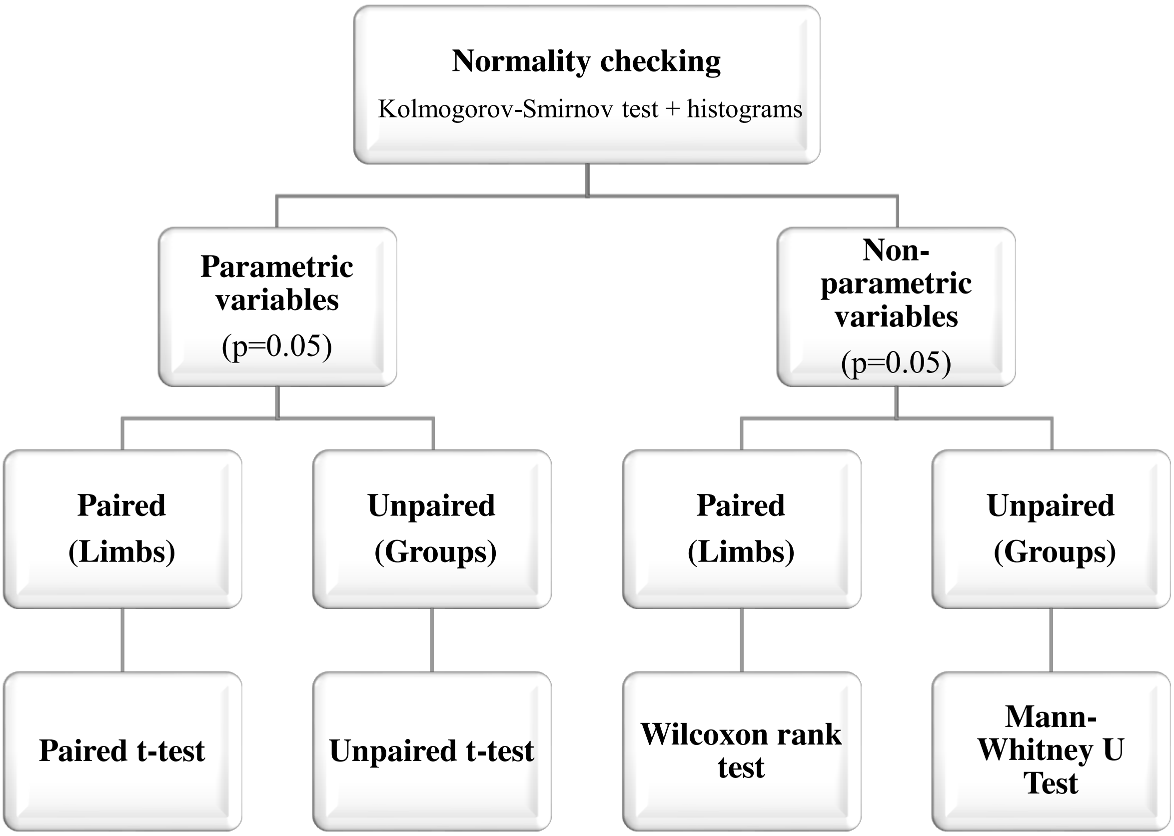


Figure 5.2: Statistical analysis outline for the return to sport study

5.3.9 Results

5.3.9.1 Patient Reported Function: The IKDC and KOOS:

IKDC, KOOS score data was collected for the ACLR group, but not distributed (Kolmogorov-Smirnov, IKDC p=0.01 and KOOS p=0.01). An independent t-test for clinical scores was deployed for the comparisons between the two groups, with group differences deemed significant at a level of p=0.05.

Table 5-1: IKDC and KOOS scores.

	Participant Group		
	ACLR	Control	*P* value
KOOS	88.40±10.09	96.15±6.06	*0.01
IKDC	88.98±11.41	94.39±9.52	*0.01

Key: ACLR=ACL reconstructed group, Control=Healthy group. *Significant differences at a level of p =0.05. *Means±S.D.*

As shown in Table 5.1, a significant difference was found in the KOOS and IKDC (p=0.01 and p=0.01 respectively) knee function scores between the participant groups. The ACLR group demonstrated significantly lower levels of subjective knee function in comparison to the control group.

Table 5-2: KOOS Subscales Scores

	KOOS				
	Symptoms	**Pain**	**Function ADL**	**Function SRA**	**Quality of Life (QoL)**
ACLR	85.5±11.8	93.1±8.7	98.4±4.9	87.7±14.3	77.8±18.6
Control	92.5±7.0	96.5±6.9	99.1±2.6	94.2±11.4	89.9±17.7
P-value	0.05*	0.17	0.55	0.01*	0.01*

*Significant differences at a level of p=0.05. *Means±S.D.*

As shown in Table 5.2 for the KOOS-Subscales, a significant difference was found in the KOOS-symptoms, KOOS-function SRA and KOOS-QoL (p=0.05, p=0.01 and p=0.01 respectively) for knee function scores between the participant groups. The ACLR group revealed significantly lower levels of subjective knee function in comparison to the control group.

5.3.10 Analysis of Running

Kinematic and kinetic outcomes, together with running performance measures, were analysed for 34 control group individuals and 34 ACLR patients. These results show the knee adduction angle rather than the knee valgus angle, as there is no valgus during the early stance (0-50) of the stance phase. Table 5.3 shows the differences between the groups for the results of the t-test and demographic parameters, in terms of standard deviations and mean deviations.

Table 5.3: Subjects demographic data.

Variable	Groups		*P* value
	Control (n=34) *(Means ± S.D)*	ACLR (n=34) *(Means ± S.D)*	
Age (years)	23.7±3.6	21.8±3.9	0.05*
Height (m)	1.7±0.1	1.7±0.1	0.32
Body mass (kg)	73.5±11.1	79..9±16.5	0.08
Male/Female	24M/10F	24M/10F	NA
Running Speed (m/s)	3.5±0.58	3.5±0.57	0.69
Activity Level (KOOS SRA)	94.1±11.4	87.6±14.3	0.01*

Key: m=metres, kg=Kilogram. m/s=Speed in metres per second. * Significant differences at p=0.05.

As shown in Table 5.3, there are no significant differences between groups for all demographic data, except for age (p=0.05). Activity levels demonstrated a significant reduction in the ACLR group compared to the control group (p=0.01). The running speeds were not significantly different between the two groups.

5.3.10.1 Hip Internal Rotation Angle

The hip internal rotation angle demonstrated a significant increase in the injured limb compared to the non-injured limb for the ACLR group (*p*=0.01; effect size 0.53). When comparing the injured limb and non-injured limb for the ACLR group against the control group, no significant differences were found (*p*=0.08 and p= 0.54 respectively). Descriptive data on hip internal rotation angle after ACLR during running are shown in Figure 5.3 and Table 5.4.

Table 5.4: Hip Internal Rotation Angle after ACLR

Stance	Injured	Non	Injured	Control	Non	Control
Mean (°)	7.14	3.43	7.14	4.74	3.43	4.74
SD	6.94	7.04	6.94	6.26	7.04	6.26
P-value(t-test)	0.01*		0.08		0.54	

*Key: °=degrees. * significant differences at p=0.05*

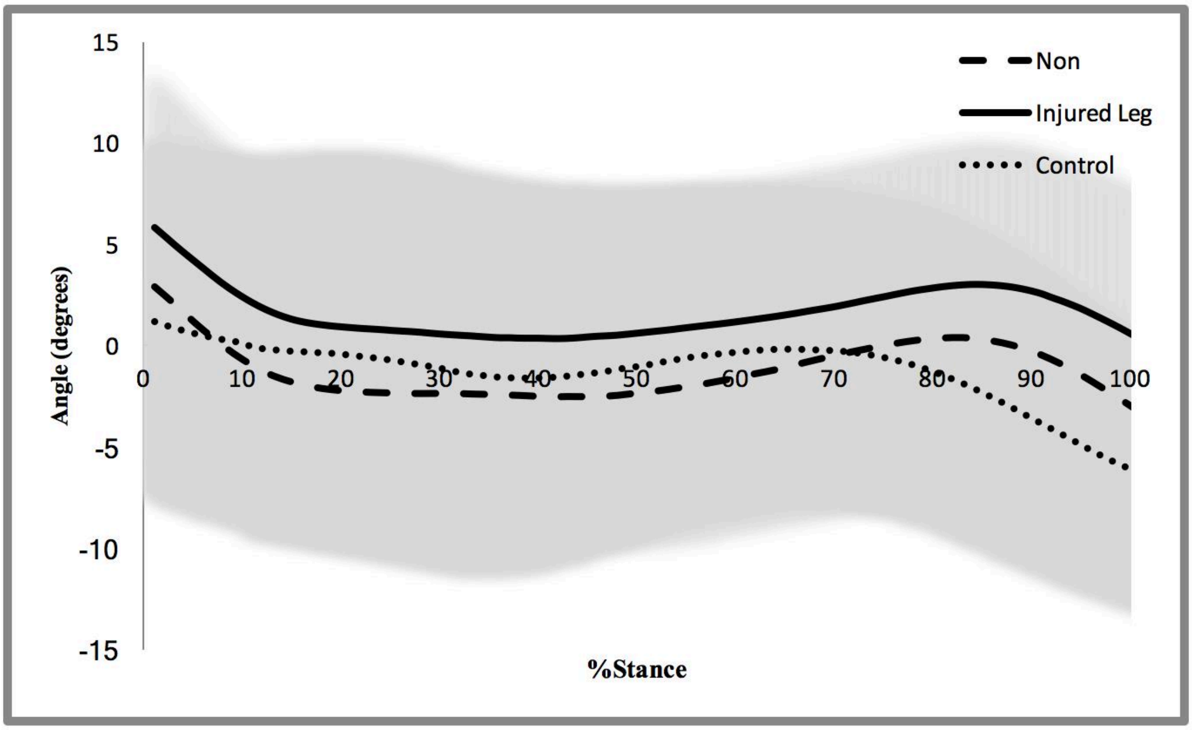


Figure 5.3: (+) Hip Internal Rotation Angle (Means ± 1SD) after ACLR during Running

5.3.10.2 Hip Internal Rotation Moment

The hip internal rotation moment demonstrated no significant difference between limbs for the ACLR group (p=0.08). When comparing the injured limb and non-injured limb for the ACLR group against the control group, no significant differences were found (p=0.55 and p= 0.29 respectively). Descriptive data on hip internal rotation moment after ACLR during running are shown in Figure 5.4 and Table 5.5.

Table 5.5: Hip Internal Rotation Moment after ACLR

Stance	Injured	Non	Injured	Control	Non	Control
Mean (Nm/Kg)	-0.73	-0.77	-0.73	-0.72	-0.77	-0.72
SD	0.21	0.27	0.21	0.22	0.27	0.22
P-value	0.08		0.55		0.29	

Key: N.m/kg=Normalised knee moment

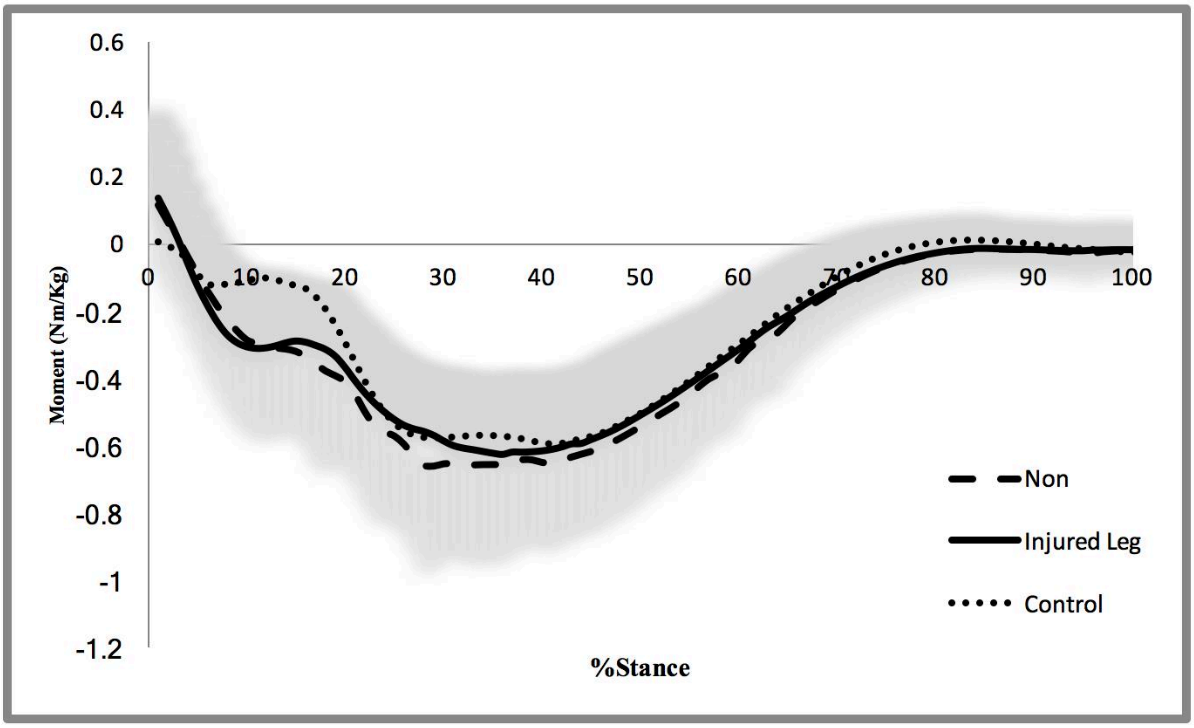


Figure 5.4: Hip Internal Rotation Moment (Means ± 1SD) after ACLR during Running

5.3.10.3 Hip Adduction Angle

The hip adduction angle demonstrated no significant difference between limbs for the ACLR group (*p*=0.30). When comparing the injured limb and non-injured limb for the ACLR group against the control group, no significant differences were found (*p*=0.66 and p= 0.11 respectively). Descriptive data on hip adduction angle after ACLR during running are shown in Figure 5.5 and Table 5.6.

Table 5.6: Hip Adduction Angle After ACLR

Stance	Injured	Non	Injured	Control	Non	Control
Mean (°)	11.15	10.16	11.15	11.96	10.16	11.96
SD	5.59	4.44	5.59	4.73	4.44	4.73
P-value	0.30		0.66		0.11	

Key: °=degrees

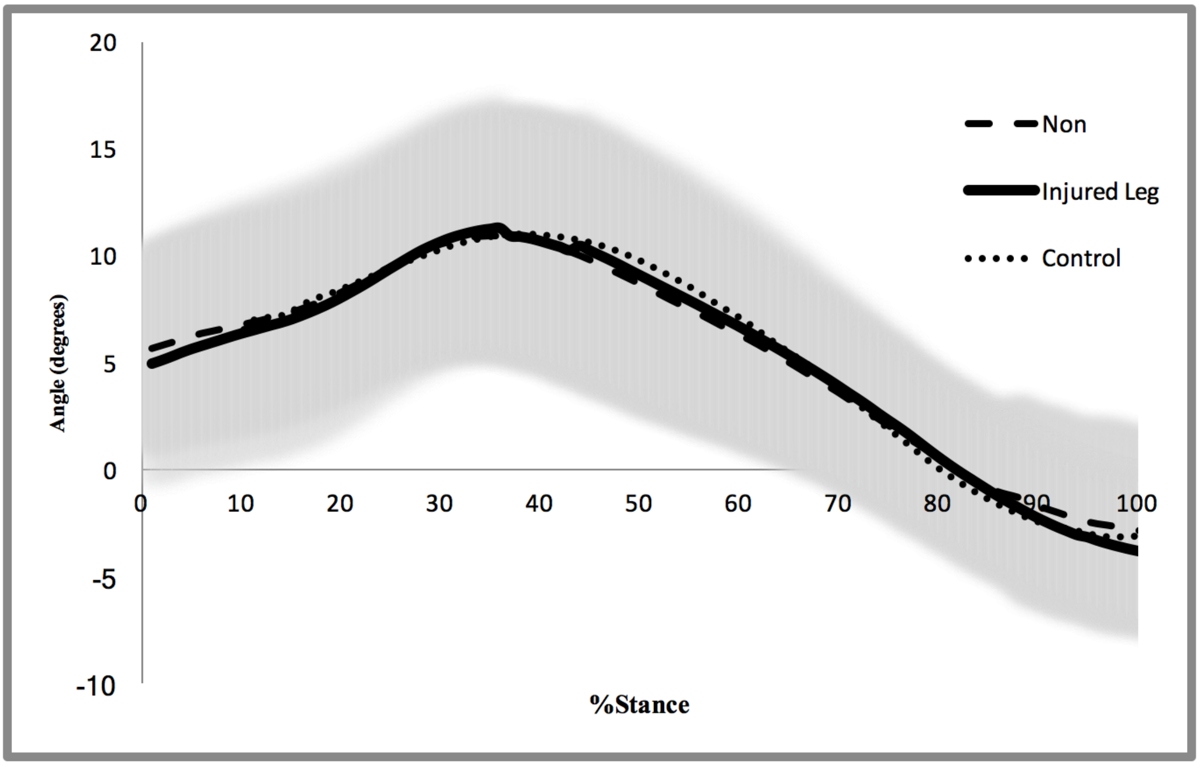


Figure 5.5: (+) Hip Adduction Angle (Means ±1SD) After ACLR during Running

5.3.10.4 Hip Adduction Moment

The hip adduction moment demonstrated no significant difference between limbs for the ACLR group (p=0.30). When comparing the injured limb and non-injured limb for the ACLR group against the control group, no significant differences were found (p=0.58 and p= 0.30 respectively). Descriptive data on hip adduction moment after ACLR during running are shown in Figure 5.6 and Table 5.7.

Table 5.7: Hip Adduction Moment After ACLR

Stance	Injured	Non	Injured	Control	Non	Control
Mean (Nm/Kg)	-2.00	-2.07	-2.00	-1.89	-2.07	-1.89
SD	0.58	0.55	0.58	0.54	0.55	0.54
P-value	0.30		0.58		0.30	

Key: N.m/kg=Normalised knee moment

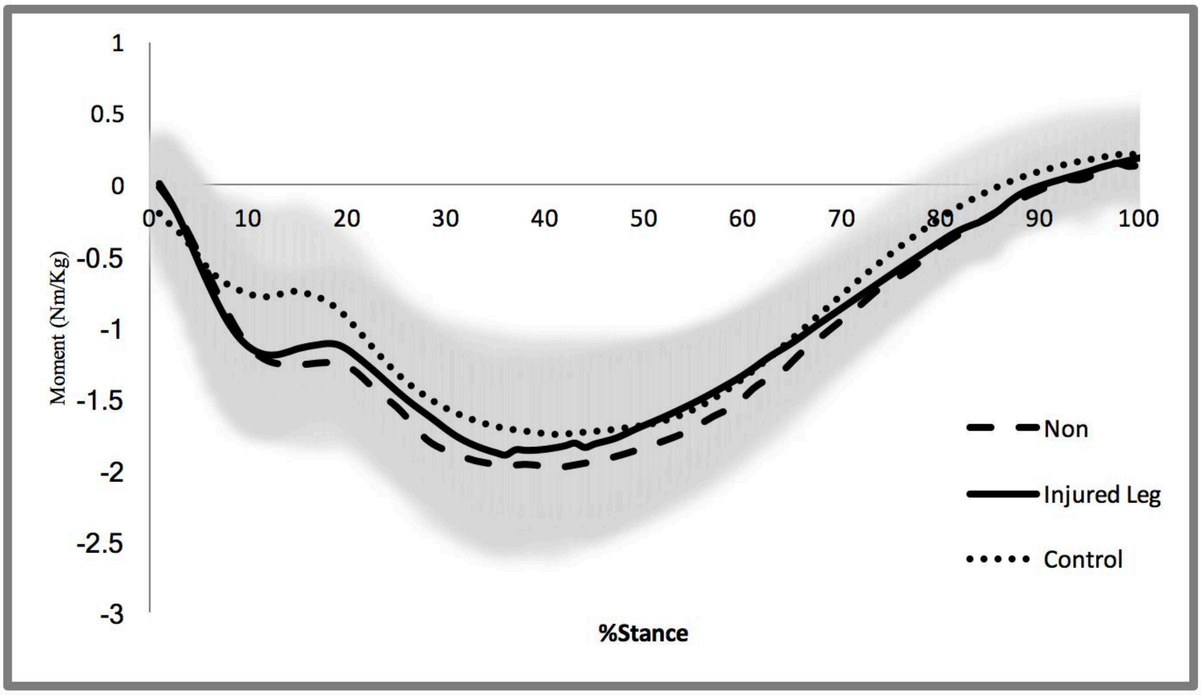


Figure 5.6: (-) Hip Adduction Moment (Means ± 1SD) After ACLR during Running

5.3.10.5 Knee Flexion Angle

The peak knee flexion angle demonstrated a significant reduction in the injured limb compared to the non-injured for the ACLR group (p=0.01; effect size 0.63). When comparing the injured limb for the ACLR group against the control group, a significant reduction was found in the injured limb (p=0.01; effect size 0.74). When comparing the non-injured limb for the ACLR group against the control group, no significant differences were found (p=0.62). Descriptive data on peak knee flexion angle after ACLR during running are shown in Figure 5.7 and Table 5.8.

Table 5.8: Peak Knee Flexion Angle after ACLR

Stance	Injured	Non	Injured	Control	Non	Control
Mean (°)	44.94	48.95	44.94	50.61	48.95	50.61
SD	6.18	6.40	6.18	8.83	6.40	8.83
P-value(t-test)	**0.01***		**0.01***		0.62	

*Key: °=degrees. * significant differences at p=0.05*

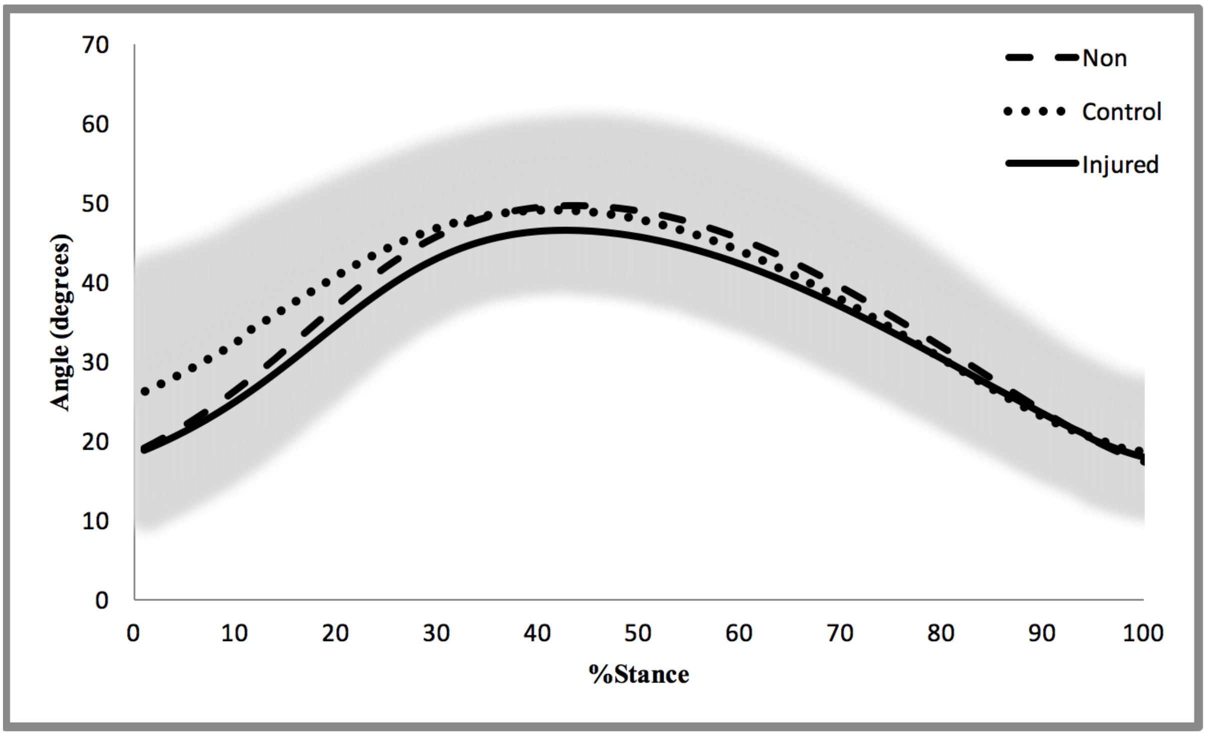


Figure 5.7: (+) Knee Flexion Angle (Means ± 1SD) after ACLR during Running

5.3.10.6 Internal Knee Extensor Moment

The internal knee extensor moment demonstrated a significant reduction in the injured limb compared to the non-injured limb for the ACLR group (*p*=0.01; effect size 0.77). When comparing the injured limb for the ACLR group against the control group, significant reductions were found in the injured limb (p=0.01; effect size 1.27). When comparing the non-injured limb for the ACLR group against the control group, no significant differences were found (*p*=0.17). Descriptive data on internal knee extensor moment after ACLR during running are shown in Figure 5.8 and Table 5.9.

