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UNIVERSITY *of* LIMERICK

OLLSCOIL LUIMNIGH

# Lower limb kinematics, kinetics and coordination during a land and cut task; the role of gender and previous ACL injury

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*Sarah Breen BSc.*

A thesis submitted for the degree of Doctor of Philosophy



Biomechanics Research Unit,

Department of Physical Education and Sport Sciences,

Faculty of Education and Health Sciences,

University of Limerick, Limerick, Ireland

Supervisors: Dr. Andrew J. Harrison and Dr. Ian C. Kenny

Submitted to the University of Limerick May 2012



## DEDICATION

To my father John Joe, for giving me what you strive to give all your students, the thirst for knowledge and motivation for self-improvement.

*“One mark of a great educator is the ability to lead students out to new places where even the educator has never been”*

# Lower limb kinematics, kinetics and coordination during a land and cut task; the role of gender and previous ACL injury

## ABSTRACT

Anterior cruciate ligament (ACL) injury continues to be a constant adversary to field sports athletes. Females are widely acknowledged as being at an increased risk of ACL injury, in comparison to males. Athletes who are successful in rehabilitation after surgery and return to their sport are reported to have an increased risk of repeated ACL injury and the development of osteoarthritis