Table 5.9: Internal knee Extensor Moment after ACLR

Stance	Injured	Non	Injured	Control	Non	Control
Mean (Nm/Kg)	2.80	3.32	2.80	3.59	3.32	3.59
SD	0.63	0.70	0.63	0.61	0.70	0.61
P-value	0.01*		0.01*		0.17	

*Key: N.m/kg=Normalised knee moment. * significant differences at p=0.05*

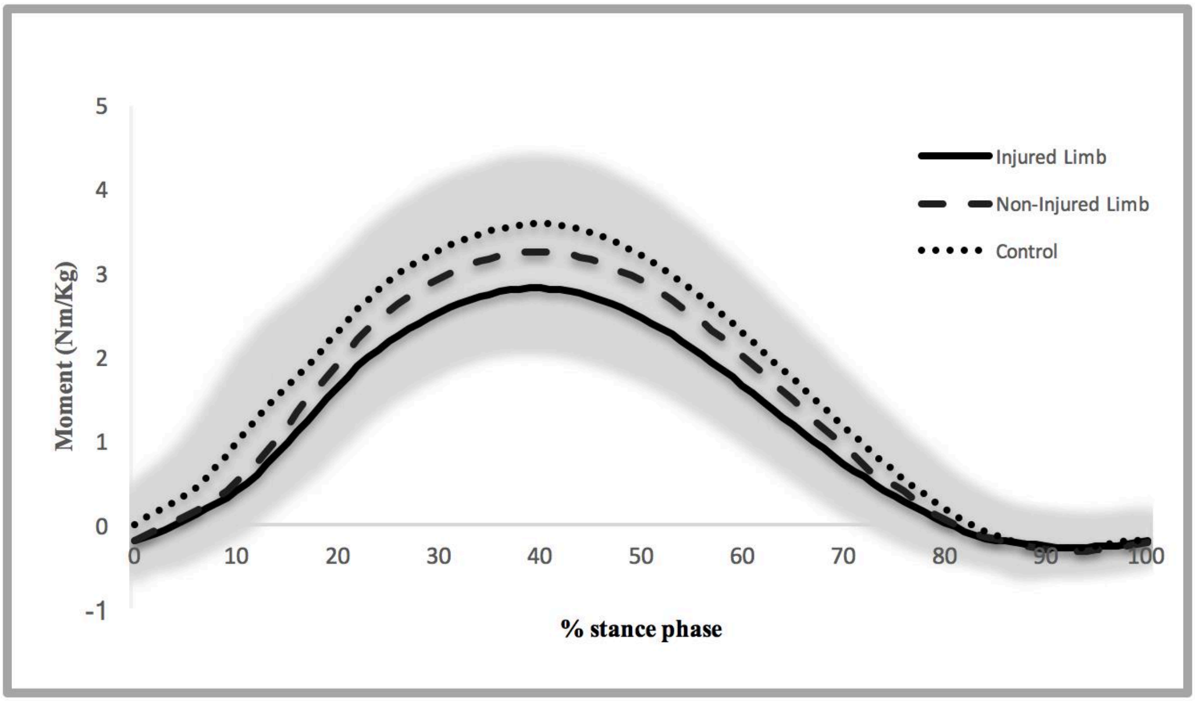


Figure 5.8:(+) Internal Knee Extensor Moment (Means ± 1SD) after ACLR during Running

5.3.10.7 Knee Extensor Impulse

The knee extensor impulse demonstrated a significant reduction in the injured limb compared to the non-injured limb for the ACLR group (*p*=0.01; effect size 1.14). When comparing the injured limb for the ACLR group against the control group, significant reductions were found in the injured limb (p=0.01; effect size 1.22). When comparing the non-injured limb for the ACLR group against the control group, no significant differences were found (*p*=0.11). Descriptive data on knee extensor impulse six months post ACLR during running are shown in Figure 5.9 and Table 5.10.

Table 5.10: Knee Extensor Impulse after ACLR

Stance	Injured	Non	Injured	Control	Non	Control
Mean (Nm/Kg*s)	0.24	0.32	0.24	0.36	0.32	0.36
SD	0.07	0.07	0.07	0.12	0.07	0.12
P-value	0.01*		0.01*		0.14	

Key: N.m/kg*s=Normalised knee impulse. * significant differences at p=0.05

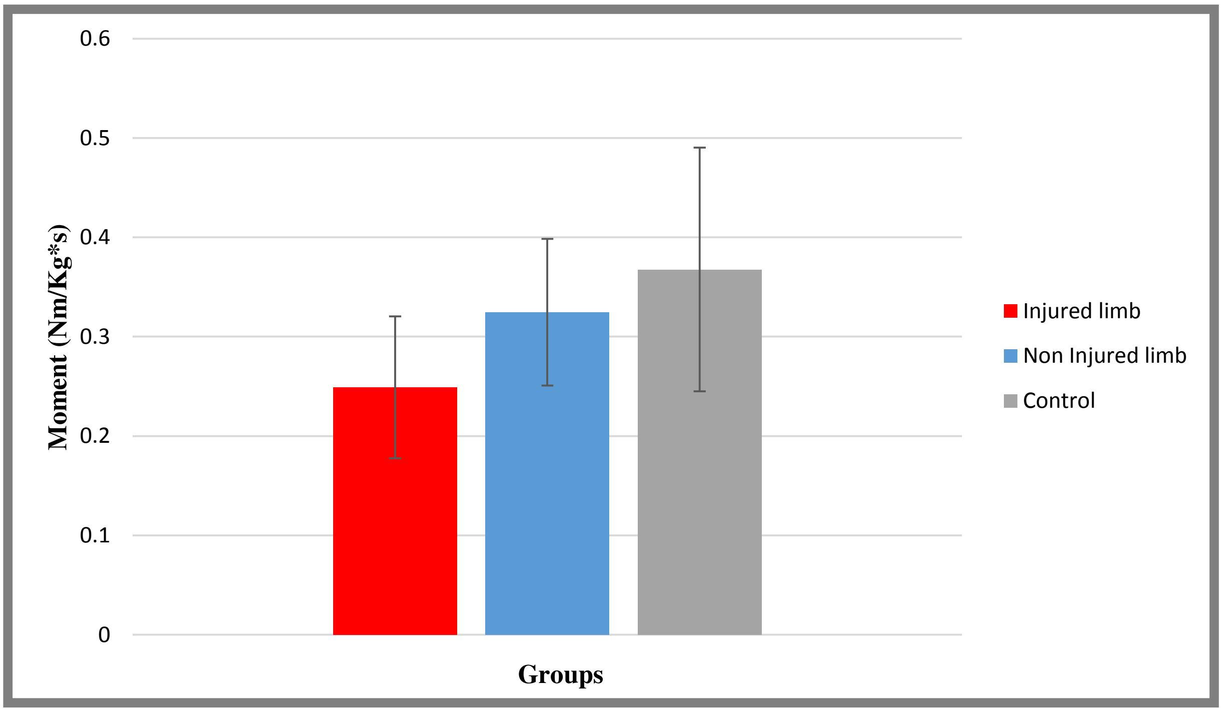


Figure 5.9: Knee Extensor Impulse (Means ± SD) after ACLR during Running

5.3.10.8 Knee Adduction angle

Knee adduction angle demonstrated no significant difference between limbs for the ACLR group, and when comparing the injured limb and non-injured limb for the ACLR group against the control group (*p*=0.28 and p= 0.80 respectively). When comparing the non-injured limb for the ACLR group against the control group, no significant differences were found (*p*=0.23). Descriptive data on knee adduction angle after ACLR during running are shown in Figure 5.10 and Table 5.11.

Table 5.11: Knee adduction angle after ACLR

Stance	Injured	Non	Injured	Control	Non	Control
Mean (°)	5.82	4.85	5.82	6.11	4.85	6.11
SD	5.01	4.48	5.01	4.20	4.48	4.20
P-value (t-test)	0.28		0.80		0.23	

*Key: °=degrees. * significant differences at p=0.05*

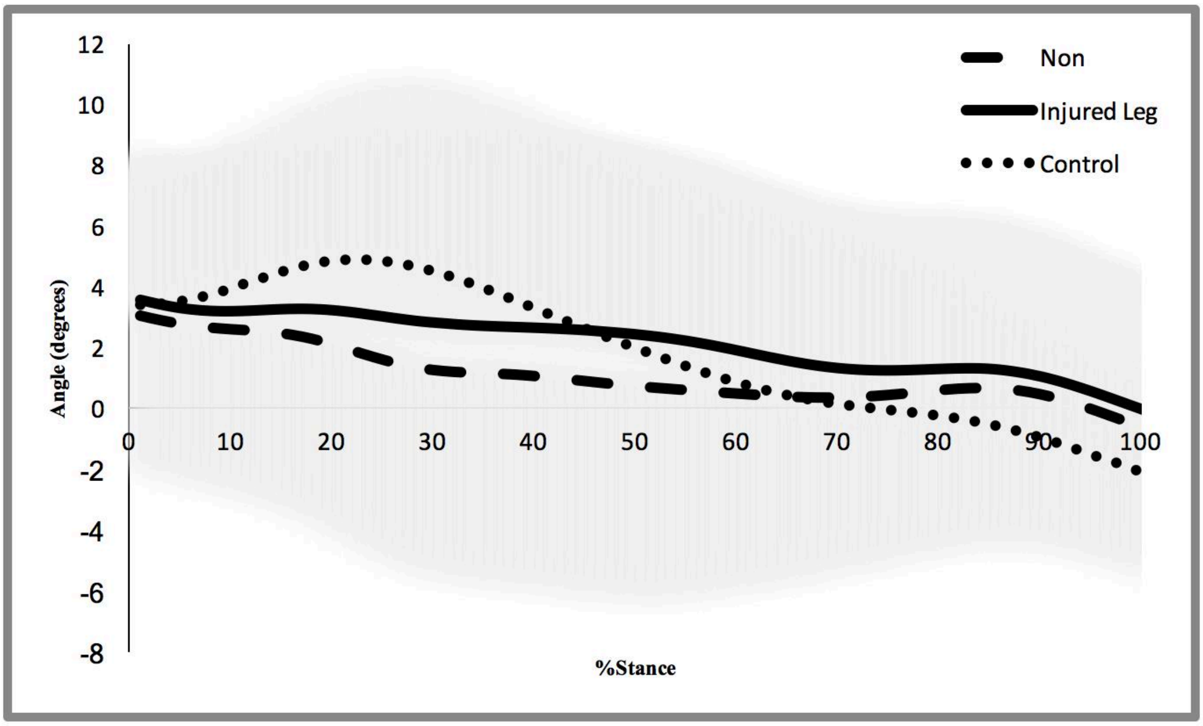


Figure 5.10: (+) Knee Valgus Angle (Means ± 1SD) After ACLR during Running

5.3.10.9 External Knee Adduction Moment

The knee adduction moment demonstrated no significant difference between limbs for the ACLR group (*p*=0.36). When comparing the injured limb for the ACLR group against the control group, significant increases for the ACLR group were found (*p*=0.04 effect size 0.66). When comparing the non-injured limb for the ACLR group against the control group no significant differences (p=0.24) were found. Descriptive data on external knee adduction moment after ACLR during running are shown in Figure 5.11 and Table 5.12.

Table 5.12: External knee Adduction Moment after ACLR.

Stance	Injured	Non	Injured	Control	Non	Control
Mean (Nm/Kg)	0.88	0.79	0.88	0.69	0.79	0.69
SD	0.37	0.36	0.37	0.36	0.36	0.36
P-value	0.36		0.04*		0.24	

*Key: N.m/kg=Normalised knee moment. * significant differences at p=0.05*

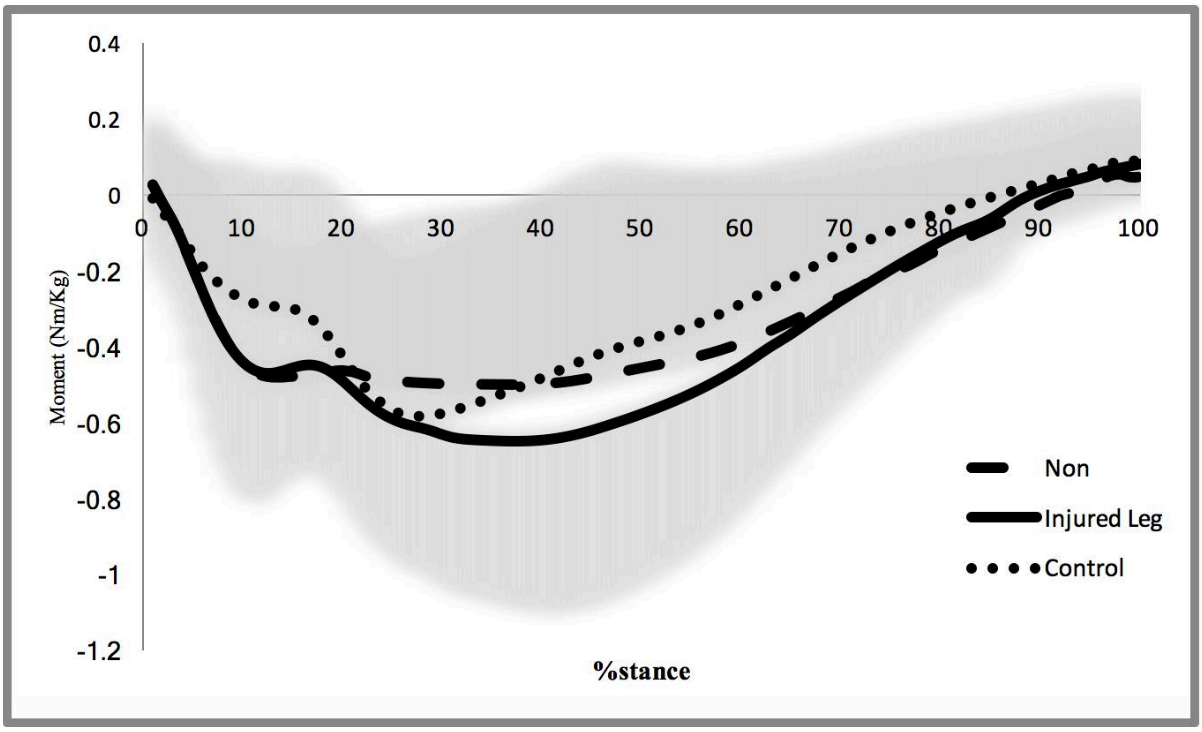


Figure 5.11: (+) External Knee Adduction Moment (Means ± 1SD) after ACLR during Running

5.3.10.10 Vertical Ground Reaction Force

The vertical ground reaction force demonstrated no significant difference between limbs for the ACLR group (p=0.15). When comparing the injured limb and non-injured limb for the ACLR group against the control group, no significant differences were found (p=0.08 and p= 0.31 respectively). Descriptive data on vertical ground reaction force after ACLR during running are shown in Figure 5.12 and Table 5.13.

Table 5.13: Vertical Ground Reaction Force After ACLR.

Stance	Injured	Non	Injured	Control	Non	Control
Mean (*BW)	2.57	2.62	2.57	2.67	2.62	2.67
SD	0.22	0.25	0.22	0.41	0.25	0.41
P-value	0.15		0.08		0.31	

Key: BW= Body Weight

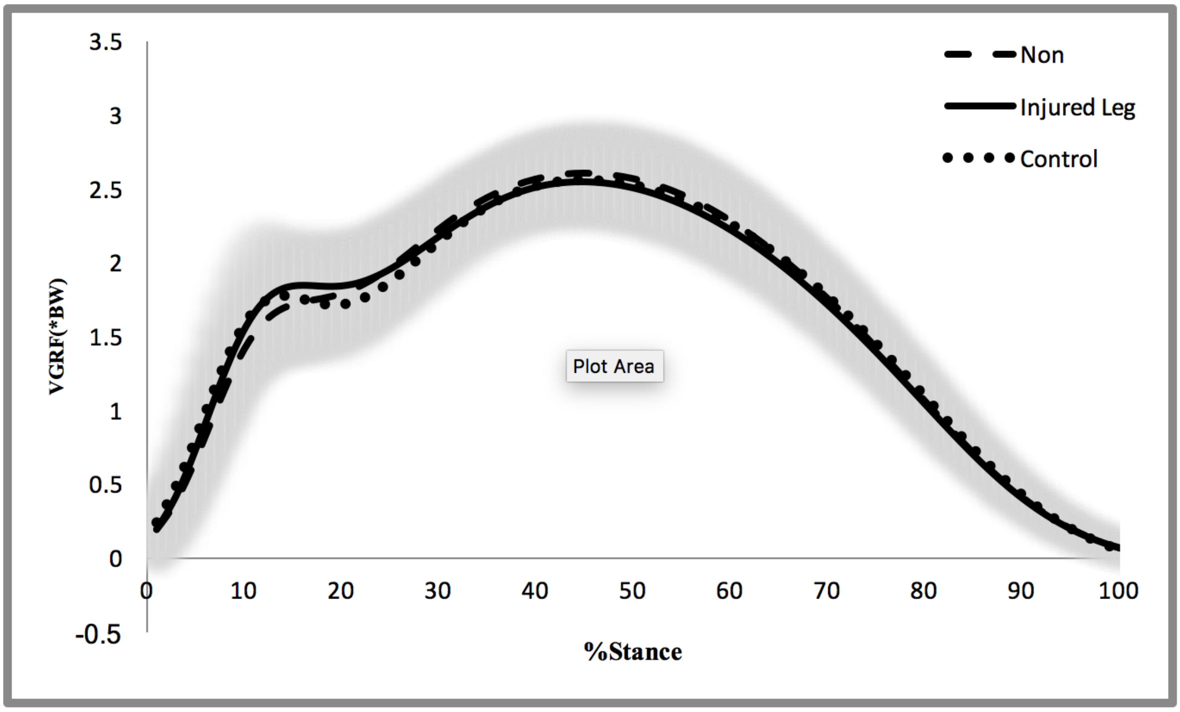


Figure 5.12: (+) Vertical Ground Reaction Force (Means ± 1SD) After ACLR during Running

5.3.11 Analysis of Single Leg Squat (SLS)

All hip and knee kinematics and kinetics were normally distributed, apart from hip internal rotation moment (p=0.01), which were found to be not normally distributed for all participants.

5.3.11.1 Kinematic and kinetic differences in injured and non-injured limbs after ACLR during SLS:

The results for kinematic and kinetics between limbs differences after ACLR, are shown in Table 5.14. The hip internal rotation angle and knee adduction angle demonstrated a significant increase in the injured limb compared to the non-injured limb for the ACLR group (*p*=0.01; effect size 0.68, *p*=0.01; effect size 0.36 respectively). The internal knee extensor moment demonstrated a significant reduction in the injured limb compared to the non-injured limb for the ACLR group (*p*=0.04; effect size 0.42). For the other kinematic and kinetic variables, the results were not significantly different between limbs for ACLR patients.

Table 5.14: Kinematics and kinetics between injured and non-injured limbs differences after ACLR

Variable	Limbs		*P* value
	Injured limb (n=34) *(Means ± S.D)*	Non-Injured limb (n=34) *(Means ± S.D)*	
Hip Internal Rotation Angle (°)	10.39±5.57	6.60±4.98	*0.01
Hip Internal Rotation Moment (Nm/Kg)	-0.46± 0.14	-0.51±0.12	0.42
Hip Adduction Angle (°)	13.28±5.07	11.72±4.96	0.17
Hip Adduction Moment (Nm/Kg)	-0.93±0.21	-0.92±0.22	0.82
Knee Adduction Angle (°)	14.57±6.69	11.16±6.59	0.01*
Knee Adduction Moment (Nm/Kg)	-0.50±0.15	-0.46±0.17	0.29
Knee Flexion Angle (°)	82.44±12.13	83.62±12.61	0.48
Knee Extensor Moment (Nm/Kg)	1.61±0.29	1.71±0.23	*0.04
VGRF (*BW)	1.10±0.03	1.11±0.04	0.08

*Key: °=degrees and N.m/kg=Normalised knee moment. * significant differences at p=0.05*

5.3.11.2 Kinematic and kinetic differences between injured limbs and the control group during SLS:

The results on the kinematic and kinetic differences between the injured limb and the control group after ACLR during SLS are shown in Table 5.15. The hip adduction angle demonstrated a significant increase in the injured limb for the ACLR group (*p*=0.05; effect size 0.68) when comparing to the control group. The internal knee extensor moment demonstrated a significant reduction in the injured limb for the ACLR group (*p*=0.01; effect size 1.10) when comparing the injured limb for the ACLR group against the control group. For the other kinematic and kinetic variables, the results were not significantly different between the control group and injured limb of ACLR patients.

Table 5.15: Kinematics and kinetics between Injured limb and control group differences.

Variable	Groups		*P* value
	Injured limb (n=34) *(Means ± S.D)*	Control (n=34) *(Means ± S.D)*	
Hip Internal Rotation Angle (°)	10.39±5.57	8.36±8.21	0.29
Hip Internal Rotation Moment (Nm/Kg)	-0.46± 0.14	-0.43±0.25	0.15
Hip Adduction Angle (°)	13.28±5.07	11.40±5.19	*0.05
Hip Adduction Moment (Nm/Kg)	-0.93±0.21	-0.97±0.19	0.89
Knee Adduction Angle (°)	14.57±6.69	12.28±7.27	0.18
Knee Adduction Moment (Nm/Kg)	-0.50±0.15	-0.42±0.19	0.06
Knee Flexion Angle (°)	82.44±12.13	90.04±10.91	0.06
Knee Extensor Moment (Nm/Kg)	1.61±0.29	1.96±0.33	*0.01
VGRF (Nm/Kg)	1.10±0.03	1.12±0.05	0.07

*Key: °=degrees and N.m/kg=Normalised knee moment. * significant differences at p=0.05*

5.3.11.3 Kinematic and kinetic differences between non-injured limbs and the control group during SLS:

The kinematic and kinetic differences between the non-injured limb after ACLR and the control group during SLS are shown in Table 5.16. The hip internal rotation moment showed a significant increase in the non-injured limb for those with ACLR compared to individuals in the control group (p=0.04: effect size 0.36). The internal knee extensor moment demonstrated a significant reduction in the non-injured limb for the ACLR group in comparison to the control group (p=0.01: effect size 0.87). For the other kinematic and kinetic variables, the results were not significantly different between the control group and non-injured limbs for ACLR patients.

Table 5.16: Kinematics and kinetics with between non-Injured and control group differences

Variable	Groups		*P* value
	Non-Injured limb (n=34) *(Means ± S.D)*	Control (n=34) *(Means ± S.D)*	
Hip Internal Rotation Angle (°)	6.60±4.98	8.36±8.21	0.18
Hip Internal Rotation Moment (Nm/Kg)	-0.51±0.12	-0.43±0.25	*0.04
Hip Adduction Angle (°)	11.72±4.96	11.40±5.19	0.88
Hip Adduction Moment (Nm/Kg)	-0.92±0.22	-0.97±0.19	0.58
Knee Adduction Angle (°)	11.16±6.59	12.28±7.27	0.51
Knee Adduction Moment (Nm/Kg)	-0.46±0.17	-0.42±0.19	0.36
Knee Flexion Angle (°)	83.62±12.61	90.04±10.91	0.22
Knee Extensor Moment (Nm/Kg)	1.71±0.23	1.96±0.33	*0.01
VGRF (Nm/Kg)	1.11±0.04	1.12±0.05	0.60

*Key: °=degrees and N.m/kg=Normalised knee moment. * significant differences at p=0.05*

5.4 Results Summary

The results reveal that there were significant differences in hip transverse (internal rotation angle); however, there were no differences in the hip frontal plane at the hip joint six to eight months after ACL-reconstruction during running and SLS, between limbs (injured and non-injured). Hence, the third and fourth null hypotheses have been partially rejected. The results also show that there were no significant differences in hip transverse at the hip joint six to eight months after ACL-reconstruction during running and SLS, between the ACLR group and the healthy control group. There also appeared to be a significant increase in hip frontal (adduction angle) during SLS; however, there were no differences in the hip frontal plane at the hip joint six to eight months after ACL-reconstruction during running, between the ACLR group and the healthy control group. Hence, the first null hypothesis is accepted and the second null hypothesis has been partially rejected.

A significant reduction in sagittal plane was found (peak knee flexion angle and peak internal knee extensor moment and impulse), although there were no significant differences in the frontal plane at the knee joint six to eight months after ACL-reconstruction during running between limbs (injured and non-injured). Hence, the fifth null hypothesis has been partially rejected. There also appeared to be a significant reduction in sagittal plane (internal knee extensor moment) and a significant increase in frontal plane (knee adduction angle) at the knee joint six to eight months after ACL-reconstruction during SLS between limbs (injured and non-injured). Therefore, the sixth null hypothesis has been rejected.

In addition, the results reveal significant reductions in the sagittal plane (peak knee flexion and internal knee extensor moment and impulse) and significant increases in the frontal plane (knee adduction moments) at the knee joint six to eight months after ACL-reconstruction during running, for ACL reconstruction individuals compared to the healthy control group. Hence, the seventh null hypothesis has been rejected. The results also show that there were significant reductions in sagittal plane (peak knee flexion and internal knee extensor moment) at the knee joint six to eight months after ACL-reconstruction during SLS when comparing ACL reconstruction individuals to the healthy control group, and so the eighth null hypothesis has been partially rejected.

5.5 Discussion

The hip and knee kinetic and kinematic outcomes during SLS and running in the intervention group (ACLR patients six to eight months post-surgery, n=34) and control group (biometrically matched healthy individuals, n=34) were measured using motion analysis techniques in the human performance laboratory. This section of the current research study has aimed to evaluate the significant differences in hip and knee kinetic and kinematic measures between the injured leg of ACLR patients six to eight months after surgery, the non-injured leg of these patients and a matched healthy control group, during running and SLS.

5.5.1 Kinetic Outcomes Discussion:

Kinetic assessment in the present study has shown there was a statistically significant reduction in internal knee extensor moment for the injured limb of the ACLR group compared to both the control group and non-injured limb during running. Stress in the ACL after injury may be the cause of reduction in internal knee extensor moment (Karanikas et al., 2009). In order to prevent undue stress on the reconstructed ACL and facilitate healing, adaptation and compensation through avoidance of extensor moment could potentially be a coping strategy (Grindem et al. 2012). Activities that could induce stress on the ACL include hard pivoting, jumping, and cutting common in a wide range of sporting activities such as volleyball, basketball, football, and gymnastics (Grindem et al., 2012). Grindem et al., (2012) concluded that reduction in internal knee extensor moment and knee-loading could potentially be adaptations to movement which may permit participation in these activities.

Since the literature suggests that adaptations leading to a reduced peak internal knee extensor moment are common in ACLR patients (Hart et al., 2010), previous studies can be compared to the present study using this measurement. In fact, one of the most important adaptions found in the ACLR group is a reduction in peak internal knee extensor moment. A study conducted by DeVita et al., (1998) found that during the early stages following ACL reconstruction, a stress prevention strategy for the new ligament reduced peak internal knee extensor moment. Such stress prevention strategies, developed with ACLR patients through sensory feedback, could be useful in protecting ligaments from further injury and reducing the pain experienced by the patient. This would result in the reorganisation of motor tasks, and re-equilibration of sensory inputs and motor outputs following surgery. Other strategies may be used for the same purpose in ACLR patients (DeVita et al., 1998; Karanikas et al., 2009; Shadmehr, 2004). In conjunction with reduced peak internal knee extensor moment , a reduction in peak knee flexion angle which is indicative of the co-contraction of the hamstring and quadriceps, provides stabilisation at the knee (Knoll et al., 2004; Andriacchi & Dyrby 2005; Von Porat et al., 2006; Alkjaer et al., 2003).

ACLR patients experience quadriceps dysfunction in many cases, which results in deficits in strength and activity. This type of dysfunction may continue following return to full activity and ADLs (Roewer et al., 2011). Links have been found between quadriceps weakness, activation failure, as well as decreased peak internal knee extensor moment and altered sagittal plane knee joint biomechanics in individuals who have undergone ACLR. It should be noted that the present study did not test quadriceps strength or activation in either the intervention or control groups. There is, however, ample evidence to clearly demonstrate a reduction in internal knee extensor moment during peak loading between the injured leg of the ACLR group and both the non-injured leg and the control group. As a result, this sagittal plane kinetic pattern may be indicative of early post-injury compensation that develops to account for decreased quadriceps function, increased knee joint effusion, and pain (Ingersoll et al., 2008; Lewek et al., 2002). It should be noted that the quadriceps of the involved limb undergoes profound weakening subsequent to ACL rupture. Even during early periods after injury, the weakness of the quadriceps can be substantial, regardless of the limited time atrophy takes to occur. These strength deficits can potentially persist in patients for prolonged periods after ACL reconstruction, and abnormal biomechanics following ACLR may significantly impact long-term joint health. Preoperative weakness of the quadriceps is a risk factor for poor quadriceps strength and low self-reported scores on validated outcome measures post-surgery. This cycle of reduced internal knee extensor moment/impulse and impairment in quadriceps strength commences at the time of injury and can persist for an extended period after reconstruction; this has significant implications for patient function.

A reduction in peak internal knee extensor moments was found in the current study. This result has also been noted in similar studies (Sigward et al., 2015; Kuenze et al., 2014; Bush-Joseph et al., 2001; Berchuck et al., 1990; Patel et al., 2003). However, there is insufficient evidence to substantiate the relationship between increased loading and incidence of OA in the ACL-injured group (Bush-Joseph et al., 2001; Berchuck et al., 1990; Patel et al., 2003). Berchuck et al., (1990) showed that the quadriceps avoidance strategy was capable of reducing the peak internal knee extensor moments and flexion angle, which helped to provide knee joint stabilisation following ACLR. Knee extensor moment/opposing moments may be high when considering muscle co-contractions, especially in consideration of eccentric contraction of the hamstrings. Increased co-contraction has been noted in individuals with ACL pathologies. This type of contraction is commonly considered to be a defence mechanism aimed at providing knee joint stabilisation following ACL injury (Andriacchi and Dyrby 2005; Alkjaer et al., 2003; Von Porat et al., 2006; Knoll et al., 2004). It should also be noted that such a defence mechanism could be the cause of increased compressive forces within the knee. If this defence mechanism is abnormal, or if the loading area of the cartilage is unable to handle increased loads, there is a high possibility of the occurrence of degenerative changes. Owing to the above findings, the null hypothesis in the current study that hypothesised no significant differences in the sagittal plane movements after surgery between ACL group and healthy controls has been rejected.

Tibone and Antich (1988) found that ACL participants had improved abilities in performing a cutting manoeuvre, and there were significant differences in relation to force plate data. The present study confirms the previously reported findings concerning the reduction in knee extensor impulse during running. There is a clear need to examine knee extensor impulse in activities other than walking and cutting, since the literature lacks supported conclusions on any differences that might exist in the ACLR group during such activities. Load reduction in the sagittal plane has been accomplished through a protective strategy resulting in the reduction in peak internal knee extensor moment (Berchuck et al., 1990; DeVita et al., 1998). At the same time, external knee adduction moment could cause knee loads that have been associated with degenerative changes (Butler et al., 2009; Webster et al., 2011). Therefore, further investigation in this area is needed, particularly in consideration as to whether different loading types are associated with different structural changes within the knee. These investigations should aim to determine what knee compartments are responsible for the changes. Impact due to increased shear, axial loading and compressive loading may affect knee structures in different ways. Tibiofemoral cartilage is well adapted to compressive loading, but considered to be poorly adapted to shear and axial loads (Andriacchi and Mündermann 2006; Andriacchi et al., 2004).

Frontal plane moments influence knee loading patterns and may affect the progression of OA. A six-fold increased risk in OA has been observed if peak external knee adduction moment is increased by 1% (Miyazaki et al., 2002). Increased peak external knee adduction moment is reported to be an important outcome measure in ACL injuries since it increases loading in the medial compartments (Butler et al., 2009). Consequently, external hip adduction, internal rotation, peak internal knee extensor moment and external knee adduction moment during SLS and running were included in this kinetic analysis. Butler et al., (2009) reported similar findings to the present study, showing that peak external knee adduction moment was significantly increased in the ACLR group compared to the control group. This is important since increased peak external knee adduction moment and greater loading of the medial compartment can initiate OA. In contrast to the present study, Webster et al., (2011) found no differences in frontal plane kinetics. A number of factors, including gender bias, may help to explain the differences between these studies. Indeed, Webster et al., (2011) did not analyse for gender effects and suggested that more research was needed to clarify whether increased external knee adduction moment was affected by gender, and if females with ACL injuries are at a higher risk of developing OA than males. According to the findings of the current study, the null hypothesis pertaining to the lack of differences in knee joint frontal after ACLR during running between the limbs has been rejected, as well as the same hypothesis with respect to SLS being rejected.

It would appear that time after surgery is an important factor in the differences seen between this and other studies. For example, in this study, patients were on average 7.1±1.6 months post-surgery, whereas Butler et al., (2009) used ACLR patients that had received ACLR 5.3±4.4 years before recruitment. It would be of interest to follow-up the kinetics and kinematics of ACLR and control subjects in this study after five years, and assess the development and progression of OA. Time from surgery is, however, unlikely to be a causal factor in the differences seen between this study and that of Webster et al., (2011).

This study has reported a statistically significant decrease in the peak internal knee extensor moment between injured and non-injured ACL legs, and between the injured leg and those in the control group. Other studies have suggested that knee loading is less during SLS and that SLS may be a safer way of assessing knee function at higher degrees of knee flexion in ACL injuries than more demanding activities such as hopping (Button et al. 2014); yet this was not the case in this study. Button et al., (2014) have reported differences in peak internal knee extensor moment between injured and non-injured legs of ACLR patients, and between the injured leg and controls during a walking task. Additionally, it was noted that ACLR patients could not resume pre-injury levels of sports activity. These findings suggest that ACLR patients may be at risk of repeated injury, which could initiate OA, unless activity levels are modified after surgery to reduce excessive knee loading (Button et al., 2014).

In this study, determination of knee structural changes was unlikely due to the similarity in biomechanics, demographics and functional outcomes in the sample population. Data presented here suggest that risk factors interact in a complex manner but could be used to forecast early onset OA; this requires further investigation. The injured leg of ACLR patients in this study showed a decreased peak internal knee extensor moment during SLS and running, suggesting that these patients attempt to implement strategies that lower knee loading and prevent further or repeated damage. These findings do not suggest that changes would be initiated by increased loading in the sagittal or frontal planes. Differences observed in levels of performance, kinematic values and knee moments in ACLR patients compared to controls suggest that this group would be at greater risk of degenerative change should the biomechanical model be a contributing factor in early onset PFJ and TFJ OA.

5.6 Kinematic Outcomes Discussion

Kinematic assessment in the current study, showed a significant increase in the hip internal rotation angle between the injured and non-injured legs of ACLR patients during SLS and running. It has to be noted that the null hypothesis pertaining to the hip frontal biomechanics has been partially rejected. Additionally, no studies on ACLR measured the hip kinematics and kinetics after ACLR. The hip internal rotation angle demonstrated statistically significant variation between limbs in the ACLR group during running, whereas comparison of the injured limb and non-injured limb of the ACLR group and control group did not exhibit significant differences. However, there was a significant increase in hip frontal (adduction angle) during SLS. Patellofemoral OA is characterised by an abnormal increase in the hip internal rotation and adduction angles, which will increase the patellofemoral pain from OA (Motlagh et al., 2013, Noehren et al., 2012). Risk factors associated with the development of PFPS include a reduction in knee flexion angle, a reduction in vertical ground reaction force, and an increase in the hip internal rotation angle during different tasks (Motlagh et al., 2013). A decrease in the quadriceps and hamstring strength, coupled with increased hip external rotator strength and increased navicular drop, are risk factors for the development of PFPS (Motlagh et al., 2013).

Knee flexion angle measurement during running in the present study showed that the injured leg of the ACLR group had a significantly lower peak knee flexion angle than the non-injured limb and the control group. This was also found in studies conducted by Sigward et al., (2015), Karanikas et al., (2009) and Gao and Zheng (2010). However, DeVita et al. (1998), Bush-Joseph et al., (2001), Georgoulis et al., (2003) and Webster et al., (2011) take the opposite view, as in their research they found that the ACLR group showed no significant difference in knee flexion angle. Karanikas et al., (2009) discovered adaptations in knee flexion angles within 3-6 months post-surgery and kinematic differences remained at 6-12 months following surgery. The 12+ month assessment showed insignificant changes in kinematic values for knee flexion angle, so time periods after surgery are important for adapting to ACLR. As the knee flexion angle measurement during running in the ACLR group revealed significant lower knee flexion angles compared to the control group and other limbs, the null hypothesis has been rejected.

The ACLR group in the running analysis in the current study was an average of 7.2±1.6 months post-surgery, indicating that the results are comparable when related to time following surgery. There could be significant consequences for long-term joint integrity in ACLR patients with reduced knee flexion angle, since this interferes with the normal ability of the knee to absorb shock during weight acceptance (Cook et al., 1997), and may be linked to early degenerative changes. ACLR may also lead to a stiffer knee that does not flex sufficiently (Rudolph et al., 1998; Bulgheroni et al., 1997; DeVita et al., 1998). Evidence provided by Lewek et al., (2002) shows that deficits in quadriceps strength might produce a reduction in knee flexion, which may lead an increased risk of early degenerative changes, and this correlates with the findings of this research study.

As far as the present study is concerned, significant differences were observed in the peak knee flexion angle, peak internal knee extensor moment and extensor impulses between the reconstructed and uninjured limbs during running 6-8 months after ACLR. This could mean that the rate of loading on articular cartilage may be more detrimental to the joint surfaces than the magnitude of the joint contact forces (Radin et al., 1991); however, this was not tested in the current study. An increased knee joint loading rate in females has also been associated with weakness in quadriceps (Mikesky et al., 2000), which is a concern when considering the integrity of the joint’s contact surfaces.

It has been hypothesised that loading is often reduced when knee flexion angle is reduced as a strategy of adaptation, because this lowers the risk of damage, reduces pain and maintains stability. Additional damage may be created when knee flexion angle is increased and normal stability and loading boundaries are exceeded, causing excessive stress on the structures of the knee. Knee loading and lever arm stress are increased when the knee is flexed more and the centre of mass of knee rotation is moved away (McGinnis 2013).

A reduction in the knee flexion angle was noted in the intervention group compared to the control group during running. A shift in knee flexion angle likely affects the location of tibiofemoral joint contact and the patellofemoral contact stress (Hall et al. 2012; Boling et al. 2009). Greater knee flexion angles, increased knee moments and loading, and greater speed, are required during running when compared to other tasks (Frobell et al., 2010). It has been hypothesised that as running places a greater demand on the knee, abnormal kinetics and kinematics could be present during this activity. Considering the frequent occurrence and abnormal loading associated with these activities, they may potentially have an influence on the long-term health of the knee (Frobell et al., 2010). Increased localisation of stress in areas of greater contact is indicative of elevated cartilage damage risk. A reduction in the knee flexion angle paves the way towards decreasing the patellofemoral contact area. Large patellofemoral loads, coupled with smaller knee flexion angles and decreased contact area, during high demand tasks, could potentially increase the risk of early patellofemoral disease progression (Boiling et al., 2009). Smaller knee flexion angles result in patellofemoral contact stress increasing, in addition to a reduction in the knee extensor moment (Boiling et al., 2009). In the present study, it has been shown that ACLR patients may reduce knee flexion in order to reduce patellofemoral compressive forces and consequently reduce pain (Dierks et al., 2010).

In this study, symptoms of pain and ADL, and patients reporting acceptable symptoms following ACLR, were within the norms reported for KOOS Sport/Rec and QOL. According to Hartigan et al., (2013), kinesiophobia levels were elevated in patients prior to ACL reconstruction. This was predominant in those with poorer dynamic knee stability and non-copers. Subsequent to ACL reconstruction, it was noted that kinesiophobia levels reduced the most in non-copers (Hartigan et al., 2013). It was also noted that reductions in kinesiophobia had a significant relationship to improvements in self-reported knee function during ADL (Hartigan et al., 2013). As far as the clinical scenario is concerned, kinesiophobia levels were reported to be high at six months, and plateaued between six and 12 months post-surgery (Hartigan et al., 2013). It is important to note that kinesiophobia levels should be monitored from the time of ACL rupture to 12 months after surgery (Hartigan et al., 2013).

As far as the present study is concerned, significant differences in the extensor moment, knee flexion and extensor impulse between the reconstructed and uninjured limbs during running six to eight months after ACLR were noted, and so the null hypothesis has been rejected. This could mean that the rate of loading on articular cartilage may be more detrimental to the joint surfaces than the magnitude of the joint contact forces; however, this was not tested in the current study (Radin et al., 1991). The current study reported a lack of statistically significant differences in the knee adduction angle, thereby paving the way for partial acceptance of the hypothesis, as there were no differences in the frontal plane movements prior to and subsequent to ACLR during running and SLS (between reconstructed limbs and uninjured group). The study found significant increases in the external knee adduction moments, which resulted in partially rejecting the null hypothesis. Here, there were significant increases in external knee adduction moments, which resulted in partial acceptance as well as rejection of the null hypothesis. The present study also noted differences in the knee joint sagittal planes movement six to eight months after ACLR during running and SLS between the ACLR group and uninjured group.

The current study also noted statistically significant differences when comparing the non-injured limb in the ACLR group and the control group. A statistically significant difference was exhibited, with an increase in the hip internal rotation moment and reduction in peak internal knee extensor moments for non-injured limbs of the ACLR group compared to the control group during SLS. The study conducted by Pahnabi et al. (2014) compared the uninjured legs of the experimental group with the healthy legs of the control group. Overloading of the uninjured leg was reported in individuals with ACLR knees, thereby leading to increased stress on the uninjured leg. Bonfim et al., (2008) report a reduction in the neural signal transmissions in the injured leg due to ACL injury; this caused motor control system malfunction and the performance of the uninjured leg was adversely affected. Bonfim et al., (2008) have also reported changes in performance when an ACLD group used their uninjured legs for locomotion. According to Jones et al., (2013), the risk of OA is high for the contralateral knee, irrespective of the occurrence of the injury. Shakoor et al., (2002) have reported a high risk of OA in the contralateral knee for patients undergoing a unilateral hip replacement.

As stated previously, in comparison with the control group and participants' uninjured leg, peak internal knee extensor moment was reduced significantly for the injured leg of ACLR subjects and knee flexion angle was directionally, but not statistically, significantly reduced in ACLR subjects during SLS. In addition, hip flexion was shown to be reduced when knee flexion angles were increased (Yamazaki et al., 2009). As a consequence of these changes, different strategies are used at the ankle and hip to achieve normal squat depth. Since knee function is measured by squat depth, this may explain the results obtained during the analysis of knee flexion angle. Increased hip flexion angle and reduced knee flexion angle are inter-related, as shown in this current study, at greater squat depths.

It was not possible to facilitate the comparison of the results obtained here relating to kinematics after ACL reconstruction during SLS, since no previous studies are available. The biomechanical study on ACLD patients previously discussed (Yamazaki et al., 2009) has described a similar in knee flexion angle. There is a need to high demand examine knee flexion angle for the ACLR group further, since the paucity of literature on the subject prevents evidence-based conclusions from being made for ACLR patients. Yamazaki et al., (2009) compared findings with a control group, and reported increased knee adduction angle in ACLD, as this study found significant increases in peak knee adduction angle, therefore the null hypothesis has been rejected.

The ACLR group in the current research study pertaining to running, demonstrated kinematics and kinetic adaptations between limbs and in comparison with the control group. There were alterations in kinetics, but kinematics were at normal levels during SLS activity. Knee health in the longer term is likely to be influenced by demographics, function, and biomechanics, as well as how these combine and inter-relate, so it is important to analyse these interrelationships individually (Noehren et al., 2012). When an injured knee joint experiences greater demands, tasks need to be assessed for single leg stance, as patients may never fully recover from injuries. Therefore, that movement control requires more muscle proprioceptive responses, muscle co-contractions and muscle contractions when activity demands are increased (Noehren et al., 2012).

5.7 Subjective Knee Evaluation

As detailed in the aforementioned sections, this study has delved deep into utilising subjective knee evaluation, as the significant changes in pain or another subscale, may have an effect on biomechanics, and if so, could be quantified by the same. Contemporary researchers such as Ingelsrud et al., (2015) have used KOOS subscale values to report pain, symptoms, ADL, poor outcomes pertaining to sports/recreation, and QOL following six months of ACLR. As far as symptoms, pain, and ADL are concerned, patients reported acceptable outcomes in symptoms, pain, and ADL following ACLR. However, the outcomes were reduced in the sports/recreation group and QOL. According to the findings of Barenius et al., (2013), 30% of patients perceived their postoperative outcomes as poor, owing to treatment failure. The study by Barenius et al., (2013) used a KOOS QOL subscale value of "< 44" points as a treatment failure cut-off point. This originated from a randomised controlled trial for ACL injuries treatment and has been used as a criterion for the crossover between surgical and nonsurgical treatment (Frobell et al., 2010). In this research study, a QOL score of 77 was considered an inferior outcome pertaining to the patient score six to eight months following surgery. It has to be noted that the KOOS subscales depicted a significant difference in the KOOS-symptoms and KOOS-QoL knee function scores between the participant groups with the ACLR groups, demonstrating significantly reduced levels of subjective knee function compared to the control participants.

The IKDC is validated for use in post-ACLR, and it offers a reliable and valid knee-specific measurement of symptoms, function and sports activity (Irrgang et al., 2001). In the current study, the ACLR group had an average score of (88.40). The study conducted by Grindem et al., (2011) aimed to examine the predictors associated with functional outcomes after ACL reconstruction using IKDC and single-legged hop tests in non-operatively treated ACL-injured individuals, and they found reduced knee function. Similar results were reported by Anderson et al., (2006). In this research study, the ACLR group exhibited similar IKDC and run-performance scores as ACLD patients one-year post injury (89.0) (Grindem et al., 2011). It has been suggested by Ardern et al., (2011) and Paterno et al., (2010) that an appropriate time of at least nine months is a mandatory requisite for maximising functional recovery following ACLR. This could potentially be the reason behind the low scores in the current sample, and the possibility of further improvements could be apparent after one year or later. The median of the IKDC scores in the group with scores within normal ranges was

88.98, which suggests average functionality during the postoperative period, six to eight months following ACLR.

5.8 PFJ and TFJ Osteoarthritis

When PFOA is present, greater flexion angles are shown to be linked to the high compressive forces of the patellofemoral joint, because these are limited by strategies that compensate for this in individuals' patterns of movement. The current study observed lower peak knee flexion angle and extensor moment for individuals with ACLR, which is not consistent with results of Teng et al., (2015). Patellofemoral biomechanics are shown to be detrimentally affected by hamstring activation when elevated antagonist is simulated. This means that individuals with ACLR show movement patterns that are distinct, and this could contribute towards respondents reporting symptoms of stiffness, probably due to the co-contraction of muscles that is increased as a result (Li et al., 2004). According to Teng et al., (2015), when cartilage lesions of the patellofemoral are assessed in older individuals using MRI technology, and diagnosed as early stage patellofemoral disease, patellofemoral stress and extensor moment shows a higher peak. These findings are not supported by the current study, as individuals with ACLR were observed to display lower movement patterns at the sagittal plane. This study has focused on the first part of the loading phase during running for assessment purposes to understand the biomechanics involved. This differs from the study by Teng et al., (2015) who focused on contralateral limb transference of weight during test exercises of gait in the second part of the stance phase, with older individuals with early stage PFOA, to investigate biomechanical features.

The reduction in internal knee extensor moment and knee flexion angle is a strategy adopted to reduce or avoid contraction of the quadriceps, that is, quadriceps avoidance (Berchuck et al., 1990). The findings from the current study were obtained by limiting the contact area over which patellofemoral loads can be distributed. This thus increases the contact stress on small areas of patellofemoral contact by using smaller knee flexion angles during high demand tasks, which may increase the risk of early patellofemoral disease progression (Boling et al., 2009). The increased risk of cartilage damage potentially follows more localised areas of contact stress. Moreover, Boling et al., (2009) report that individuals with PFPS also exhibit patellar misalignment. The patellofemoral misalignment stems from decreased knee-extensor moment (Kuenze et al., 2014). A decrease in such moments would increase the dynamic of the patella. Patellar misalignment would predispose the risk of increased loading and PFPS/PFOA (Boling et al., 2009, Culvenor et al.,2016, Waryasz and Mcdermott, 2008).

Stress pain and contact pain are increased for PFPS subjects that present greater lateral patella tilt, and are connected to the femur internal rotation that is increased during running tasks. Pain results from patellofemoral joint stress at the same structures and locations because running tasks required a range of motion that is constrained (Draper et al. 2009). In running tasks, the control group applied limitations on specific areas of the patellofemoral joint, so that repetitive stress was reduced, and throughout the running tasks these subjects presented internal rotation that was increased, but less motion overall was shown when compared with PFPS subjects (Noehren et al. 2012). In the current study, comparisons between limbs for SLS and running showed differences that were significant, but still within the error of measurement defined as standard.

Kuenze et al., (2014) have explored the relationship between muscle weakness and incidence of knee OA, and have shown that progressive loading following ACLR could lead to the progression of OA. The incidence of muscle weakness in those with knee OA has been reported by Palmieri-Smith et al., (2010). Patients with PFJ OA may be prone to a greater risk of muscle weakness in comparison to those with TFJ OA. Youssef et al., (2009) found that weakness in the quadriceps muscle led to increased degeneration in the retro-patellar cartilage in an animal model, and this suggests that muscle weakness is a potential risk factor associated with the onset and progression of osteoarthritis (OA). It was also noted that quadriceps weakening led to subsequent degeneration in the patellofemoral joint rather than the tibiofemoral joint. Protection against lateral PFJ but not TFJ cartilage loss in humans was conferred by greater quadriceps strength. Quadriceps weakness may be a contributor to the development of PFJ OA, with strength deficits persisting for up to six years following ACLR (Keays et al., 2007). However, Keays et al., (2010) claim that there is no association between quadriceps strength and PFJ OA. Thus, the relationship between PFJ OA development and quadriceps weakness is unclear.

Quadriceps weakness in post-ACLR patients is reported to be associated with lower knee extensor moments, because quadriceps strength deficit is linked to lower knee flexion angle and knee extensor moment during running (Lewek et al. 2002). In contrast to the findings of Keays et al., (2007), Wang et al., (2015) suggest that development of chondral lesions in the patellofemoral joint after ACLR is linked to weakness of the quadriceps. The increase in knee flexion angle has been considered an efficacious strategy that enables running without pain, as the shock of running is reduced by the quadriceps when their strength is greater (Wang et al., 2015). Atraumatic individuals have a greater risk of developing TFOA, as this is consistently associated with PFOA patterns. This is due to high axial compression, which causes adverse effects in the tibiofemoral joint over a period of time for individuals with early stage PFOA. It should be noted that the active mechanisms for absorbing shock are lower in individuals with PFOA (Duncan et al., 2008). Various studies suggest that when individuals present with quadriceps weakness they are also likely to be at a higher risk of degenerative changes at the patellofemoral and tibiofemoral joint; this is a significant concern since it may be followed by symptoms of OA (Amin et al., 2009; Oiestad et al., 2015). At the same time, external knee adduction moment could cause knee loads that have been associated with degenerative OA changes in TFJ (Butler et al., 2009; Webster et al., 2011). Tibiofemoral cartilage is well adapted to compressive loading but considered to be poorly adapted to shear and axial loads (Andriacchi and Mündermann 2006; Andriacchi et al., 2004). The progression of OA may be enhanced by compressive loading once knee OA is present (Andriacchi et al., 2004).

There are some limitations in the current study since the 3D motion analysis was conducted using cluster markers. These are susceptible to error due to soft tissue artefacts, and because frontal and transverse planes of motion are susceptible to this type of error when high-velocity movements are involved (Cappozzo et al., 1996; McGinley et al., 2009). Joint degeneration could result from increased knee joint loading, and the risk factors that contribute to this include altered lower extremity biomechanics and quadriceps weakness following ACLR. The implications in the longer term from such differences remain unclear, so further research should focus on these factors (McGinley et al., 2009). Individuals who have a history of ACLR should be observed in order to identify sources of biomechanical alterations, and allow conclusions to be drawn about the role of quadriceps strength, which is a concern for both academics and clinicians (Cappozzo et al., 1996; McGinley et al., 2009). Therefore, in patients with severe PFOA in knee OA, gait deviations from observed loading responses could be helped by improving the knee flexion range of motion and quadriceps strength through effective strategies (McGinley et al., 2009). It should be emphasised that there are positive clinical implications pertaining to biomechanical alterations (using feedback) of knee flexion and gait retraining using sensors, smart gait retraining, EMG, and so on (McGinley et al., 2009).

Other limitations within the current study are related to task standardisation, such as squat depth, which is known to elevate knee-joint loads (Besier et al., 2001). The highly-controlled laboratory environment is another limitation of this study, as how do these findings translate to the day-to-day conditions experienced by ACLR patients? Future investigators should seek to transfer the findings from such standardised investigations into more ecologically valid evaluations of loading and injury risk in actual sports environments and training sessions, taking into account ongoing evolutions in technology. The long-term implications of these differences are unclear, yet the interaction of quadriceps weakness following ACLR, together with the altered lower extremity biomechanics, may be an important risk factor for increased knee joint loading and subsequent degeneration of the joint.

All participants in the control group were healthy, highly active university students, which is another limitation. It is unclear whether age or activity levels would affect such comparisons; these results may not be applicable to elite athletes, adolescents or older age groups. Patients undertaking different types of sport may affect the generalisations contained in these results. Since ACL injury has been linked to excessive hip adduction and internal rotation angles, as well as knee valgus angle during different functional tasks, the reliability of running and SLS tasks in a population with ACL injury requires further investigation (Hewett et al., 2004; Willson and Davis 2008). Subjects with ACL injuries develop adaptations to structure and movement that change over time leading to the changes in knee structure movements and limb movements described in the literature. The ACLR groups in the current study may be categorised in terms of post-injury period in the short-to-medium term; it would be beneficial to reassess these patients in a follow-up study at longer intervals post-surgery.

5.9 Conclusion

This study suggests that six to eight months following ACL reconstruction, athletes in the intervention group exhibited a specific degree of altered hip and knee joint kinematics and kinetics. The reduction in peak internal knee extensor moment and peak knee flexion angle is a strategy adopted to reduce or avoid contraction of the quadriceps; this is called quadriceps avoidance. However, these reductions decrease the patella contact area, resulting in increased patellofemoral contact stress over time. Repetitive movements may, therefore, contribute to patellofemoral disorders and increase the risk of degenerative joint disease- a progression commonly found in ACLR patients after surgery. The results of this study may help to guide the development of new or alternative treatment options for improving long-term joint health after ACL injury.

Previous research studies on patients with knee OA show that increased knee adduction moment is related to significant narrowing of the medial joint space. As far as the author is aware, no published prospective studies have investigated the biomechanical differences between patients during the preoperative phase, and the same patients during the rehabilitation phase after ACLR. The following chapter will address this issue by analysing the biomechanical alterations observed during running and SLS before and after ACLR surgery. Finally, in accordance with the results obtained here, some of the null hypotheses have been rejected, and a few have been partially rejected or accepted.

CHAPTER SIX

FUNCTIONAL TASKS BEFORE AND AFTER ANTERIOR CRUCIATE LIGAMENT (ACL) RECONSTRUCTION– AN EVALUATION

6.1 Background

Anterior cruciate ligament (ACL) injuries, along with long-term changes in dynamic loading, can lead to early knee osteoarthritis (Andersen et al., 2004; Myklebust and Bahr 2005). After a period of 10 years, half of patients show radiological signs of osteoarthritis, and almost all patients will suffer from osteoarthritis (OA) within fifteen to twenty years (Myklebust and Bahr 2005). Almost 80% of ACL injuries occur during non-contact situations such as landing from a jump or running during an activity (Renstrom et al., 2008). Running is the most important sport for many, because of its health benefits and economical nature. However, running injuries have been well illustrated in previous studies. Macera et al., (1989) report running injuries at annual rates of 24–65%. The knee is the most commonly injured area during running, and it is believed that some of these injuries are affected by abnormal knee motion (James et al., 1979). ACL injury produces changes in lower extremity kinematics and kinetics during running (Sigward et al., 2015).

Bulgheroni et al., (1997) assessed the effect of ACL reconstruction on kinematic and kinetic patterns during walking. They have shown that patients exhibit an abnormal gait pattern after the surgical procedure, with different knee extensor moments. It has also been shown that abnormal gait patterns might have consequences in the long term, leading to pathological knee conditions (Georgoulis et al., 2003; Noyes et al., 1992). There is little data on why individuals with an ACL-deficient (ACLD) or ACL-reconstructed (ACLR) knee subsequently develop OA (Chaudhari et al., 2008). ACL injury is a contributing factor in up to 30,000 knee arthroplasties annually in the United States (Mather et al., 2013), and altered kinematics and kinetics are one of the hypothesised reasons. Numerous studies that have investigated walking gait have found significant reductions in internal knee extensor moment, and small reductions in peak knee flexion angle, with individuals adopting a quadriceps avoidance gait pattern during walking (DeVita et al., 1998). One of the most common activities pre- and post-surgery is running, and it is not known whether individuals before and after ACLR have different knee kinematic and kinetic patterns to healthy individuals. In the absence of typical functioning of the ACL, the ACL-D individuals may subconsciously avoid contraction of the quadriceps to avoid displacing the tibia anteriorly during running (Berchuck et al., 1990) when dynamic loads are greater, and this alteration may continue after ACLR. This may then increase the risk of developing degenerative diseases due to excess loading; therefore, understanding if there are differences is of utmost importance.

The kinematics and kinetics between before and after ACL reconstruction have been compared in various studies. Knoll et al., (2004) examined the walking gait pattern of ACL injured individuals during walking before surgery and two weeks, six weeks, four months, eight months and 12 months post-surgical reconstruction. The subjects showed a quadriceps avoidance gait prior to surgery and six weeks post-surgery. The strategy of a quadriceps avoidance gait is described by Berchuck et al., (1990) as a reduction in internal extensor moment This gait pattern was still evident five weeks post-surgery, but more similar to the control group. The reason for this is that before and after ACL reconstruction, the patient may have different kinematics and kinetics, which results in different performance abilities during running. Moreover, this may have consequences for long-term rehabilitation and return to sport for the individual. Currently, there is no research detailing kinematics and kinetics before and after ACL reconstruction during running. However, as mentioned previously, in general clinical space, the option to actually ask the individual to run and assess this is limited; and so a more space-optimised clinical assessment is needed. Therefore, the single leg squat (SLS) was chosen as the measure to assess these individuals.

Whatman et al., (2011) investigated the links between lower-limb kinematics during running and those occurring during SLS. They found moderate to very large correlations to be shown by the Pearson correlation coefficients, between the peak ankle, knee and hip angles, recorded during both SLS and during running (r= 0.70 to 0.89). Despite using a small sample size and control velocity, Claiborne et al., (2006) describe SLS as a controlled, yet dynamic, manoeuvre that can be extrapolated to many functional actions, such as single leg landing, running and changing direction tasks. As such, they mimicked most of the movements that athletes have to perform during the course of their sporting activity. The inference here is that FPTs in the form of squatting tests allow for an examination of the biomechanics of ACLR / ACLD knees in the real-life settings of athletes. Regarding leg hop tests, although they provide more objective measurements compared to time postoperatively, they fail to address the multi-planar motion that is characterised in a cutting task (Coats et al., 2013). Therefore, the single leg squat has also been used as a valid and reliable assessment tool for the analysis of faulty movement patterns, especially regarding preventing injury at the trunk, hip, and knee (Myer et al., 2012).

Yamazaki et al. (2010) compared the kinematics of SLS between ACLD individuals and controls. When comparing the injured and non-injured legs of participants, the injured leg of both male and female individuals showed more knee adduction than the non-injured leg. The injured leg of the male individuals showed less knee and hip external rotation angles, less knee flexion and more knee adduction than those of the non-injured subjects, which are risk factors for PFJ and TFJ OA. The investigation was carried out before and after ACL reconstruction because the patient may have different kinematics and kinetics, which will result in different performance abilities during sporting activities, and this may have consequences for knee OA. To date, no studies have been found that have compared the kinematics and kinetics after ACLR during an SLS task.

Therefore, the purpose of this study is to compare the hip and knee kinematics and kinetics before and after for ACLR individuals and a control group during running and SLS.

6.2 Objectives and Null Hypotheses

1. To investigate the hip joint frontal and transverse plane movements between limbs before and after ACL reconstruction during running (injured and non-injured).

   Null hypotheses: there will be no significant difference between limbs in the hip joint frontal and transverse planes movements before and after acl reconstruction during running (injured and non-injured).

2. To investigate the hip joint frontal and transverse plane movements between limbs before and after ACL reconstruction during SLS (injured and non-injured).

   Null hypotheses: there will be no significant difference between limbs in the hip joint frontal and transverse planes movements before and after ACL reconstruction during SLS (injured and non-injured).

3. To compare the hip joint frontal and transverse plane movements before and after acl reconstruction with healthy control group during running.

   Null hypotheses: there will be no significant difference in the hip joint frontal and transverse planes movements between before and after ACL reconstruction individuals, and healthy control group during running.

4. To compare the hip joint frontal and transverse plane movements before and after acl reconstruction with healthy control group during SLS.

   Null hypotheses: there will be no significant difference in the hip joint frontal and transverse plane movements between before and after ACL reconstruction individuals and healthy control group during SLS.

5. To investigate the knee joint sagittal and frontal plane movements between limbs before and after ACL reconstruction during running (injured and non-injured).

   Null hypotheses: there will be no significant difference between limbs in the knee joint sagittal and frontal plane movements before and after ACL reconstruction during running (injured and non-injured).

6. To investigate the knee joint sagittal and frontal plane movements between limbs before and after ACL reconstruction during SLS (injured and non-injured).

   Null hypotheses: there will be no significant difference between limbs in the knee joint sagittal and frontal planes movements before and after ACL reconstruction during SLS (injured and non-injured).

7. To compare the knee joint sagittal and frontal plane movements before and after ACL reconstruction with healthy control group during running.

   Null hypotheses: there will be no significant difference in the knee joint sagittal and frontal plane movements between before and after ACL reconstruction individuals and healthy control group during running.

8. To compare the knee joint sagittal and frontal planes movements before and after ACL reconstruction with healthy control group during SLS.

   Null hypotheses: there will be no significant difference in the knee joint sagittal and frontal planes movements between before and after ACL reconstruction individuals and healthy control group during SLS.

6.3 Methods

6.3.1 Research Environment

The running gait and SLS analysis work was all completed at the Human Performance Laboratory (situated in the Mary Seacole Building, University of Salford), which has a strong record in musculoskeletal research and clinical gait analysis.

6.3.2 Participants:

The study was conducted on six anterior cruciate ligament deficient subjects two months prior to and three and six months following ACL reconstructive surgery. The surgery technique was four hamstring graft and two patellar tendon grafts. The College of Health Sciences Research Governance and Ethical Committee approved the study (HSCR13/74). Additionally, this study received a favourable opinion from the NHS ethical committee (NHS rec number 14/LO/0255) (Appendix1). The recruitment was initially from orthopaedic consultants at Stepping Hill Hospital, Stockport, or private hospitals. Participant Identification Centres (PICs) were arranged at this trust, and invitation letters (see Appendix 6) were made available to give to patients, along with the participant information sheet (Appendix 4). Individuals were requested to return the data access forms in the stamped addressed envelope provided to indicate they were interested in the study. Once the form was received, the principal investigator would contact the individual to check their eligibility. If they were eligible and still interested in the study, an appointment was made for them at the Human Performance laboratory at the University of Salford. All of the participants attended the same post-operative rehabilitation, which involved introducing weight-bearing exercises, and a range of movement exercises; neuromuscular retraining, and a step-by-step return to sport.

The six subjects that made up the healthy control group were highly active; all of them are students recruited from the university, and they volunteered for the study. These participants had no history of lower extremity surgery, and had not suffered any lower extremity injuries during the previous six months. The definition of injury was all musculoskeletal complaints that had prevented engagement in normal exercise routines for participants. Ethical approval was obtained (HSCR12/64), and all participants were required to read and sign the informed consent statement before testing.

6.3.3 Inclusion criteria

To be eligible for the study, the participants must have met the following criteria:

- Aged between 18 to 40 years old
- Adult with a unilateral ACL injury and have had or plan to have an ACL reconstruction with a hamstring graft and patellar tendon graft, and with or without an accompanying meniscal tear
- Confident of being able to run for five minutes without an adverse reaction. This was also determined by the referring clinician, along with gaining approval for the individual to run.

6.3.4 Exclusion criteria

Individuals with any of the following criteria were excluded from the study:

- Unable to give informed consent or comply with the study procedures
- Subjects with cardiovascular, pulmonary or neurological conditions that limit physical activity
- Subjects with any lower limb, pelvic or spinal pathology that limits their ability to run comfortably for five minutes
- Those who did not run regularly prior to the injury
- Surgery involving procedures other than an isolated ACL reconstruction using either a hamstring or patella tendon autograft

6.3.5 Procedures

Once subjects demonstrated that they were interested in the study, they were given an appointment at the Human Performance Laboratory at the University of Salford. When participants attended the Human Performance laboratory for their initial visit, they were briefed on the study and had all the equipment and procedures explained to them. Any questions were answered in full, and if happy, they were asked to sign the informed consent form (see Appendix 5). Participants also completed the Knee Injury and Osteoarthritis Outcome Score (KOOS) questionnaire (see Appendix 2), which measures subjective outcomes (symptoms, stiffness, pain, daily living, sports and recreational activities function, and quality of life) to determine the outcomes after ACL injury. The International Knee Documentation Committee Scale (IKDC) evaluation scales (Appendix 3) were utilised before and three and six months post ACL reconstruction for the ACLR group; they were also used with the control group one time before the session. A demographics form was also completed (age, height and weight) for each of the participants. Each participant took part at three points in time- before and three and six months after ACLR. A twelve-camera motion analysis system (Pro-Reflex, Qualisys, Sweden), sampling at 240 Hz, was used to collect kinematic data. A force platform was embedded into the floor (AMTI, USA), and sampling at 1200 Hz was used to collect kinetic data. As previously outlined in the methodology chapter (Chapter 4, Sections 4.1-3), the same instrumentation, calibration, training shoes, filtration, biomechanical model and marker list were deployed. Sections 4.3.1-2 have already described the running and SLS tasks.

6.3.6 Outcome measures

The discrete variables were selected based on basis of their frequency of use in relation to possible ACL and PFPS and OA injury studies, as discussed in Chapter Two. Alongside the clinical questionnaire results, the following discrete variables were calculated during the early stance (0-50) phase for each trial: peaks of hip adduction angle and moment; peaks of hip internal rotation angle and moment; peak knee flexion angle; peak internal knee extensor moment; knee extensor impulse; peak knee valgus angle; peak external knee adduction moment, and peak vertical ground-reaction force (VGRF).

6.3.7 Data analysis

Descriptive analysis (mean and standard deviation) has been carried out for each dependent variable in the target tasks (running and SLS). A Kolmogorov-Smirnov test was used to check whether the data was normally distributed or not (parametric or non-parametric). ANOVA tests for parametric variables and a Friedman test for non-parametric variables were used to assess differences in hip and knee joint kinematics and kinetics across the three time points for ACL patients: before surgery, and three months and six months after ACL reconstruction surgery. Where appropriate, Bonferroni post hoc multiple comparisons were performed using estimated marginal means. Comparisons between the injured limb and non-injured limb for ACLR were carried out using a paired t-test for parametric variables, and a Wilcoxon Rank Test for non- parametric variables. Comparing the injured limb and non-injured limb for ACLR against the control group was done using an independent t-test for parametric variables, and a Mann-Whitney U test for non-parametric variables, as shown in Figure 6.1. The p-value was set at p=0.05 or less to be statistically significant. All statistical analyses were performed using SPSS (v. 20, SPSS Inc., USA).

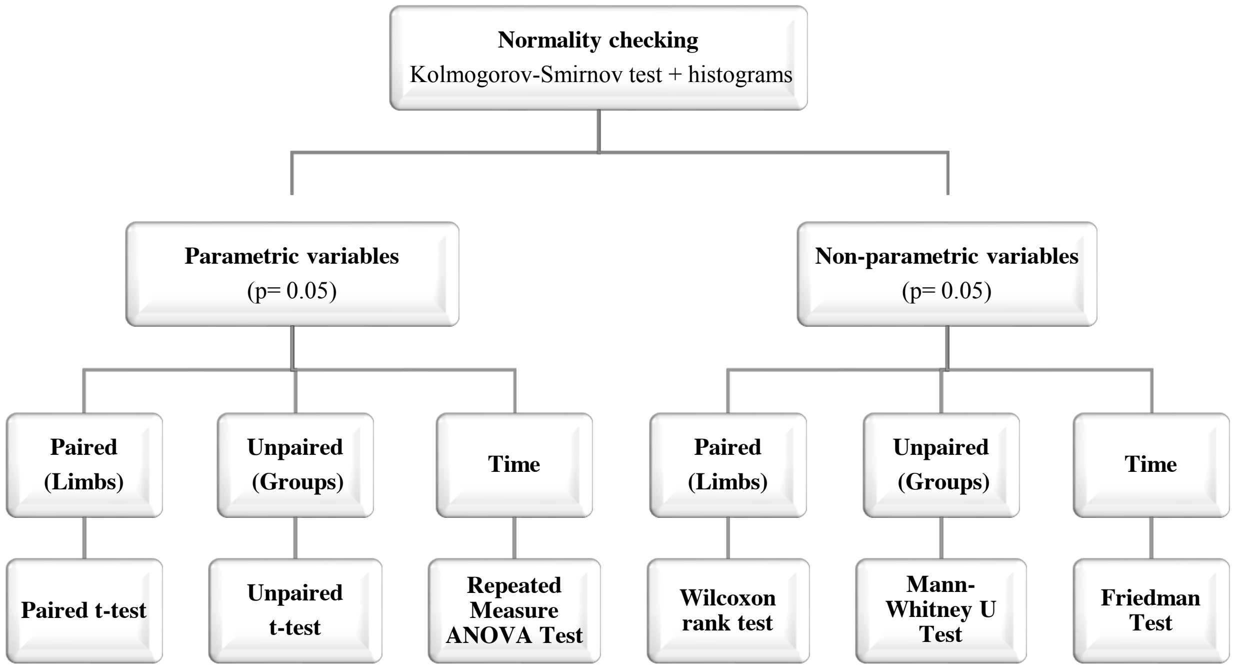


Figure 6.1: Statistical analysis outline

6.4 Results:

6.4.1 Patient Reported Function: The IKDC and KOOS

IKDC and KOOS score data was collected for the ACLR group; the data was normally distributed (Kolmogorov-Smirnov, IKDC p=0.200 and KOOS p=0.200). KOOS and IKDC scores were shown to improve for knee function scores three and six months after ACLR.

Table 6.1: IKDC and KOOS scores before and three and six months after ACLR.

	Participant			
	Pre *(Means ± S.D)*	3 months *(Means ± S.D)*	6 months *(Means ± S.D)*	*P* value
KOOS	70.05±18.07	80.93±11.96	92.60±4.06	0.04*
IKDC	58.23±16.52	73.13±7.82	91±5.64	0.01*

Key: ACLR=ACL injured group, Control=Healthy group. *Signifies group differences at a level of p=0.05.

As shown in Table 6.1, prior to the repair, KOOS and IKDC scores were low, indicating poor knee function (KOOS= 70.05 ±18.07; IKDC= 58.23 ±16.52). However, at three months post ACL surgery, both KOOS and IKDC improved (KOOS= 80.93 ±11.96; IKDC= 83.13 ±782). Patient-reported knee function at six months also improved (KOOS=92.6 ±4.06; IKDC= 91±5.64). Significant differences were noted across all time points for KOOS (p-0.04) and IKDC (p=0.01).

Table 6.2: KOOS Subscales before, three and six months after ACLR

	Symptoms	Pain	Function ADL	Function SRA	Quality of Life
Pre-Op.	64.95±23.76	77.83±20.66	87.25±24.30	55.83±25.00	35.54±21.88
3 Months	77.31±27.87	90.81±6.16	96.88±4.24	74.17±17.50	40.71±24.14
6 Months	92.86±11.06	96.76±2.27	98.53±0.73	86.67±13.23	66.67±15.31
p-value	0.09	0.06	0.25	0.05*	0.02*

Scores (Means ± S.D).

As shown in Table 6.2, in the KOOS subscales, there were significant differences in the KOOS- Quality of Life (p=0.02) and KOOS function SRA (p=0.05) knee function scores between the participants at three time points. The participants during pre-op time demonstrated significantly reduced levels of subjective knee function compared to three and six months post-operatively. The other KOOS scores are not significant due to the small sample size.

6.4.2 Running Analysis

Kinematic and kinetic outcomes, together with running performance measures, were analysed for six controls and six ACLR individuals. These results show the knee adduction angle rather than the knee valgus angle, as there is no valgus during the early stance (0-50) of the stance phase. The testing of kinematic and kinetic outcome parameters for normal distributions utilised Kolmogorov-Smirnov, and the reported values indicate that all hip and knee kinematics and kinetics were normally distributed. Table 6.3 shows there are no significant differences between ACLR group and control group for all data.

Table 6.3: Subjects' Demographic Data

Variable	Participant Group		*P* value
	Control (n=6) *(Means±S.D)*	ACLR (n=6) *(Means±S.D)*	
Age (years)	24.1±3.9	22.3±5.7	0.44
Height (m)	1.72±0.06	1.70±0.08	0.92
Body mass (kg)	65.8±12.5	68±10.9	0.31
Male/Female	4M/2F	4M/2F	NA
Running Speed (m/s)	3.19±0.33	3.20±0.23	0.98
Activity Level (KOOS SRA subscale)	96.67±8.16	86.6±13.2	0.09

Key: m=metres, kg=Kilogram. m/s=Speed in metres per second * Significant differences at p=0.05

6.4.2.1 Kinematic and kinetic differences in the injured limb before, three and six months after ACLR during running:

The kinematic and kinetic results are shown in Table 6.4. The results for all kinematic and kinetic variables show no significant differences in the ACL injury participants measured before, and three and six months after, reconstruction surgery (N=6). The knee adduction angle and external knee adduction moment were increased in the ACL injury participants measured before and after ACLR (p=0.16, p=59 respectively) but this is not significant.

Table 6.4: Kinematics and kinetics before, three and six after ACLR.

Variable	Groups			P -value
	Pre (n=6) *(Means±SD)*	3 Months (n=6) *(Means±SD)*	6 Months (n=6) *(Means±SD)*	
Hip Internal Rotation Angle (°)	4.65±6.44	6.29±6.34	5.37±7.72	0.83
Hip Internal Rotation Moment (Nm/Kg)	-0.59±0.21	-0.62±0.10	-0.66±0.08	0.54
Hip Adduction Angle (°)	14.33±4.31	11.86±7.31	9.97±7.46	0.15
Hip Adduction Moment (Nm/Kg)	-2.05±0.49	-1.79±0.53	-1.76±0.29	0.28
Knee Adduction Angle (°)	3.00±6.13	5.76±7.44	6.19±5.71	0.16
Knee Adduction Moment (Nm/Kg)	-0.62±0.16	-0.59±0.28	-0.68±0.23	0.59
Knee Flexion Angle (°)	46.85±7.11	44.06±5.53	47.69±6.17	0.32
Knee Extensor Moment (Nm/Kg)	2.54±0.66	2.30±0.42	2.48±0.77	0.63
VGRF (BW)	2.41±0.24	2.34±0.23	2.44±0.25	0.50

*Key: °=degrees and N.m/kg=Normalised knee moment * significant differences at p=0.05*

6.4.2.2 Kinematic and kinetic differences between injured and non-injured limbs before, and three and six months after ACLR during running:

The kinematic and kinetic results are shown in Table 6.5. The peak knee flexion angle, and peak internal knee extensor moments and impulse, demonstrated significant reductions in the injured limb compared to the non-injured limb three and six months after ACLR (*p*=0.01*, effect size=1.66, *p*= 0.01*; effect size 0.86, *p*=0.01*; effect size 2.27, and *p*=0.01*; effect size 2.40, *p*=0.01*; effect size 0.90, and 0.01*; effect size 0.90 respectively). The knee adduction moment showed a significant reduction in the injured limb compared to the non-injured limb before ACLR (p= 0.02; effect size 0.90). For all other kinematic and kinetic variables, the results show no significant differences in the ACL injury participants measured before, and three and six months after, reconstruction surgery.

Table 6.5: Kinematics and kinetics between injured and non-injured limbs differences.

Variable	Participant Group			*P*-Value
	Time	Injured Limb (N=6) *(Means±S.D)*	Non-Injured Limb (N=6) *(Means±S.D)*	
Hip Internal Rotation Angle (°)	Pre	4.65±6.44	11.45±7.32	0.16
	3 Months	6.29±6.34	6.58±6.84	0.90
	6 Months	5.37±7.72	4.49±7.24	0.75
Hip Internal Rotation Moment (Nm/Kg)	Pre	-0.59±0.21	-0.54±0.27	0.26
	3 Months	-0.62±0.10	-0.58±0.17	0.40
	6 Months	-0.66±0.08	-0.61±0.10	0.09
Hip Adduction Angle (°)	Pre	14.33±4.31	12.48±3.12	0.29
	3 Months	11.86±7.31	9.39±2.53	0.36
	6 Months	9.97±7.46	8.84±2.38	0.73
Hip Adduction Moment (Nm/Kg)	Pre	-2.05±0.48	-1.99±0.57	0.67
	3 Months	-1.79±0.53	-1.65±0.29	0.52
	6 Months	-1.76±0.29	-1.71±0.24	0.69
Knee Adduction Angle (°)	Pre	3.00±6.13	6.89±5.22	0.06
	3 Months	5.76±7.44	6.84±5.12	0.64
	6 Months	6.19±5.71	5.54±2.89	0.89
Knee Adduction Moment (Nm/Kg)	Pre	-0.62±0.16	-0.92±0.22	*0.02
	3 Months	-0.59±0.28	-0.66±0.39	0.47
	6 Months	-0.68±0.23	-0.64±0.31	0.78
Knee Flexion Angle (°)	Pre	46.85±7.11	49.69±5.51	0.16
	3 Months	44.05±5.53	52.38±4.16	*0.01
	6 Months	42.67±5.37	47.69±6.17	*0.01
Knee Extensor Moment (Nm/Kg)	Pre	2.54±0.66	2.99±0.54	0.12
	3 Months	2.30±0.42	3.28±0.44	*0.01
	6 Months	2.48±0.77	3.04±0.64	*0.01
Knee Extensor Impulse (Nm*S/Kg)	Pre	0.24±0.08	0.30±0.07	0.14
	3 Months	0.22±0.05	0.37±0.07	*0.01
	6 Months	0.25±0.10	0.34±0.10	*0.01
VGRF (Bw)	Pre	2.41±0.24	2.51±0.14	0.17
	3 Months	2.34±0.23	2.45±0.20	0.23
	6 Months	2.44±0.25	2.47±0.29	0.78

*Key: °=degrees and N.m/kg=Normalised knee moment * significant differences at p=0.05*

6.4.2.3 Kinematic and kinetic differences between the injured limb before, three and six months after ACLR and control group during running:

The kinematic and kinetic results are shown in Table 6.6. The peak internal knee extensor moment demonstrated a significant reduction in the injured limb for the ACLR group compared to the control group before, and three and six months after, ACLR (p=0.02; effect size 1.06, p=0.01; effect size 1.06, p=0.02; effect size 1.06 respectively). In addition, knee extensor impulse demonstrated a significant reduction in the injured limb for the ACLR group compared to the control group before and three and six months after ACLR (p=0.03; effect size 1.06, p=0.01; effect size 1.06 and p=0.05; effect size 1.03 respectively). For all other kinematic and kinetic variables, the results show no significant differences in the ACL injury participants measured before, and three and six months after, reconstruction surgery.

Table 6.6: Kinematics and kinetics differences between the injured and the control group

Variable	Time	Injured Limb (N=6) *(Means±S.D)*	Control (N=6) *(Means±S.D)*	*P* Value
Hip Internal Rotation Angle (°)	Pre	4.65±6.44	5.31±5.67	0.85
	3 Months	6.29±6.34	5.31±5.67	0.78
	6 Months	5.37±7.72	5.31±5.67	0.98
Hip Internal Rotation Moment (Nm/Kg)	Pre	-0.59±0.21	-0.62±0.12	0.76
	3 Months	-0.62±0.10	-0.62±0.12	0.95
	6 Months	-0.66±0.08	-0.62±0.12	0.55
Hip Adduction Angle (°)	Pre	14.33±4.31	9.29±4.12	0.06
	3 Months	11.86±7.31	9.29±4.12	0.47
	6 Months	9.97±7.46	9.29±4.12	0.84
Hip Adduction Moment (Nm/Kg)	Pre	-2.05±0.48	-1.56±0.65	0.17
	3 Months	-1.79±0.53	-1.56±0.65	0.53
	6 Months	-1.76±0.29	-1.56±0.65	0.52
Knee Adduction Angle (°)	Pre	3.00±6.13	8.78±5.79	0.12
	3 Months	5.76±7.44	8.78±5.79	0.45
	6 Months	6.19±5.71	8.78±5.79	0.45
Knee Adduction Moment (Nm/Kg)	Pre	-0.62±0.16	-0.75±0.53	0.59
	3 Months	-0.59±0.28	-0.75±0.53	0.52
	6 Months	-0.68±0.23	-0.75±0.53	0.77
Knee Flexion Angle (°)	Pre	46.85±7.11	47.27±3.46	0.89
	3 Months	44.05±5.53	47.27±3.46	0.25
	6 Months	42.67±5.37	47.27±3.46	0.10
Knee Extensor Moment (Nm/Kg)	Pre	2.54±0.66	3.66±0.80	*0.02
	3 Months	2.30±0.42	3.66±0.80	*0.01
	6 Months	2.48±0.77	3.66±0.80	*0.02
Knee Extensor Impulse (Nm*s/Kg)	Pre	0.24±0.08	0.37±0.13	*0.03
	3 Months	0.22±0.05	0.37±0.13	*0.01
	6 Months	0.25±0.10	0.37±0.13	*0.05
VGRF (Bw)	Pre	2.41±0.24	2.67±0.34	0.15
	3 Months	2.34±0.23	2.67±0.34	0.07
	6 Months	2.44±0.25	2.67±0.34	0.21

*Key: °=degrees and N.m/kg=Normalised knee moment * significant differences at p=0.05*

6.4.2.4 Kinematic and kinetic differences between the non-injured limb before, three and six months after ACLR and the control group during running:

The kinematic and kinetic results are shown in Table 6.7. The peak knee flexion angle demonstrated a significant increase in the non-injured limb for the ACLR group compared to the control group three months after ACLR (p=0.04; effect size 1.06). For all other kinematic and kinetic variables, the results show no significant differences in the ACL injury participants measured before, and three and six months after, reconstruction surgery.

Table 6.7: Kinematics and kinetics differences between the non-injured and control group

Variable	Participant Group			*P* Value
	Time	Control (N=6) *(Means±S.D)*	Non-Injured Limb (N=6) *(Means±S.D)*	
Hip Internal Rotation Angle (°)	Pre	5.31±5.67	11.45±7.32	0.13
	3 Months	5.31±5.67	6.58±6.84	0.73
	6 Months	5.31±5.67	4.49±7.24	0.83
Hip Internal Rotation Moment (Nm/Kg)	Pre	-0.62±0.12	-0.54±0.27	0.51
	3 Months	-0.62±0.12	-0.58±0.17	0.61
	6 Months	-0.62±0.12	-0.61±0.10	0.86
Hip Adduction Angle (°)	Pre	9.29±4.12	12.48±3.12	0.16
	3 Months	9.29±4.12	9.39±2.53	0.96
	6 Months	9.29±4.12	8.84±2.38	0.81
Hip Adduction Moment (Nm/Kg)	Pre	-1.56±0.65	-1.99±0.57	0.25
	3 Months	-1.56±0.65	-1.65±0.29	0.77
	6 Months	-1.56±0.65	-1.71±0.24	0.63
Knee Adduction Angle (°)	Pre	8.78±5.79	6.89±5.22	0.56
	3 Months	8.78±5.79	6.84±5.12	0.55
	6 Months	8.78±5.79	5.54±2.89	0.24
Knee Adduction Moment (Nm/Kg)	Pre	-0.75±0.53	-0.92±0.22	0.47
	3 Months	-0.75±0.53	-0.66±0.39	0.75
	6 Months	-0.75±0.53	-0.64±0.31	0.68
Knee Flexion Angle (°)	Pre	47.27±3.46	49.69±5.51	0.38
	3 Months	47.27±3.46	52.38±4.16	*0.04
	6 Months	47.27±3.46	47.69±6.17	0.88
Knee Extensor Moment (Nm/Kg)	Pre	3.66±0.80	2.99±0.54	0.12
	3 Months	3.66±0.80	3.28±0.44	0.32
	6 Months	3.66±0.80	3.04±0.64	0.17
Knee Extensor Impulse (Nm*S/Kg)	Pre	0.37±0.13	0.30±0.07	0.12
	3 Months	0.37±0.13	0.37±0.07	0.65
	6 Months	0.37±0.13	0.34±0.10	0.49
VGRF (BW)	Pre	2.67±0.34	2.51±0.14	0.30
	3 Months	2.67±0.34	2.45±0.20	0.18
	6 Months	2.67±0.34	2.47±0.29	0.28

*Key: °=degrees and N.m/kg=Normalised knee moment. * significant differences at p=0.05.*

6.4.3 Analysis of Single Leg Squats

To test the kinematic and kinetic outcome parameters for normal distribution, Kolmogorov-Smirnov was used, and the reported values indicate that all hip and knee kinematics and kinetics were normally distributed, apart from the hip internal rotation moment (p=0.01), which was found not to be normally distributed for all participants.

6.4.3.1 Kinematics and kinetics differences in injured limb before, three and six months after ACLR during SLS:

The kinematic and kinetic results are shown in Table 6.8. For all kinematic and kinetic variables, the results show no significant differences in the ACL injury participants measured before, and three and six months after, reconstruction surgery (N=6). The knee adduction angle and external knee adduction moment were increased in the ACL injury participants measured before and after ACLR (p=0.54, p=0.41 respectively) but this is not significant.

Table 6.8: Kinematics and kinetics between injured limb differences before and after ACLR.

Variable	Participant Group			*P* value
	Pre(n=6) *(Means±S.D)*	Post 3M(n=6) *(Means±S.D)*	Post 6M(n=6) *(Means±S.D)*	
Hip Internal Rotation Angle (°)	8.11±6.35	9.21±7.10	9.63±5.50	0.83
Hip Internal Rotation Moment (Nm/Kg)	-0.49±0.12	-0.49±0.10	0.49±0.10	0.97
Hip Adduction Angle (°)	16.24±5.44	14.13±4.39	12.20±6.93	0.18
Hip Adduction Moment (Nm/Kg)	-1.10±0.31	-0.97±0.25	-0.93±0.20	0.20
Knee Adduction Angle (°)	9.43±6.86	11.38±6.69	12.67±5.41	0.54
Knee Adduction Moment (Nm/Kg)	-0.38±0.11	-0.44±0.10	-0.43±0.09	0.41
Knee Flexion Angle (°)	81.72±3.38	76.17±10.96	76.52±8.87	0.24
Knee Extensor Moment (Nm/Kg)	1.54±0.22	1.51±0.28	1.51±0.20	0.88
VGRF (*BW)	1.11±0.03	1.09±0.04	1.10±0.02	0.46

*Key: °=degrees and N.m/kg=Normalised knee moment * significant differences at p=0.05*

6.4.3.2 Kinematic and kinetic differences between injured and non-injured limbs before, three and six months after ACLR during SLS:

The kinematic and kinetic results are shown in Table 6.9. The peak internal knee extensor moments demonstrated a significant reduction in the injured limb before ACLR (p= 0.01; effect size 1.39) when compared with the non-injured limb for the ACLR group. There was reduction in peak internal knee extensor moment for the injured limb when compared to the non-injured limb for the ACLR group three months after ACLR, but this is not significant (p= 0.06). There was a significant reduction in peak knee flexion angle for the injured limb when compared to the non-injured limb for the ACLR group three months after ACLR (p=0.04; effect size 0.70). For all other kinematic and kinetic variables, the results show no significant differences in the ACL injury participants measured before, and three and six months after, ACLR.

Table 6.9: SLS Kinematics and kinetics between injured and non-injured limbs differences.

Variable	Limbs			*P* value
	Time	Injured limb (n=6) *(Means±S.D)*	Non-Injured limb (n=6) *(Means±S.D)*	
Hip Internal Rotation Angle (°)	Pre	8.11±6.35	10.93±6.57	0.54
	3 months	9.21±7.10	7.34±6.49	0.44
	6 months	9.63±5.50	8.57±4.21	0.50
Hip Internal Rotation Moment (Nm/Kg)	Pre	-0.48±0.12	-0.47±0.08	0.69
	3 months	-0.48±0.10	-0.42±0.05	0.23
	6 months	-0.4919±0.10	-0.51±0.07	0.45
Hip Adduction Angle (°)	Pre	16.24±5.44	11.52±5.11	0.21
	3 months	14.13±4.39	11.69±4.70	0.49
	6 months	12.20±6.93	10.96±2.55	0.73
Hip Adduction Moment (Nm/Kg)	Pre	-1.10±0.31	-1.00±0.40	0.23
	3 months	-0.97±0.25	-0.87±0.18	0.49
	6 months	-0.93±0.20	-0.96±0.24	0.83
Knee Adduction Angle (°)	Pre	9.43±6.86	13.52±5.98	0.28
	3 Months	11.38±6.69	10.23±5.75	0.60
	6 Months	12.67±5.41	13.33±5.92	0.78
Knee Adduction Moment(Nm/Kg)	Pre	-0.38±0.11	-0.52±0.19	0.14
	3 Months	-0.44±0.10	-0.41±0.08	0.57
	6 Months	-0.43±0.09	-0.54±0.15	0.21
Knee Flexion Angle (°)	Pre	81.72±3.38	86.87±5.95	0.13
	3 months	76.18±10.97	83.28±8.73	*0.04
	6 months	76.52±8.87	78.93±12.25	0.57
Knee Extensor Moment (Nm/Kg)	Pre	1.54±0.22	1.81±0.14	*0.01
	3 months	1.51±0.28	1.76±0.19	0.06
	6 months	1.51±0.20	1.64±0.23	0.21
VGRF (*BW)	Pre	1.11±0.03	1.10±0.03	0.68
	3 months	1.09±0.04	1.10±0.03	0.80
	6 months	1.10±0.02	1.11±0.03	0.56

*Key: °=degrees and N.m/kg=Normalised knee moment * significant differences at p=0.05*

6.4.3.3 Kinematic and kinetic differences between the injured limb before, three and six months after ACLR and for the control group during SLS:

The kinematic and kinetic results are shown in Table 6.10. The peak internal knee extensor moments demonstrated a significant reduction when comparing the injured limb for the ACLR group against the control group six months after ACLR (p=0.05 effect size 1.14). There were reductions in peak internal knee extensor moment before, and three months after, ACLR, but these are not significant (p=0.08, p=0.08, respectively). For all other kinematics and kinetics, there are no significant differences between the injured limb for the ACLR group compared to the control group before and after ACLR.

Table 6.10: Kinematics and kinetics differences between injured limb and control group:

Variable	Groups			*P* value
	Time	Injured limb (n=6) *(Means±S.D)*	Control (n=6) *(Means±S.D)*	
Hip Internal Rotation Angle (°)	Pre	8.11±6.35	10.42±5.62	0.52
	3 months	9.21±7.10	10.42±5.62	0.75
	6 months	9.63±5.50	10.42±5.62	0.81
Hip Internal Rotation Moment (Nm/Kg)	Pre	-0.48±0.12	-0.47±0.11	0.76
	3 months	-0.48±0.10	-0.47±0.11	0.74
	6 months	-0.4919±0.10	-0.47±0.11	0.71
Hip Adduction Angle (°)	Pre	16.24±5.44	11.61±5.09	0.15
	3 months	14.13±4.39	11.61±5.09	0.37
	6 months	12.20±6.93	11.61±5.09	0.86
Hip Adduction Moment (Nm/Kg)	Pre	-1.10±0.31	-0.92±0.19	0.24
	3 months	-0.97±0.25	-0.92±0.19	0.67
	6 months	-0.93±0.20	-0.92±0.19	0.90
Knee Adduction Angle (°)	Pre	9.43±6.86	13.73±8.94	0.37
	3 Months	11.38±6.69	13.73±8.94	0.61
	6 Months	12.67±5.41	13.73±8.94	0.81
Knee Adduction Moment (Nm/Kg)	Pre	-0.38±0.11	-0.35±0.29	0.80
	3 Months	-0.44±0.10	-0.35±0.29	0.47
	6 Months	-0.43±0.09	-0.35±0.29	0.54
Knee Flexion Angle (°)	Pre	81.72±3.38	84.10±11.38	0.63
	3 months	76.18±10.97	84.10±11.38	0.24
	6 months	76.52±8.87	84.10±11.38	0.22
Knee Extensor Moment (Nm/Kg)	Pre	1.54±0.22	1.82±0.26	0.08
	3 months	1.51±0.28	1.82±0.26	0.08
	6 months	1.51±0.20	1.82±0.26	*0.05
VGRF (*BW)	Pre	1.11±0.03	1.12±0.01	0.53
	3 months	1.09±0.04	1.12±0.01	0.17
	6 months	1.10±0.02	1.12±0.01	0.14

*Key: °=degrees and N.m/kg=Normalised knee moment * significant differences at p=0.05*

6.4.3.4 Kinematic and kinetic differences between the non-injured limb before, three and six months after ACLR and the control group during SLS:

The kinematic and kinetic results, and the statistical analysis results, are shown in Table 6.11. For all kinematic and kinetic variables, the result revealed no significant difference between the non-injured limb for the ACLR group compared to the control group before, and three and six months after ACLR, during SLS. The non-injured limb group showed a reduced peak internal knee extensor moment compared to the control group. However, this is also not significant.

Table 6.11: Kinematics and kinetics differences between the non-injured limb and control group.

Variable	Groups			*P* value
	Time	Control (n=6) *(Means±S.D)*	Non-Injured limb (n=6) *(Means±S.D)*	
Hip Internal Rotation Angle (°)	Pre	10.42±5.62	10.93±6.57	0.88
	3 months	10.42±5.62	7.34±6.49	0.40
	6 months	10.42±5.62	8.57±4.21	0.53
Hip Internal Rotation Moment (Nm/Kg)	Pre	-0.47±0.11	-0.47±0.08	0.93
	3 months	-0.47±0.11	-0.42±0.05	0.37
	6 months	-0.47±0.11	-0.51±0.07	0.38
Hip Adduction Angle (°)	Pre	11.61±5.09	11.52±5.11	0.97
	3 months	11.61±5.09	11.69±4.70	0.97
	6 months	11.61±5.09	10.96±2.55	0.78
Hip Adduction Moment (Nm/Kg)	Pre	-0.92±0.19	-1.00±0.40	0.67
	3 months	-0.92±0.19	-0.87±0.18	0.66
	6 months	-0.92±0.19	-0.96±0.24	0.75
Knee Adduction Angle (°)	Pre	13.73±8.94	13.52±5.98	0.96
	3 months	13.73±8.94	10.23±5.75	0.43
	6 months	13.73±8.94	13.33±5.92	0.93
Knee Adduction Moment (Nm/Kg)	Pre	-0.35±0.29	-0.52±0.19	0.25
	3 Months	-0.35±0.29	-0.41±0.08	0.65
	6 Months	-0.35±0.29	-0.53±0.15	0.20
Knee Flexion Angle (°)	Pre	84.10±11.38	86.87±5.95	0.60
	3 months	84.10±11.38	83.28±8.73	0.89
	6 months	84.10±11.38	78.93±12.25	0.46
Knee Extensor Moment (Nm/Kg)	Pre	1.82±0.26	1.81±0.14	0.95
	3 months	1.82±0.26	1.76±0.19	0.69
	6 months	1.82±0.26	1.64±0.23	0.24
VGRF (*BW)	Pre	1.12±0.01	1.10±0.03	0.24
	3 months	1.12±0.01	1.10±0.03	0.21
	6 months	1.12±0.01	1.11±0.03	0.57

*Key: °=degrees and N.m/kg=Normalised knee moment * significant differences at p=0.05*

6.5 Results Summary

The results suggest that there are no significant differences at the hip joint before and after ACLR during running and SLS between limbs (injured and non-injured), and between the ACLR group and healthy control group. Hence, the first, second, third and fourth null hypotheses are accepted. There were significant differences noted in the peak knee flexion angle, extensor moment, and extensor impulse at three and six months during running between limbs (injured and non-injured). This suggests significant differences in knee joint sagittal plane movements. Hence, the fifth null hypothesis has been partially rejected. However, there were no significant differences in the frontal planes movements before and after ACL reconstruction during running between limbs (injured and non-injured). As shown in the results above, there were no significant differences in the knee adduction angle and moment. Hence, null hypothesis five has been partially accepted. There were significant differences in the sagittal planes movements before and after ACL reconstruction during SLS between limbs (injured and non-injured), as exemplified by significant differences in peak knee flexion angle and extensor moment at baseline and at three and six months after ACL. Hence, null hypothesis six has been partially rejected. However, there were no significant differences in frontal planes during SLS, as demonstrated by the lack of significant differences between the knee adduction angle and moment. Hence, null hypothesis six has been partially accepted.

Null hypothesis seven has been partially rejected, since there will be differences in knee joint sagittal planes movement before and after ACL reconstruction during running between the ACLR group and the healthy control group. Specifically, there were significant differences in the peak knee flexion angle, extensor moment, and impulse. However, there were no significant differences in the knee frontal plane movements. Hence, null hypothesis seven has been partially accepted. Null hypothesis eight is partially rejected since there were significant differences in the peak internal knee extensor moment and impulse between the ACLR group and healthy control group at baseline, and at three and six months after ACL reconstruction. This means that there were significant differences in the knee joint sagittal plane movements during SLS. However, there were no significant differences in the frontal plane movements before and after surgery between the ACL group and the healthy control group before and after ACL surgery. Hence, null hypothesis eight has been partially accepted.

6.6 Discussion

This pilot study has investigated running and SLS tasks in individuals pre- and post-ACL reconstruction. To the author's knowledge, this is the first study that has compared individuals during these activities in this subject group. In this chapter, the results have been presented at the beginning, and the following section will discuss those results. It is felt that these results will help inform readers about whether impaired knee function can increase the risk of patellofemoral (PF) and tibiofemoral (TF) osteoarthritis following ACL injury and repair.

6.6.1 Kinematics and Kinetics of Patients during Running and SLS

As identified in the previous chapter, players who return to their sporting activities six to eight months after ACL reconstruction have a deficit in their peak internal knee extensor moment and peak knee flexion angle. This observation has been confirmed in the present small pilot study. The results of the present study have identified that knee kinetic outcomes for the ACLR group showed a reduction in the peak internal knee extensor moment and impulse after three and six months of ACLR between the non-injured and injured limbs. A reduction in the knee flexion angle will decrease the patellofemoral contact area. In turn, this increases patellofemoral contact stress and decreases knee extensor moment can lead to decrease the dynamic mobility of the patella and lead to the development of PFPS (Boling et al., 2009). This is also confirmed by Butler et al., (2009) who emphasise that PF osteoarthritis risk is increased with changes in the dynamic mobility of the patella and the presence of PFPS. In the present study, participants may have reduced their knee flexion in order to reduce patellofemoral compressive forces and subsequently reduce any pain (Dierks et al., 2010). It is hypothesised that a shift in knee flexion angle likely affects the location of tibiofemoral joint contact and patellofemoral contact stress (Hall et al., 2012; Boling et al., 2009).

As far as this study is concerned, symptoms of pain and ADL and patients reporting acceptable symptoms following ACLR were within the norms of the population pertaining to Sport/Rec and QOL. According to Hartigan et al., (2013), kinesiophobia levels were elevated in patients prior to ACL reconstruction; this was predominant in those with poorer dynamic knee stability/ non-copers. Subsequent to ACL reconstruction, it was noted that kinesiophobia levels reduced the most in non-copers (Hartigan et al., 2013). It was also noted that reductions in kinesiophobia have a significant relation to improvements in self-reported knee function during ADL (Hartigan et al., 2013). As far as the clinical scenario is concerned, kinesiophobia levels were reported to be high at six months, and plateaued between six and 12 months post-surgery (Hartigan et al., 2013). It should also be emphasised that kinesiophobia levels should be monitored from the time of ACL rupture to 12 months after surgery (Hartigan et al., 2013).

The current study also found significant increases in peak knee flexion angle on comparing with the non-injured limb in the ACLR group and the control group. The study conducted by Pahnabi et al., (2014) compared the uninjured legs of the experimental group with the healthy legs of the control group. Overloading of the uninjured leg was reported in individuals with ACLR knees, therefore leading to increased stress on the uninjured leg. Bonfim et al., (2008) report a reduction in the neural signal transmissions in the injured leg due to ACL injury; this paved the way towards motor control system malfunctioning, whereby the performance of the uninjured leg was adversely affected. Bonfim et al., (2008) also report changes in performance when ACL group respondents used their uninjured legs for locomotion. According to Jones et al., (2012), the risk of OA is high for the contralateral knee, irrespective of the occurrence of the injury. Shakoor et al., (2002) found a high risk of OA in the contralateral knee for patients undergoing a unilateral hip replacement.

Abnormal gait patterns might have consequences in the long term, leading to pathologic knee conditions (Georgoulis et al., 2003; Noyes et al., 1992). There is little data on why individuals with an ACL-deficient (ACLD) or ACL-reconstructed (ACLR) knee consequently develop OA (osteoarthritis), but altered kinematics and kinetics are one of the hypothesised reasons (Chaudhari et al., 2008).

Previous investigations have examined biomechanical alterations separately during running in ACLD (Lewek et al., 2002) and ACL-reconstructed limbs (Lewek et al., 2002; Karanikas et al., 2009; Bush-Joseph et al., 2001). They have concluded that individuals with ACL injury ambulate during running with less flexed knee joint angles (Lewek et al., 2002; Karanikas et al., 2009), and exhibit more abducted knee joint angles during walking (Gao et al., 2012). The current study detected less flexed knee joints in ACL injured limbs in the absence of kinematic alterations in the frontal plane during running. A shift in the tibiofemoral contact pattern could be detrimental to long-term joint health, as this would lead to abnormal load distribution on the articular cartilage (Andriacchi et al., 2004). As such, it seems plausible that the cumulative effects of small alterations in knee joint flexion angles over time, whether it is at the peak range of motion or not, may have detrimental effects on knee joint degeneration. Clinically, however, it remains unclear how these relatively small alterations in knee flexion angle during the stance phase contribute to post-traumatic knee joint degeneration.

In the present pilot study, ACL injured patients had reduced activity levels based on their KOOS SR subscale scores compared to the control group. However, following ACL reconstruction, activity levels improved, suggesting that ACL injured patients went on to engage in increased activities and had been subjected to increase load in their knee joints. However, this increased load may affect knee kinematics. The present study suggests that these alterations in peak knee flexion angle occur mainly during running, which has been shown to be a more demanding biomechanical task than walking (Bush-Joseph et al., 2001). The increased load might explain the small alterations in knee flexion angle during the stance phase. Although this is the first investigation to track patients prospectively, before and after ACL reconstruction, it is clear that more research is needed to fully understand the extent of these kinematic and kinetic alterations and, ultimately, what effect they have on post-traumatic joint degeneration.

An interesting finding from this study suggests that patients with an ACL injury use less knee flexion angle throughout the stance phase of running. It is likely that a decrease in knee flexion angle initiates abnormal movement patterns proximal in the kinetic chain, which in turn might affect hip joint motion. However, in this study, increased frontal plane hip joint range of motion was not noted. Lower knee flexion angle have also been observed during walking in patients with knee osteoarthritis (Childs et al., 2004). In another study (Goerger et al., 2014), patients with osteoarthritis demonstrated increased frontal hip motion when performing dynamic tasks. It is suggested that increased frontal hip motion will help individuals with an ACL injury to create new strategies to complete dynamic tasks such as cutting (Goerger et al., 2014). In the present study, the reduction in sagittal plane motion in the knee might be a compensatory strategy in response to the ACL injury and subsequent lack of quadriceps strength. Furthermore, alterations in the sagittal plane motion of the knee in the current set of patients may contribute towards the development of post-traumatic osteoarthritis over time; however, more research is needed to determine the influence these biomechanical alterations have on the progression of joint degeneration.

Participants in the ACLR group showed improved KOOS and IKDC scores from baseline until the three and six month follow-up periods, which is indicative of normal knee function (Anderson et al., 2006; Grindem et al., 2011). When compared to the results of Grindem et al., (2011), the IKDC run-performance scores of the current patients at six months were similar to ACLD patients at one-year post injury. It is suggested that patients have to wait for nine months to gain maximum functional recovery following ACL repair (Ardern et al., 2011; Paterno et al., 2010). In the current set of patients, the median IKDC score was 91, which suggests that patients were able to attain maximum functional recovery earlier than patients in other studies. Meanwhile, the KOOS subscales were similar to the study of Lynch et al. (2015).

In the present study, pain during running was recorded based on the KOOS subscale on pain. The results show that the mean pain scores of the injured limb of the ACLR group was 77.83 (S.D. ±20.66) at baseline, 90.81± at three months and 96.76 ± 2.27 at six months. The presence of pain in the pilot study group is an important predictor of the risk of PF OA. In the study by Wyndow et al., (2016), the presence of patellofemoral (PF) pain in adolescents and young people has been associated with the development of PF joint OA. A possible explanation for this relationship includes the altered neuromotor control and biomechanical factors. The altered sagittal plane motions of the knees in the current set of patients are some of the biomechanical factors that might increase the risk of these patients suffering PF OA later in life. In addition, Culvenor et al., (2016) demonstrated that patients who exhibit distinct kinematic and kinetic features during a high-load landing task are more likely to develop early PF OA within the first two years following repair of the ACL. The findings of Culvenor et al., (2016) have important implications for the present study, since common biomechanical patterns might be associated with PF OA. This suggests the need to employ management strategies, which include altering the individual's knee load during the early stages of the disease, in order to reduce the risk of early PF OA (Culvenor et al., 2016).

Similar to previous investigations, smaller knee joint extensor moments were observed in the ACL-injured limb compared to those in the non-injured limb (Sigward et al., 2015; Karanikas et al., 2009) and healthy individuals (Lewek et al., 2002; Kuenze et al., 2014; Bush-Joseph et al., 2001) during running. This reduction in knee extensor moment is thought to result from persistent deficits in quadriceps strength, which are commonly observed in patients with injured and reconstructed ACL (Ingersoll et al., 2008). Weakness of the muscle causes a decrease in the patient's ability to produce enough quadriceps force to eccentrically control the limb through the entire range of motion (Andriacchi et al., 2004). Furthermore, smaller moments in the involved limb may develop because of planned biomechanical adaptations, where the individual with an ACL injury attempts to reduce the stress on the injured joint and avoid painful movements related to compressive forces (Sturnieks et al., 2008). Previous investigations have observed reduced knee extensor moments at times greater than six months after reconstruction (Kuenze et al., 2014; Noehren et al., 2013; Karanikas et al., 2009; Lewek et al., 2002; Bush-Joseph et al., 2001), and this is the first investigation to identify these alterations before and after reconstruction in the same patients during running. Therefore, it is likely that changes in knee joint function develop in response to the acute ACL rupture, and that reconstructive surgery and traditional rehabilitation does not appear to restore normal biomechanics during daily tasks in these patents. Hence, there is a need to investigate rehabilitation strategies, such as hip strengthening, as that may improve the biomechanics of individuals post ACL reconstructive surgery. At present, deficits in the strength of the hip and knee joints post-ACLR may explain why normal biomechanics are not restored. Although recent evidence suggests that baseline movement patterns are altered after injury and reconstruction (Goerger et al., 2014), further work in the area of ACL injury risk and prevention should aim to prospectively evaluate lower extremity biomechanics during a variety of tasks.

Investigators have continued to discover higher knee extensor joint moments in the non-injured ACL limb (Sigward et al., 2015; Kuenze et al., 2014; Karanikas et al., 2009), which seems to warrant an investigation into the rates of contralateral ACL injury and post-traumatic osteoarthritis of the contralateral limb. The unloading of the injured joint, whether due to quadriceps weakness or pain avoidance, may place undue stress on the contralateral limb, creating a scenario for injury to the uninvolved limb. Rates of osteoarthritis in the contralateral limb are not investigated as regularly as post-traumatic osteoarthritis in the injured limb; however, rates of contralateral limb osteoarthritis are reported to be approximately 20%–30% (Barenius et al., 2014; Murray et al., 2012), which is significantly less frequent than reports of ipsilateral post-traumatic osteoarthritis of 50%–60% (Barenius et al., 2014). However, the findings of Barenius et al., (2014) should be considered with caution, since 50% of the patients who were followed-up might have developed osteoarthritis. It is noteworthy that not all patients were followed-up in the study. Arguably, those that fail to reply during follow-ups are likely to do so since they may have no residual issues, and thus the percentage of people with OA is perhaps far lower than reported.

Although instinctively counterintuitive, it has been suggested that the unloading observed in the injured limb may actually contribute to these high rates of osteoarthritis, as unloading previously loaded articular cartilage may negatively affect joint health by changing the biochemical composition of the cartilage and compromising its structure (Arokoski et al., 2000; Chaudhari et al., 2008). Similarly, less is known regarding the rates of contralateral ACL rupture, however reported rates of contralateral ACL injury range from 5% to 30% (Hettrich et al., 2013; Paterno et al., 2012). It is also important to note that knee joint asymmetry is suggested to contribute towards secondary ACL rupture (Paterno et al., 2012) and posttraumatic osteoarthritis (Oiestad et al., 2010). The asymmetrical shift in biomechanics after injury and reconstruction may ultimately negatively affect joint health and stability in these patients. More research is needed to understand the clinical reason for patients with an ACL injury shifting joint loads to the uninvolved limb, and how this adaptation affects bilateral long-term joint health.

The return to high level dynamic activities on a reconstructed, but not normal, knee may increase loads on the knee joint, which over time can lead to joint degeneration and osteoarthritis (Andriacchi et al., 2004; Svoboda 2014). However, little is known regarding the cumulative effects of small biomechanical alterations that are potentially adopted during highly demanding activities, such as running and SLS. The patients with an ACL injury in this study experienced biomechanical differences to the control group before surgery, potentially indicating that these alterations develop early in the injury process. Furthermore, traditional reconstructive surgery and therapeutic rehabilitation did not restore these biomechanical alterations to normal levels. This is in agreement with previous investigations evaluating level ground walking, which found early biomechanical responses to injury (DeVita et al., 1998). The lack of any gait retraining or biofeedback interventions during the current rehabilitation protocol stand out as areas for which further consideration is warranted, to help correct these abnormal gait patterns. Therefore, it is possible that patients would benefit from early, pre-surgery gait retraining interventions to correct abnormal gait adaptations before they become permanent movement patterns. Gait retraining programmes have shown varying levels of success in patients with reconstructed ACL (Decker et al., 2004); however, it is possible that after reconstruction, poor biomechanical strategies have already been adopted. Therefore, clinicians may have better success if they institute gait-retraining interventions before surgery, and training with a variety of low and high demand activities, to help limit the development of deleterious biomechanical strategies. However, gait retraining interventions alone may not restore normal biomechanics, and may need to be supplemented with neuromuscular control and strengthening programmes to help combat the deficits in muscle function common in this population. Furthermore, outside of advancements in therapeutic rehabilitation, improvements in surgical procedures must also be evaluated in an attempt to minimise joint and limb asymmetry in both structure and function.

There is good evidence that a high proportion of individuals with ACL injury go on to develop disabling OA within 10 years of their injury (Georger et al., 2014). When joint biomechanics are affected following ACL repair, the risk of developing PF and TF osteoarthritis are increased. Determining abnormal movement patterns at the hip joint following ACL injury and repair is crucial, since this has the potential to influence joint biomechanics, both proximal and distal to the injury (Goerger et al., 2014). The present study did not find any significant increase in hip frontal and transverse plane movements. An increase in hip frontal hip motion is often observed in patients with reconstructed ACL during landing tasks (Georger et al., 2014). This is in contrast to the findings of the current study where increased hip motion was not observed. Increased strength at the hip will play a role in allowing patients to utilise more motion at the hip joint, creating a new strategy in which they will be able to complete landing tasks (Georger et al., 2014). It is noteworthy that all participants who had undergone ACL repairs in the study underwent hip strengthening as part of standard rehabilitation protocols. This strategy has been previously used by clinicians to improve knee joint biomechanics, or help correct abnormal biomechanics following knee joint surgery (Ferber et al., 2011; Powers 2010). However, some data suggests that hip strength is not associated with hip and knee joint biomechanics during walking (Bolgla et al., 2008) (specifically in patellofemoral pain syndrome patients), making it unclear whether the post-surgical rehabilitation positively affected hip biomechanics in the current set of patients.

6.6.2 PF and TF Osteoarthritis

Limiting the contact area over patellofemoral loads would lead to increased contact stress with small areas of patellofemoral contact. Also, using smaller knee flexion angles during high demand tasks may increase the risk of early patellofemoral disease progression. The increased risk of cartilage damage potentially follows more localised areas of contact stress. Smaller knee flexion angles may represent an alternative pain adaptation. When PFOA is present, greater flexion angle is linked with compressive forces of the patellofemoral joint. However, the present study observed lower peak internal knee extensor moment and less knee flexion for individuals with ACLR, which is not consistent with greater flexion angles in PFOA (Teng et al., 2015). This lower knee flexion may be an adaptation of the participants in the present study against pain during running or SLS. Individuals with ACLR show movement patterns that are distinct, and this could contribute towards respondents reporting symptoms of stiffness (Li et al., 2004). According to Teng et al., (2015), when cartilage lesion of the patellofemoral is assessed by MRI technology, which is considered to be patellofemoral disease at an early stage for individuals described as older, patellofemoral stress and flexion moment show a higher peak. However, these findings are not supported by this current study, which has found that individuals with ACLR were observed to display movement patterns at the sagittal plane that were lower. This current study focused on running during the first part of the loading phase for assessment purposes to understand the biomechanics involved, which differs from the study by Teng et al., (2015), who report a focus on contralateral limb transference of weight during test exercises for gait in the second part of the stance phase.

Quadriceps weakness is reported to be associated post-ACLR with lower knee extensor moments, because quadriceps strength deficit is linked, during running, to lower knee flexion angle and knee extensor moment (Lewek et al., 2002). Other findings suggest that individuals with chondral lesions of the patellofemoral that are post ACLR are linked to weakness of the quadriceps (Wang et al., 2015). Therefore, individuals resort to increased knee flexion angles as a learned strategy to be able to run without pain. However, this requires increased strength of the quadriceps to absorb the shock from running. Individuals with ACL injuries have a greater risk of developing TFOA and PFOA, since active mechanisms for absorbing shock are lower (Duncan et al., 2008). When compared with the present study, knee flexion angles were significantly lower in the reconstructed limb compared to the non-injured limb. However, the observations during running were carried out a few times. It is also important to note that in the present study, there was increased knee flexion angle at three months during running and significantly increased peak internal knee extensor moment before ACLR during SLS in the contralateral limb. These changes may be an adaptation due to the decreased knee function or sagittal plane movements of the injured limb. Kobayashi et al., (2016) suggest that PFOA has high prevalence in both symptom-based and population-based cohorts. Importantly, the presence of PFOA was also associated with the development of TFOA. Risks of developing radiographic TFOA and PFOA could be lower if quadriceps strength can alter the movement of the sagittal plane. However, this current study has not evaluated the factor of quadriceps strength. Various findings suggest that when individuals present quadriceps weakness, they are also likely to be at higher risk of degenerative changes, at a future date, to their patellofemoral and tibiofemoral, which is a significant concern (Amin et al., 2009; Oiestad et al., 2015).

In the present study, the link between running and SLS is based on the observation of Whatman et al., (2011) and Lewis et al., (2015), which suggests that SLS can predict running outcomes. In the present study, the SLS task allowed the participants to increase knee abduction (dynamic valgus), whilst the step-down task was performed with more hip adduction. Increased hip adduction has been linked to PFPS, and increased knee valgus angle to ACL injury (Willson and Davis 2008; Noehren et al., 2013; Hewett 2005). Hence, the findings of the present study suggest that participants with ACL reconstruction have an increased risk of PFPS, which in turn, might increase risk of PFOA.

There are important limitations to the present study. It was difficult to establish whether there were significant differences in the mean values of the knee valgus, knee adduction moment, hip adduction moment, hip internal rotation moment, and dorsiflexion moment in the injured limb of the ACLR group and control group, due to the very small sample size (n=6 in the ACLR group and n=6 in the control group). The sample size calculation used the difference between the means and standard deviation of six ACLR and six control group participants during running. This indicated that at least 18 participants in the ACLR group, and another 18 participants in the control group, are needed for the present study to achieve a power of 90% and detect an effect size of 0.36, which was calculated using the partial eta squared =0.120, with statistical significance deemed to be α=0.05. Therefore, the sample size is insufficiently powered to detect an effect size. Despite this limitation, the study provides baseline data on reported knee function and kinematics and kinetics during running and single leg squat following ACL repair. The lack of statistically significant differences in some of the variables in the present study should be taken with caution due to the small sample size of the present study. Furthermore, variables that demonstrated significant differences between the ACLR group and control group exceeded the standard error of measurement (SEM). This suggests that the results need to be verified in a larger trial to be conclusive. The parametric t-test was also used several times to compare means between groups, which may have increased the Type 1 error.

6.7 Conclusion

As identified in the previous chapter, players who return to their sporting activities six to eight months after ACL reconstruction, have a deficit in their peak internal knee extensor moment and peak knee flexion angle. This observation has been confirmed in the current small pilot study. This study has revealed that before and three and six months following ACL reconstruction, the athletes in this study showed some specifically altered knee joint kinematics and kinetics. The reduction in knee extensor moment and knee flexion angle was in an effort to reduce or avoid contraction of the quadriceps; this is called ‘quadriceps avoidance’. However, these reductions will decrease the patella contact area and this will increase the patellofemoral contact stress over time; therefore, repetitive movements may contribute towards patellofemoral disorders, thereby increasing the risk of degenerate joint disease, which is commonly found post-surgery. The results of this study may help to guide the development of new or alternative treatment options for improving long-term joint health after an ACL injury and after ACLR, by adding gait retraining to rehabilitation programs before and after ACLR to correct the biomechanical alterations (feedback training). The patellofemoral cartilage’s long-term life could be increased if biomechanics that are abnormal, and their connection with neuromuscular deficits, are used to inform rehabilitation strategies for patients, as PFOA development exposes the risk factors of reduced knee extensor moment or knee flexion angle. This may be confirmed by larger prospective studies.

CHAPTER SEVEN

SUMMARY

7.1 Summary

There is little data on why individuals with an ACL-deficient (ACLD) or ACL-reconstructed (ACLR) knee go on to develop osteoarthritis (OA) of both the tibiofemoral and patellofemoral joints, but altered kinematics and kinetics at the knee are one of the hypothesised reasons. Numerous studies investigating walking gait have found significant reductions in peak internal knee extensor moment and small reductions in peak knee flexion angle, with individuals adopting a quadriceps avoidance gait pattern during walking. This may then increase the risk of the development of degenerative diseases due to changes in loading patterns, so understanding if there are any differences is of utmost importance. One of the most common activities pre- and post-surgery is running, and it is not known if individuals before and after ACLR knee have different knee kinematic and kinetic patterns compared to healthy individuals. Currently, there is no research detailing kinematics and kinetics before and after ACL reconstruction during running. Therefore, the overall purpose of this study was to compare the knee kinematics and kinetics before and after ACL reconstruction during running.

The most widely adopted system used to analyse the motion of the human body during complex movements across three dimensions is 3D analysis systems. The use of these systems allows researchers and clinicians to quantify mechanical factors when investigating knee injuries and the biomechanics of the lower limb. Assessments used for clinical or research purposes should be reliable and valid both within session, and also after a period of time, to justify its use as an assessment technique. Therefore, measurement techniques need to recognise associated errors in measurements and the reliability of the data collected to expose whether differences or changes in performance are due to actual performance changes or to errors in measurements. This will help in achieving accurate assessments, which is essential, as the values of measurement errors and knowledge of reliability are critically important. Standard error of measurement (SEM) allows researchers to determine whether the changes or improvements are more than the measurement error of the assessment, while minimal detectable change (MDC) values reveal if any changes in a specific variable across time are due to real performance changes. This ensures that the analysis of data makes allowances for measurement errors through effective research techniques to identify the values of reliability and values of measurement errors.

The reason for this is that before and after ACL reconstruction, the patient may have different kinematics and kinetics, which results in different performance abilities during running, and may have consequences for long-term rehabilitation and return to sport for the individual. Currently, there is no research detailing kinematics and kinetics before and after ACL reconstruction during running and SLS. Therefore, the research question focuses on determining whether there is an alteration in the kinematics and kinetics of the hip and knee joints, or related risk factors for patellofemoral pain syndrome and OA, before and after ACLR during running and SLS.

This has been answered through the following objectives:

1. Investigate the reliability of using a 3D motion-analysis system to measure lower-limb kinematic and kinetic variables during running and single-leg squat (SLS) tasks.
2. To investigate the hip joint frontal and transverse plane movements six to eight months after ACL reconstruction during running and SLS between limbs (injured and non-injured).
3. To compare the hip joint frontal and transverse plane movements six to eight months after ACL reconstruction with a healthy control group during running and SLS.
4. To investigate the knee joint sagittal and frontal plane movements six to eight months after ACL reconstruction during running and SLS between limbs (injured and non-injured).
5. To compare the knee joint sagittal and frontal plane movements six to eight months after ACL reconstruction with a healthy control group during running and SLS.
6. To investigate the hip joint frontal and transverse plane movements before and after ACL reconstruction during running and SLS between limbs (injured and non-injured).
7. To compare the hip joint frontal and transverse plane movements before and after ACL reconstruction with healthy control group during running and SLS.
8. To investigate the knee joint sagittal and frontal plane movements before and after ACL reconstruction during running and SLS between limbs (injured and non-injured).
9. To compare the knee joint sagittal and frontal plane movements before and after ACL reconstruction with a healthy control group during running and SLS.

7.2 Conclusion

With regard to the first aim, this was to establish the within-day and between-days reliability for using 3D motion analysis to measure the biomechanical variables collected from running and single leg squat tasks. This study concludes that for between and within-day sessions, specific variables demonstrated good and excellent levels of consistency, and exhibited standard errors of measurement that have relatively low values. This is the first study that has shown repeatability with the cluster model adopted for this thesis. The data has also shown the repeatability of the marker set and kinematic and kinetic data at three different sessions, which enabled the collection of kinematic and kinetic data at multiple time points for an individual before and after ACL reconstruction to be taken forward into the main study.

In order to achieve the second, third, fourth and fifth aims of this thesis, 34 patients six to eight months after ACLR, and 34 healthy participants, were recruited in order to investigate the hip and knee joint kinematics and kinetics during running and a SLS task to compare between the injured limb and the contralateral limb and a control group. This study found that knee kinetic outcomes within the ACLR group showed that there was a reduction in the peak internal knee extensor moment and impulse. This reduction was found to exist even when the ACLR group was compared to the control group. At the same time, between the ACLR and control group, there was a significant increase in external peak knee adduction moment during running, as well as SLS between limbs. The kinematic assessment showed that there was a significant difference, with an increase in hip internal rotation angle between limbs during running as well as SLS. Although there was a reduction in peak knee flexion angle for running between limbs and the control group, for SLS, there was a significant increase in the knee adduction angle between limbs.

The lower peak knee flexion angles or moments and impulse observed during running and SLS might also reflect deficits in quadriceps strength, as low internal knee extensor moments post-ACLR have been linked to quadriceps weakness. Quadriceps weakness has also been associated with post-ACLR patellofemoral chondral lesions. With greater strength, the quadriceps can more effectively attenuate the shock from running by producing a ‘soft-contact’ strategy via increased knee flexion angles. Lower active shock absorbing mechanisms in those with ACLR may also adversely affect the tibiofemoral joint over time due to high axial compression, which is consistent with the pattern of PFOA increasing the risk of TFOA in atraumatic patients. Although quadriceps strength was not evaluated in the current study, the altered sagittal plane movement patterns suggest that addressing deficits in movement control and strength may potentially lower the risk of radiographic PFOA and TFOA development. This is particularly pertinent given the elevated risk of both future patellofemoral and tibiofemoral degenerative changes identified in the presence of quadriceps weakness. The lower peak knee flexion angle or internal knee extensor moment and impulse, are risk factors for PFOA development, therefore the patellofemoral cartilage could have an extended life period if biomechanics that are abnormal are related to appropriate strategies for rehabilitation for patients that successfully overcome neuromuscular deficits. The TF joint experiences changes that are degenerative, and impulse loading that is harmful, due to the knee absorbing shock which when normally active is reduced, therefore PF joint pain and joint compression is decreased as a result of strategies that compensate for these factors.

ACLR individuals can experience function and pain improvements that are significant during running tasks when peak external knee adduction moment is increased. The sample population in this current study indicates that the onset of knee OA at an early stage could result from the mechanism at the tibiofemoral joint, when the medial compartment experiences loads that are increased. This is because patients have a higher risk of developing knee OA when they present a peak external knee adduction moment that has increased following ACL reconstruction surgery. This is a clinically meaningful finding from this current study as differences in peak internal knee extensor moment have been observed. Early onset knee OA incidence could be reduced if patients are regularly examined after an ACL rupture to decelerate the progression of knee OA through the selection of treatment options that are effective, and based on research studies, over the longer term. Many patients with an ACL rupture wish to return to their sport and previously attained activity levels, which should also influence the care plan for these patients to help them maintain joint integrity over the long term, based on the findings of this current study.

Regarding the sixth, seventh, eighth and ninth objectives of this research, for the prospective study, six patients before and three and six months after ACLR, and six healthy participants, were recruited. This allowed an investigation of the hip and knee joint kinematics and kinetics during running and a SLS task to compare between the injured limb and contralateral limb and the control group. The findings clearly demonstrate that there were significant hip and knee kinetic outcomes within the ACLR group, and that there was a reduction in the peak internal knee extensor moment and impulse, three and six months after ACLR between the limbs. In addition, when comparing the injured limb for the ACLR group to the control group, significant differences were noted before and after three and six months of ACLR during running, as well as SLS between limbs before and after three months of ACLR. At the same time, within the ACLR group, there was a significant reduction in peak knee flexion angle during running three and six months after ACLR between limbs. This reduction was found to exist even when the non-injured limb ACLR group was compared to the control group three months after ACLR; although there was an increase in knee valgus angle between the limbs before ACLR during SLS.

Smaller knee flexion angles during high demand tasks may increase the risk of early patellofemoral disease progression by limiting the contact area over which large patellofemoral loads can be distributed. This deficit potentially creates more localised areas of contact stress, increasing the risk of cartilage damage. Alternatively, smaller knee flexion angles may represent a pain adaptation. However, this is unlikely given that no participant reported any knee pain during the running task.

The lower peak internal knee extensor moment was observed in the current study in those with ACL injuries. These movement patterns may reflect a compensatory strategy to limit high patellofemoral joint compressive forces associated with greater flexion angles. A greater resultant muscle co-contraction (i.e., apparent stiffness) may also contribute to the distinct movement patterns observed in those with ACL injuries, particularly as simulated elevated antagonist hamstring activation appears detrimental to patellofemoral biomechanics.

To sum up, patients suffering from ACL injuries may exhibit reduced loading response knee flexion angles and lower single-leg stance peak internal knee extensor moments during running and SLS in comparison to the contralateral limb or a control group. Furthermore, the alteration noted in sagittal-plane gait biomechanics seems to be related in part to knee-specific impairments of quadriceps weakness, as well as limited knee flexion in range of motion or gait pattern. Therefore, future research will be necessary to discover if addressing knee-specific impairments can result in gait biomechanics improving in individuals following ACLR, and whether this will protect them from PFJ OA and TFJ OA.

7.3 Clinical Implications

No previous research has addressed kinematics and kinetics before and after ACL reconstruction during running and SLS. Therefore, the results of this thesis could assist in guiding the development of new or alternative treatment options to improve patients' long-term joint health after an ACL injury. In addition, it could be used as a basis for further research to explore the findings further.

This thesis has highlighted the importance of the patellofemoral joint in influencing recovery across individuals undergoing ACLR. Patellofemoral pain syndrome (PFPS) and its sequel, patellofemoral osteoarthritis (PFOA), is frequently witnessed across individuals who have had ACLR, significantly reducing their quality of life (Culvenor et al., 2016). The thesis shows that individuals presenting with ACLR exhibit abnormal movement patterns (such as decreased peak knee flexion angles and peak internal knee extensor moments) compared to controls. Such changes are evident in Chapter Five, concerning six to eight months after ACLR. These individuals also exhibit increased external knee-adduction moments, and this causes alterations in gait mechanics, which in turn causes more stress around the knee joint in areas that do not normally absorb such forces, leading to TFJ OA. Hip internal rotation angles demonstrated statistically significant variations between limbs in the ACLR group during running, and there was a significant increase in hip frontal (adduction angle) during SLS. This has implications for treatment programmes, as pain from patellofemoral OA has been linked to an abnormal increase in the hip internal rotation and adduction angles (Motlagh et al., 2013, Noehren et al., 2012). Therefore, assessment of these characteristics should be carried out post ACLR and suitable programmes put in place to manage patellofemoral dysfunction.

The thesis illustrates how individuals presenting with ACLR patients exhibit abnormal movement patterns (such as decreased peak knee flexion angles and peak internal knee extensor moments) compared to controls. Such changes are evident in Chapter Six, concerning before and three and six months post-ACLR intervention. These findings are supported by Kuenze et al. (2014) who explain that individuals undergoing ACLR show a marked reduction in peak knee flexion angles and knee extensor moments during running (Kuenze et al., 2014). These behaviours are typical of a quads avoidance strategy. In such circumstances, individuals avoid the use of the quadriceps during gait or physical activity. ACLR patients exhibit significant reductions in quadriceps' strength, force development and activation levels (degree of inhibition) (Lepley and Palmieri-Smith 2015; Laudner et al., 2015; Kline et al., 2015; Harput et al., 2015). These changes increase the risk of osteoarthritis and PFPS across concerned participants (Culvenor et al., 2016).

Loading was also shown to be an issue, as the current study revealed significant differences in the peak knee flexion angle, peak internal knee extensor moment and extensor impulses between the reconstructed and uninjured limbs during running six to eight months after ACLR (see Chapter 5). Loading may be reduced if knee flexion angle is reduced as a strategy to lower the risk of damage, reduce pain and maintain stability. During running, a reduction in the knee flexion angle was noted in the ACLR group compared to the control group. Such a shift in knee flexion angle is likely to alter the location of tibiofemoral joint contact and the patellofemoral contact stress (Hall et al. 2012; Boling et al. 2009), creating greater contact stress in a smaller contact area, and increasing the risk of cartilage damage (Boiling, 2009). Furthermore, limiting the contact area over patellofemoral loads would lead to increased contact stress on small areas of patellofemoral contact due to using smaller knee flexion angles during high demand tasks, and this may increase the risk of early patellofemoral disease progression, as well as tibiofemoral (TF) OA. Although ACLR patients may reduce knee flexion in order to reduce patellofemoral compressive forces and consequently reduce pain, as shown by Dierks et al. (2010), this is a short term solution, and it needs to be addressed in the long term through training and education.

One of the main adaptations found in this thesis is a reduction in internal knee extensor moments in the injured leg, which probably acts as a stress prevention strategy to reduce load following ACLR, protect the ligaments from further damage, as well as reduce pain. This was also found and pointed out in the study by DeVita et al. (1998). To implement an effective stress prevention strategy, the reorganisation of motor tasks, and the re-equilibration of sensory inputs and motor outputs following surgery would be required, and this could form part of gait retraining. In addition, similar to Butler et al. (2009), peak external knee adduction moment was significantly increased in the ACLR group compared to the control group, which along with greater loading of the medial compartment could lead to OA. The differences observed in levels of performance, knee angles and moments in ACLR patients compared to controls suggest that the risk of degenerative change is greater in this group and could lead to early onset PFJ and TFJ OA. Therefore, assessment should be carried out, and where necessary strategies should be employed to improve gait and rectify potentially harmful movements following surgery. This may be useful as a matter of course to reduce the risk of PFJ and TFJ OA, and the subsequent pain, as well as the need for patients to seek further interventions.

The results from this study also suggest that single leg squats could be used as part of the assessment process. Compared to the control group and participants' uninjured leg, peak internal knee extensor moment was reduced significantly for the injured leg of ACLR subjects and knee flexion angle was directionally, but not statistically, significantly reduced in ACLR subjects during SLS. In addition, hip flexion was shown to be reduced when knee flexion angles were increased (Yamazaki et al. 2009), requiring different strategies to be used at the ankle and hip to achieve normal squat depth. The current study has revealed that increased hip flexion angle and reduced knee flexion angle are inter-related at greater squat depths. In fact, knee flexion angle for an ACLR group should be studied further as there is a scarcity of literature on the subject. The need for this is highlighted by the current study, as the ACLR group showed alterations in kinetics compared to the control group during SLS, although kinematics remained at normal levels.

This section presents the clinical implications of PFPS, PFJ and TFJ OA associated with ACLR surgery. Furthermore, this chapter also elucidates the probable interventions that should help to minimise the prevalence of PFPS, PFJ and TFJ OA in at-risk individuals. In addition, a critical analysis of the clinical implications may help to design effective rehabilitation strategies in the near future.

7.3.1 Strengthening Exercises:

ACLR patients often experience quadriceps dysfunction in many cases, which leads to deficits in strength and activity. Although this thesis has not addressed quadriceps strength or activation in either, it has shown a reduction in internal knee extensor moment during peak loading between the injured leg of the ACLR group and both the non-injured leg and the control group. This sagittal plane kinetic pattern suggests the development of early post-injury compensation to make up for the decreased quadriceps function, increased knee joint effusion, and pain (Ingersoll et al. 2008; Lewek et al. 2002). Therefore, exercises to strengthen the quadriceps should be seen as an essential part of rehabilitation. Reduced knee extensor moments are often considered a result of strength and neural deficits in the quadriceps musculature, and net knee extensor moments were reduced in the ACL group; therefore, increased knee flexor activity should be encouraged to help prevent excessive anterior tibial translation and strain on the reconstructed ACL.

It is recommended for quadriceps strengthening exercises to be implemented prior to surgery in preparation (Culvenor et al. 2016), as well as afterwards. However, care needs to be taken with at risk patients, particularly as there are some issues with open chain exercises, for example, it is suggested that such exercises could lead to anterior tibial translation and may place stress on the ACL graft. However, as there is strong evidence that quadriceps weakness is related to PFPS and PFOA (Nobre, 2012), at-risk individuals should adhere to appropriate exercise training programmes (Nobre, 2012), including a preoperative training programme. Open chain knee extension exercises are likely to be beneficial, as well as traditional neuromuscular exercises, as these should improve the alignment and capability of the affected limb (Nobre, 2012).

7.3.2 Gait retraining:

The results of this thesis suggest that appropriate assessment and management of patellofemoral dysfunction would be beneficial during the rehabilitation phase to mitigate the impact from abnormal movement patterns and preserve knee function. Therefore, it confirms that rehabilitation strategies should address the neuromuscular deficits associated with abnormal knee-extension rotations to improve patellofemoral alignment (Lepley et al., 2015). One method could be for the patient to exercise using a forefoot strike (FFS) pattern, as that would increase the knee-extensor moments and reduce the vertical force during early stance (because of lower rates of loading). Running with a forefoot strike (FFS) pattern has been shown to reduce impact loading and can be considered an effective approach for reducing patellofemoral joint pain, which is linked to increases in the patellofemoral contact stresses (Farrokhi et al. 2011; Heino Brechter and Powers 2002). This is because running with a FFS pattern increases the vertical ground reaction force, causing increases in the knee-extensor.

Gait retraining has also been used to reduce high-impact loading among runners, for example the study by Crowell and Davis (2011). Runners were simply asked to run softly on a treadmill with a monitor in front of them displaying the tibial accelerations with each foot strike. All of the runners stated that the new gait pattern felt natural by the sixth training session, and the tibial shock was reduced on average by almost fifty percent, and vertical instantaneous and average load rates were reduced by 34% and 32% (Crowell and Davis 2011). These changes persisted at the one-month follow-up. The problem with loading has been highlighted in the current study, and clearly needs addressing. Crowell and Davis (2011) show that re-training is possible and effective, and a similarly simple method could be used by ACLR patients following surgery to increase the knee flexion angle

Real-time feedback training has also been shown to be an effective solution in the orthopaedic literature, with significant improvements reported. For example, White and Lifeso (2005) provided patients who had undergone a hip replacement, and had an asymmetrical gait pattern, with real-time force feedback from an instrumented treadmill, resulting in significant improvements. Dingwell et al. (1996) used an instrumented treadmill to address the gait patterns of unilateral, transtibial amputees, and revealed a positive impact on asymmetries. However, these studies examined gait patterns during walking, whereas the current study recorded gait patterns during running, which is more physically demanding. Therefore, it would be interesting to carry out further research into whether gait training has similar outcomes for running following ACLR; in particular, whether it is beneficial for increasing knee flexion angle and knee extensor moments. However, a major difficulty with this is that while real-time motion analysis systems can provide powerful feedback on gait, they are unlikely to be readily available in clinical settings.

Overall, this thesis study has added to the knowledge on the importance of gait and the differences between normal and ACLR limbs during running. The results suggest that various strategies can be used to address potentially damaging gait patterns, and the results from other relevant studies confirm this.

7.3.3 Orthotics:

This thesis has highlighted the importance of addressing the problem of joint degeneration in the early stages, and orthotics may provide an effective and simple means of doing so. Conservative treatments (e.g, lateral wedge insoles and valgus knee braces) have been shown to reduce the symptoms associated with knee OA, along with reducing the peak knee-adduction moment (Butler et al., 2007; Pollo et al., 2002). Furthermore, conservative interventions that alter mechanical alignment have been shown to be most effective in the early stages of joint degeneration. Therefore, patients not exhibiting radiographic joint degeneration and are asymptomatic, but are at risk of disease progression due to other risk factors (i.e knee injury), may benefit from such interventions (Shimada et al., 2006), and longitudinal studies are required to assess this.

Conservative mechanical interventions have been designed to reduce loading at the compartment affected by joint degeneration, which in the knee, is usually the medial compartment. Despite this area being addressed, individuals can still develop knee OA in the lateral compartment. The current research found that two of the participants in the ACL group are likely to be at greater risk of suffering knee OA in the lateral compartment, which highlights the need to address problems pertaining to the lateral compartment. Therefore, an initial gait analysis after ACLR may help to detect the compartment of the tibiofemoral joint that is at greater risk of knee OA. The results from this would be beneficial in guiding the design of conservative mechanical interventions. Moreover, it is essential to gain this information, as if the incorrect side is unloaded, the intervention could worsen the degenerative disease process and speed up early onset knee OA. Therefore, the long-term maintenance of joint integrity should be considered for ACL patients following surgical intervention, and not only the desire to return to previous activity levels. Long-term research studies should be carried out to examine the efficacy of conservative treatment options aimed at slowing down knee OA progression after an ACL rupture, and to discover whether they are effective for reducing the incidence of early knee OA onset.

7.3.4 Combination protocol:

Eccentric exercise based rehabilitation programmes lead to increased strength and activation of the quadriceps in affected individuals. Such rehabilitation programmes are preferred over neuromuscular electrical stimulation (NMES) for improving quadriceps function (Gerber et al., 2007; Lepley et al., 2015). According to Lepley et al. (2015), a combination of NMES and eccentric exercise could restore biomechanical limb symmetry across individuals undergoing ACL reconstruction (Lepley et al., 2015, Kim et al., 2010, Gerber et al., 2007). This is relevant to the current study considering the differences found between the uninjured and the ACLR limb during running and SLS. The study by Lepley et al. (2015) complements the findings of the current study because both studies have elucidated the importance of preserving quadriceps among concerned participants (Kuenze et al., 2014). Moreover, Lepley et al. (2015) indicate that the coupling of eccentric exercises and NMES leads to improved knee-flexion angle. On the other hand, it is well established that improved knee flexion angle and increased the knee extensor moment would reduce the stress on the knee joints during exercise (Boling et al., 2009, Gerber et al., 2007). Therefore, strengthening exercises should be coupled with NMES to reduce the risk of OA and patellofemoral pain in individuals with a history of ACLR (Lepley et al., 2015, Gerber et al., 2007).

7.3.5 Surgery:

The choice of grafts used for ACLR may have implications for quadriceps strength. Moreover, although reconstruction will restore the stability of the knee, it does not improve biomechanics. Therefore, it is important to discover exactly which types of reconstruction reduce or increase the likelihood of alterations to biomechanics. In this study, the participants were thirty-four ACLR elite athletes who had all undergone ACL reconstructive surgery; 20 had undergone hamstring grafts, and 14 had undergone bone-patellar tendon bone (BTB) grafts. These individuals were compared to a control group, as well as the ACLR limb compared to the uninjured limb. Furthermore, there is a range of literature (e.g. Pinczewski et al., 2007; Gobbi et al.,2003; Keays et al., 2007 and Kobayashi et al., 2004) on the different types of grafts, such as Webster et al (2005), who found that altered biomechanics during gait are more common in patients who have undergone BTB grafts. Combining the most effective type of surgery with the most appropriate intervention post-surgery is likely to achieve optimal results. Furthermore, while strategies involving exercise are essential, care must be taken not to stress the ACL graft, and simply focus on alignment and the work capacity of the impacted limb, as explained by Nobre (2012).

7.3.6 Conclusion:

The majority of studies in the literature have investigated single tasks, and no investigation to date has addressed kinematics and kinetics before and after ACL reconstruction during running and SLS.

Gait training to address issues around loading and knee flexion angle should be helpful in preventing problems such as damage to cartilage and OA in individuals post ACLR, based on the altered gait patterns noted in this study for the ACLR group compared to the control group and the uninjured limb. Furthermore, it may be noted that strengthening exercises should be coupled with neuromuscular stimulation and orthostatic devices to ensure an increase in knee-flexion angle; increase in peak knee extensor-moments, and a reduction in knee-adduction moments. Improved knee-flexion would lead to a greater absorption of impact forces at the knee joint. Such features would reduce the chance of effusion and inflammation at the joints. These adaptations would further reduce the risk of OA and patellofemoral pain.

Exercising using a forefoot strike (FFS) pattern would increase the knee-extensor moments and reduce the vertical force during early stance (because of lower rates of loading), while eccentric exercises (with NMES) would improve knee-flexion angles. On the other hand, lateral wedge insoles should be administered to the individuals concerned to decrease the knee-adduction moments. The lateral wedge insole aligns the foot into pronation to produce valgus moment at the ankle. Therefore, the centre of pressure of the ground reaction force in the foot is shifted laterally. The lateral shifting of the centre of the ground reaction force decreases the moment arm at the knee, and so the external knee-adduction moment and medial compartment loading are both decreased. Hence, a combination of NEMS, strengthening exercises and orthostatic implants might help in the successful rehabilitation of at-risk individuals.

7.4 Limitations and Future studies

This study's findings are based on variables that are acceptable and reliable, but with some technical limitations. For example, the results may not necessarily be replicated in other laboratories, as this is dependent on placing the markers accurately according to the ability of the individual; therefore, although the models and laboratory setting of this current study are consistent with other settings reported in the literature, the results should only be considered within the context of this setting. These limitations have restricted the ability to generalise the findings.

In addition, cluster markers could lead to errors in analysing 3D motion during high-speed movements, as transverse plane motion and frontal plane motion, as well as soft tissue artefact, are highly susceptible to errors (McGinley et al., 2009; Cappozzo et al., 1996).

Knee injuries, such as PFPS and ACL, present a high risk of misalignment, therefore susceptible individuals need to be identified before they actually sustain injuries, through assessment tasks to measure joint loading and joint angles. Athletes' training sessions and real sports environments need to evaluate standardised methods based on technological advances that are continuous, to inform and identify risks to athletes. Therefore, these findings need to achieve ecological validity, as the laboratory environment, which is tightly controlled, creates another limitation for this research. It is important to devise protocols that can effectively prevent injuries, or help towards this aim, by undertaking further research into squat tasks and running tasks that could reveal the cause of poor knee mechanics.

Marker positioning is another limitation of this current study of ACL reconstructed individuals, as test-retest reliability was not undertaken, and trainers used on AstroTurf and boots with studs used on grass mean it has failed to accurately represent real sporting interactions on various surfaces. This study has only tested using a Mondo running surface, with respondents wearing trainers defined as standard, which is also noted as a limitation. Respondents were asked to avoid losing their balance during the squat task and return to a standing position on a single leg, but there was insufficient control of individual respondents in terms of the depth of the squat. Therefore, further research is needed using a reliability study for populations with ACL injuries.

In Chapter Five, some limitations were also noted, including the heterogeneity of the subject population in regards to the level and kind of physical activity. As data from more subjects becomes available, it should be possible to investigate the influence of these confounding factors in a more robust manner. Joint degeneration could result from increased knee joint loading, therefore risk factors that could contribute towards this could be lower extremity biomechanics that are altered, and quadriceps weakness, following ACLR. However, the implications in the longer term concerning such differences remain unclear, so further research should focus on these factors. Individuals who have a history of ACLR should be observed to identify sources of biomechanical alterations, in order for conclusions to be drawn, as there is limited quadriceps strength data in this thesis, which is a concern of academics and clinicians. Therefore, for severe PFOA in knee OA patients, their gait deviations from observed loading responses could be helped by improving knee flexion range of motion, and quadriceps strength, through effective strategies.

Regarding this study, some other limitations can be noted, including the heterogeneity of the subject population in regards to the level and kind of physical activity; in addition, the sample size was rather small for detecting the reliability of such relationships. These factors could have influenced some of the results reported here. Changes over time were somewhat inconsistent across the subjects, and may have been dependent, to some extent, on subject-specific factors. This is an ongoing study, and data on all of the confounding variables listed above have been collected. As data from more subjects becomes available, it should be possible to investigate the influence of these confounding factors in a more robust manner. Another concern is that the level and kind of physical activities participants usually partake in were not taken into account. Future studies need to involve a prospective study for the same kind of sport, activity level, and with a sufficient sample size for specific populations. The patellofemoral cartilage's long-term life could be increased if biomechanics that are abnormal, and their connection with neuromuscular deficits, are used to inform rehabilitation strategies for patients, as PFOA development exposes them to the risk factors of reduced knee extensor moment or knee flexion angle. This could be determined by larger prospective studies.

Future studies that provide a better understanding of the source of altered biomechanical deviations at the knee may help clinicians to evaluate readiness for return to activity more effectively, as well as targeting treatment during the key stages of rehabilitation. Alterations in kinetics and kinematics experienced by ACLR individuals are different when compared to healthy individuals and between limbs, as they show lower knee extensor moments and lower knee flexion angles, although there are usually no apparent problems during a state of rest. Reduced long-term joint health and higher risks of knee joint injuries result from the impact of kinetic and kinematic alterations of the hip and knee joints when running persistently, but there is insufficient clarity on the source of these alterations.

Future studies based on the results of this thesis, including rehabilitation and prevention programs for early PFOA post-ACLR, may focus on improving the kinematic and kinetic deficits observed during running. Such a focus may include dynamic neuromuscular control and strengthening, functional retraining, and potentially prophylactic knee braces or insoles to control abnormal knee adduction moment. Prospective evaluations are required to establish whether optimising these biomechanical parameters prior to the development of advanced disease reduces the risk of progression to OA disease.

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APPENDICES

Appendix 1

NRES Committee London - Queen Square
HRA Head Office
Skipton House
80 London Road
London
SE1 6LH

Telephone: 020 797 22580

05 February 2014

Dr Richard Jones
Senior Lecturer in Clinical Biomechanics
University of Salford
PO18 Brian Blatchford Building
Frederick Road Campus
University of Salford, United Kingdom
M6 6PU

Dear Dr Jones

Study title:	**Running gait analysis and functional assessment before and after anterior cruciate ligament (ACL) reconstruction. RACL study**
REC reference:	**14/LO/0255**
IRAS project ID:	**139683**

The Proportionate Review Sub-committee of the NRES Committee London - Queen Square reviewed the above application on 04 February 2014.

We plan to publish your research summary wording for the above study on the NRES website, together with your contact details, unless you expressly withhold permission to do so. Publication will be no earlier than three months from the date of this favourable opinion letter. Should you wish to provide a substitute contact point, require further information, or wish to withhold permission to publish, please contact the Co-ordinator Hayley Fraser NRESCommittee.London-QueenSqaure@nhs.net

Ethical opinion

After review of the PRS application no ethical issues were raised.

On behalf of the Committee, the sub-committee gave a favourable ethical opinion of the above research on the basis described in the application form, protocol and supporting documentation, subject to the conditions specified below.

Ethical review of research sites

The favourable opinion applies to all NHS sites taking part in the study, subject to management permission being obtained from the NHS/HSC R&D office prior to the start of the study (see "Conditions of the favourable opinion" below).

Conditions of the favourable opinion

The favourable opinion is subject to the following conditions being met prior to the start of the study.

Stockport NHS
NHS Foundation Trust

Research & Development Office
F08, Pinewood House
Stepping Hill Hospital
Poplar Grove
Stockport
SK2 7JE

Tel: 0161 419 5801 / 5814
E-mail: jan.smith@stockport.nhs.uk

Dr Richard Jones
Senior Lecturer in Clinical Biomechanics
University of Salford
PO18 Brian Blatchford Building
Frederick Road Campus
Salford
M6 6PU

25 February 2014

Dear Richard,

Research Office Reference Number:
Project Title:
REC number:
NIHR CSP No:

Thank you for your application for Research Office approval for the above study.

I am pleased to confirm that we have now received and reviewed all necessary documentation, and Stockport NHS Foundation Trust has no objection to being a Participant Identification Centre (PIC) for this study.

Please note, as a PIC, activity at Stockport will be limited to providing potential participants an invitation letter and data access form. If interested in participating, patients will contact the study team at University of Salford direct.

I would like to take this opportunity to wish you well with your research.

Yours sincerely,

Jan Smith
Research & Development Manager

cc: Saud Alarifi
David Johnson
Lindsey Barber

University of Salford MANCHESTER

Research, Innovation and Academic Engagement Ethical Approval Panel

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

T +44(0)161 295 7016
r.shuttleworth@salford.ac.uk

www.salford.ac.uk/

16 December 2013

Dear Saud,

RE: ETHICS APPLICATION HSCR13/74 – Running gait analysis and functional assessment before and after anterior cruciate ligament (ACL) reconstruction

Based on the information you provided, I am pleased to inform you that application HSCR13/74 has now been approved.

If there are any changes to the project and/ or its methodology, please inform the Panel as soon as possible.

Yours sincerely,

Rachel Shuttleworth

Rachel Shuttleworth
College Support Officer (R&I)

Appendix 2

University of Salford MANCHESTER

KOOS KNEE SURVEY

Answer each question by checking the appropriate box, only <u>one</u> box for each question.

Symptoms
These questions should be answered thinking of your knee symptoms during the **<u>last week</u>**.

S1. Do you have swelling in your knee?

Never	Rarely	Sometimes	Often	Always
☐	☐	☐	☐	☐

S2. Do you feel grinding, hear clicking or any other type of noise when your knee moves?

Never	Rarely	Sometimes	Often	Always
☐	☐	☐	☐	☐

S3. Does your knee catch or hang up when moving?

Never	Rarely	Sometimes	Often	Always
☐	☐	☐	☐	☐

S4. Can you straighten your knee fully?

Always	Often	Sometimes	Rarely	Never
☐	☐	☐	☐	☐

S5. Can you bend your knee fully?

Always	Often	Sometimes	Rarely	Never
☐	☐	☐	☐	☐

Stiffness
The following questions concern the amount of joint stiffness you have experienced during the **<u>last week</u>** in your knee. Stiffness is a sensation of restriction or slowness in the ease with which you move your joints.

S6. How severe is your joint stiffness after first wakening in the morning?

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

S7. How severe is your stiffness after sitting, lying or resting **later in the day**?

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

Pain

P1. How often do you experience knee pain?

Never	Monthly	Weekly	Daily	Always
☐	☐	☐	☐	☐

What amount of knee pain have you experienced the **last week** during the following activities?

P2. Twisting/pivoting on your knee

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P3. Straightening knee fully

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P4. Bending knee fully

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P5. Walking on flat surface

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P6. Going up or down stairs

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P7. At night while in bed

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P8. Sitting or lying

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P9. Standing upright

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

Function, daily living
The following questions concern your physical function. By this we mean your ability to move around and to look after yourself. For each of the following activities please indicates the degree of difficulty you have experienced in the **last week** due to your knee.

A1. Descending stairs

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A2. Ascending stairs

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

For each of the following activities please indicate the degree of difficulty you have experienced in the last week due to your knee.

A3. Rising from sitting

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A4. Standing

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A5. Bending to floor/pick up and object

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A6. Walking on flat surface

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A7. Getting in/out of car

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A8. Going shopping

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A9. Putting on socks/stockings

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A10. Rising from bed

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A11. Taking off socks/stockings

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A12. Lying in bed (turning over, maintaining knee position)

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A13. Getting in/out of bath

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A14. Sitting

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A15. Getting on/off toilet

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

For each of the following activities please indicate the degree of difficulty you have experienced in the **last week** due to your knee.

A16. Heavy domestic duties (shoveling snow, scrubbing floors etc)

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A17. Light domestic duties (cooking, dusting etc)

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

Function, sports and recreational activities

The following questions concern your physical function when being active on a higher level. The questions should be answered thinking of what degree of difficulty you have experienced during the **last week** due to your knee.

SP1. Squatting

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

SP2. Running

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

SP3. Jumping

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

SP4. Twisting/pivoting on your injured knee

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

SP5. Kneeling

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

Quality of Life

Q1. How often are you aware of your knee problem?

Never	Monthly	Weekly	Daily	Constantly
☐	☐	☐	☐	☐

Q2. Have you modified your lifestyle to avoid potentially damaging activities to your knee?

Not at all	Mildly	Moderately	Severely	Totally
☐	☐	☐	☐	☐

Q3. How much are you troubled with lack of confidence in your knee?

Not at all	Mildly	Moderately	Severely	Totally
☐	☐	☐	☐	☐

Q4. In general, how much difficulty do you have with your knee?

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

Appendix 3

University of Salford MANCHESTER

IKDC KNEE EVALUATION FORM

Your Full Name ____

Your Date of Birth _________/__________/_________
Day Month Year

Your Gender: ❑ **Male** ❑ **Female**

Occupation ____

Today's Date _________/__________/_________
Day Month Year

SYMPTOMS:
Grade symptoms at the highest activity level at which you think you could function without significant symptoms, even if you are not actually performing activities at this level.

1. What is the highest level of activity that you can perform without significant knee pain?

❑Very strenuous activities like jumping or pivoting as in basketball or soccer
❑Strenuous activities like heavy physical work, skiing or tennis
❑Moderate activities like moderate physical work, running or jogging
❑Light activities like walking, housework or yard work
❑Unable to perform any of the above activities due to knee pain

2. During the past 4 weeks, or since your injury, how often have you had pain?

	0	1	2	3	4	5	6	7	8	9	10	
Never	❑	❑	❑	❑	❑	❑	❑	❑	❑	❑	❑	Constant

3. If you have pain, how severe is it?

	0	1	2	3	4	5	6	7	8	9	10	
No pain	❑	❑	❑	❑	❑	❑	❑	❑	❑	❑	❑	Worst pain

4. During the past 4 weeks, or since your injury, how stiff or swollen was your knee?

❑Not at all
❑Mildly
❑Moderately
❑Very
❑Extremely

5. What is the highest level of activity you can perform without significant swelling in your knee?

❑Very strenuous activities like jumping or pivoting as in basketball or soccer
❑Strenuous activities like heavy physical work, skiing or tennis
❑Moderate activities like moderate physical work, running or jogging
❑Light activities like walking, housework, or yard work
❑Unable to perform any of the above activities due to knee swelling

6. During the past 4 weeks, or since your injury, did your knee lock or catch?

❑Yes ❑No

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7. What is the highest level of activity you can perform without significant giving way in your knee?

❑Very strenuous activities like jumping or pivoting as in basketball or soccer
❑Strenuous activities like heavy physical work, skiing or tennis
❑Moderate activities like moderate physical work, running or jogging
❑Light activities like walking, housework or yard work
❑Unable to perform any of the above activities due to giving way of the knee

SPORTS ACTIVITIES:

8. What is the highest level of activity you can participate in on a regular basis?

❑Very strenuous activities like jumping or pivoting as in basketball or soccer
❑Strenuous activities like heavy physical work, skiing or tennis
❑Moderate activities like moderate physical work, running or jogging
❑Light activities like walking, housework or yard work
❑Unable to perform any of the above activities due to knee

9. How does your knee affect your ability to:

		Not difficult at all	Minimally difficult	Moderately Difficult	Extremely difficult	Unable to do
a.	Go up stairs	❑	❑	❑	❑	❑
b.	Go down stairs	❑	❑	❑	❑	❑
c.	Kneel on the front of your knee	❑	❑	❑	❑	❑
d.	Squat	❑	❑	❑	❑	❑
e.	Sit with your knee bent	❑	❑	❑	❑	❑
f.	Rise from a chair	❑	❑	❑	❑	❑
g.	Run straight ahead	❑	❑	❑	❑	❑
h.	Jump and land on your involved leg	❑	❑	❑	❑	❑
i.	Stop and start quickly	❑	❑	❑	❑	❑

FUNCTION:

10. How would you rate the function of your knee on a scale of 0 to 10 with 10 being normal, excellent function and 0 being the inability to perform any of your usual daily activities which may include sports?

FUNCTION PRIOR TO YOUR KNEE INJURY:

Cannot perform daily activities | No limitation daily activities

0	1	2	3	4	5	6	7	8	9	10
❑	❑	❑	❑	❑	❑	❑	❑	❑	❑	❑

CURRENT FUNCTION OF YOUR KNEE:

Cannot perform daily activities | No limitation daily activities

0	1	2	3	4	5	6	7	8	9	10
❑	❑	❑	❑	❑	❑	❑	❑	❑	❑	❑

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Appendix 4

University of Salford MANCHESTER

Participant Information Sheet

Part 1

Study title: Running gait analysis and functional assessment before and after anterior cruciate ligament (ACL) reconstruction: **RACL** study

You are being invited to take part in a research study. So, please read this information sheet carefully to understand why this study is being conducted and what your part in the study is. Please feel free to ask at any time if you require any further information.

- **Part 1** tells you the purpose of this study and what will happen to you if you take part.
- **Part 2** gives you more detailed information about the conduct of the study.

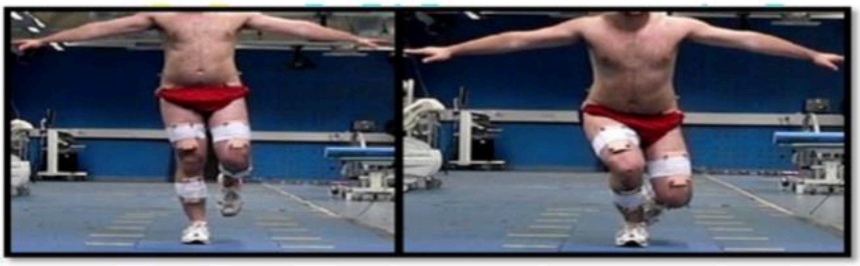
Please ask if there is anything that is not clear or if you would like more information. Take time to decide whether or not you wish to take part.

Part 1

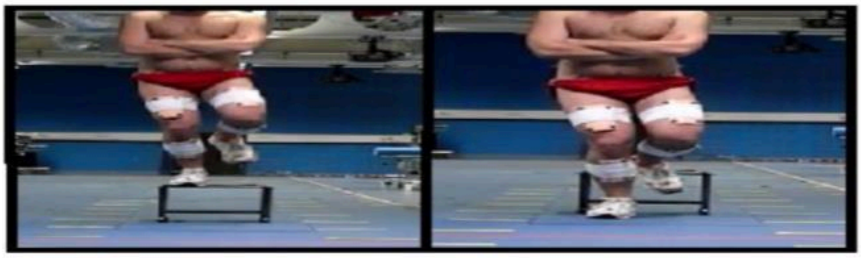
What is the purpose of the study?

Anterior Cruciate Ligament (ACL) injuries are common sports injuries but it is unknown whether there are any differences in running mechanics after an ACL injury. The purpose of the study is to investigate the differences in the movement of your limbs before and after ACL reconstruction during running, single leg squat and single leg landing (see pictures 1,2 and 3), which may be linked to predicting knee re-injury risk in the future. This means that the tests we are conducting will be recorded to see how the results correlate between before and after ACL reconstruction.

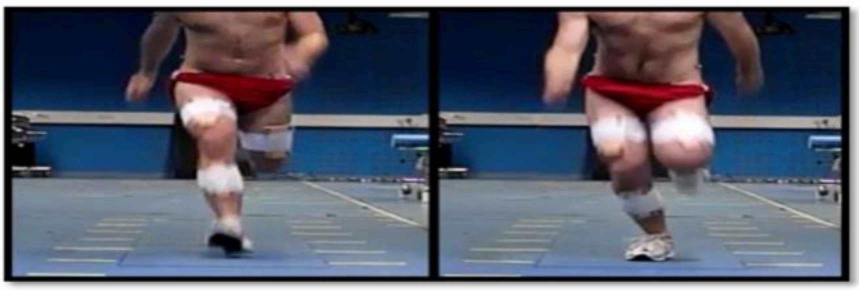
Picture 1. Single Leg Squat Task



Picture 2: Single Leg Landing Task



Picture 3: Running Task



Why have I been asked to participate?

You have been asked to participate as the referring surgeon has identified you that you have ACL injury and you had or plan to have reconstruction on the damaged ligament.

Do I have to take part?

You do not have to take part if you do not want to. You are free to withdraw from the study at any time. You do not have to give any reason and this will not affect the care you receive, either now or in the future. If you do not wish to take part in study we would be grateful if you could send back the tear-off slip confirming you would not like to take part so that we can update the records.

What will happen to me if I take part?

You will be asked to attend the Human Performance Laboratory at the University of Salford for an assessment before and after your surgery. In this session, you will complete some questionnaires in regards to your knee and also have some functional tests undertaken on your knee. After this, you will be asked to wear a loose pair of shorts to expose your lower limb and trunk in order to allow the video cameras to record you during the study.

The researcher will then attach a set of small reflective markers directly to the skin on your lower limbs and back and you will be provided with running shoes where markers are already attached. You will then be required to undertake five successful trials for each test, which include running, single leg squat and single leg landing. The testing will not involve any exertion that you are not accustomed with through your current activity levels, taking no longer than 1 hour for each session and requires you to attend three sessions: before surgery, after 3 and 6 months after surgery.

Your identity will not be revealed; the video recording will only be of your lower limbs and trunk. Moreover, your identity will not be disclosed at any time and the information obtained from this clip will be treated as privileged and will remain confidential.

<u>What are potential benefits from taking part in this study?</u>

There is no direct benefit to you but your participation will contribute to healthcare research and the information we obtain may help in investigating running and functional test analysis before and after ACL surgery.

<u>What are the possible risks?</u>

There is an inherent risk with any type of testing, however the testing for this study will be in a controlled laboratory environment and therefore any risks are minimal. Additionally, the movements are standard movements that you will be aware of and would have undertaken previously.

<u>Is the researcher being paid for including me in the study?</u>

Support towards your travel expenses will be reimbursed for your participation in this research.

THIS COMPLETES PART 1 OF THE INFORMATION SHEET.

Part2

What will happen if I do not want to carry on with study?

You can stop the testing at any time without giving a reason or it affecting your health care in any way. If you do decide to withdraw, any data we have collected will be retained and used as part of the study, unless you specifically wish it to be deleted.

What if there is a problem?

If you have a concern about any aspect of this study, you should ask to speak with the researcher (Saud Alarifi S.M.Alarifi@edu.salford.ac.uk) who will do their best to answer your questions. If you remain unhappy and wish to complain formally, you can do this by contacting the Chief Researchers (Dr. Lee Herrington L.C.Herrington@salford.ac.uk and Dr. Richard Jones R.K.Jones @salford.ac.uk) or through the NHS Complaints Procedure. In the event that something does go wrong and you are harmed during the research study there are no special compensation arrangements. If you are harmed and this is due to someone's negligence then you may have grounds for a legal action for compensation against Salford University but you may have to pay your legal costs.

Will my taking part in this study be kept confidential?

Any information obtained in connection with this study will be treated as privileged and confidential. All information will be anonymised so that you cannot be identified, except by a single paper form which will be stored securely in a lockable filing cabinet at Salford University. The research team, their colleagues, the sponsors and people who need to audit the conduct of our research will have access to the identifiable forms. The data will be analysed to complete the study as outlined above. We will also keep the data for at least five years and may use it in future studies to improve our understanding of movement problems. For example, we may wish to combine the data from this study with that of future studies to enable us to use more powerful analysis techniques. Ethical approval will not normally be sought for these studies.

What happen to the results of the study?

The final results of the study will be available to you as a short synopsis after the study has finished. The plan is for the results of the study to be published and as part of the postgraduate PhD qualification of the lead researcher.

Who is organising and funding the research?

Saud Alarifi, a PhD student in the School of Health Sciences at University of Salford, under the supervision of Dr. Lee Herrington and Dr. Richard Jones, conducts the research.

Who has reviewed the study?

The College of Health and Social Care Research Governance and Ethical Committee have reviewed the study. Additionally, this study has received a favourable opinion from the NHS ethical committee (NHS rec number 14/LO/0255).

Please feel free to ask any further questions about the nature or demands of the project at any time.

Many thanks for your participation.

Saud Alarifi

S.M.Alarifi@edu.salford.ac.uk

Appendix 5

University of Salford MANCHESTER

<u>Consent Form</u>

1. *Saud Alarifi,* who is a Postgraduate research student at the University of Salford, has requested my participation in a research study. My involvement in the study and its purpose has been fully explained to me. ☐

2. My participation in this research will involve a number of tests, which include Running, Single Leg Squat and Single Leg Landing. ☐

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

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

5. I understand that the results of this research may be published but that my name or identity will not be revealed at any time. In order to keep my records confidential, *Saud Alarifi* will store all information as numbered codes in computer files that will only be available to him. ☐

6. I have been informed that any questions I have at any time concerning the research or my participation in it will be answered by Saud Alarifi and I can contact him at (S.M.Alarifi@edu.salford.ac.uk). ☐

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

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

Researcher: ………………… **Signed:** ……………… **Date:** ……………

College Ethics Panel Approval Form, PGR Version 2013-14

Appendix 6

University of Salford MANCHESTER

Saud Alarifi
Tel:07427614059
Email: S.M.Alarifi@edu.salford.ac.uk

Insert Address

21 January 2014

Study Title: Running gait analysis and functional assessment before and after anterior cruciate ligament (ACL) reconstruction: **RACL** study

Principal Investigator: Saud Alarifi
Chief investigator: Dr. Lee Herrington and Dr. Richard Jones.

Dear *(enter patients name)*

I would like to invite you to participate in the above **RACL** research study taking part at Salford University. Please find enclosed a Patient Information Sheet outlining details of the trial. Please take your time to read the information and if you have any questions please do not hesitate to contact myself or a member of the research team

Please remember you are under no obligation to take part, however if you would like to be contacted by a member of the research team I would be very grateful if you could sign the Data Protection Consent Form and return it in the pre-paid envelope with the tear off slip below.

If you do not wish to participate we would be grateful l if you could still return the form for our records. We would like to thank you for your help and interest with this research.

Yours sincerely,

Saud Alarifi - Principal Investigator
PhD Student at university of Salford
Chief investigators: Dr. Lee Herrington and Dr. Richard Jones.

Enc: Patient Information Sheet
Data Protection Consent Form
Pre-Paid Envelope

✂--

Study Title: Running gait analysis and functional assessment before and after anterior cruciate ligament (ACL) reconstruction: RACL study

NAME: ____

Please tick the appropriate box below:

- ☐ I am interested in taking part in the above study and would like to be contacted
- ☐ I am not interested in taking part in the above study.

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Appendix 7

Supplementary material B: Modified Downs and Black Scale

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

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

Appendix 8

Chapter Five Running Results:

Kinematics and kinetics between injured and non-injured limbs differences after ACLR.

Variable	Group		P -value
	Injured limb (n=34)	Non-Injured limb (n=34)	
Hip Internal Rotation Angle (°)	7.14±6.94	3.43±7.04	*0.01
Hip Internal Rotation Moment (Nm/Kg)	-0.73±21	-0.77±0.27	0.08
Hip Adduction Angle (°)	11.15±5.59	10.16±4.44	0.30
Hip Adduction Moment (Nm/Kg)	-2.00±0.58	-2.07±0.55	0.30
Knee Adduction Angle (°)	5.82±5.01	4.85±4.48	0.28
Knee Adduction Moment (Nm/Kg)	0.88±0.37	0.79±0.36	0.37
Knee Flexion Angle (°)	44.94±6.18	48.95±6.40	*0.01
Knee Extensor Moment (Nm/Kg)	2.80±0.63	3.32±0.70	*0.01
Knee Extensor Impulse (Nm/Kg*s)	0.24±0.07	0.32±0.7	*0.01
VGRF (BW)	2.57±0.22	2.62±0.25	0.15

Kinematics and kinetics between Injured limb and control group differences.

Variable	Groups		P -value
	Injured limb (n=34)	Control (n=34)	
Hip Internal Rotation Angle (°)	7.14±6.94	4.74±6.24	0.08
Hip Internal Rotation Moment (Nm/Kg)	-0.73±21	-0.72±0.22	0.55
Hip Adduction Angle (°)	11.15±5.59	11.96±4.73	0.66
Hip Adduction Moment (Nm/Kg)	-2.00±0.58	-1.89±0.54	0.58
Knee Adduction Angle (°)	5.82±5.01	6.11±4.20	0.80
Knee Adduction Moment (Nm/Kg)	0.88±0.37	0.69±0.36	*0.04
Knee Flexion Angle (°)	44.94±6.18	50.61±8.83	*0.01
Knee Extensor Moment (Nm/Kg)	2.80±0.63	3.59±0.61	*0.01
Knee Extensor Impulse (Nm/Kg*s)	0.24±0.07	0.36±0.12	*0.01
VGRF (BW)	2.57±0.22	2.67±0.41	0.08

Kinematics and kinetics with between non-Injured and control group differences.

Variable	Groups		P -value
	Non-Injured limb (n=34)	Control (n=34)	
Hip Internal Rotation Angle (°)	3.43±7.04	4.74±6.24	0.54
Hip Internal Rotation Moment (Nm/Kg)	-0.77±0.27	-0.72±0.22	0.29
Hip Adduction Angle (°)	10.16±4.44	11.96±4.73	0.11
Hip Adduction Moment (Nm/Kg)	-2.07±0.55	-1.89±0.54	0.30
Knee Adduction Angle (°)	4.85±4.48	6.11±4.20	0.23
Knee Adduction Moment (Nm/Kg)	0.79±0.36	0.69±0.36	0.24
Knee Flexion Angle (°)	48.95±6.40	50.61±8.83	0.62
Knee Extensor Moment (Nm/Kg)	3.32±0.70	3.59±0.61	0.17
Knee Extensor Impulse (Nm/Kg*s)	0.32±0.07	0.36±0.12	0.14
VGRF (BW)	2.62±0.25	2.67±0.41	0.31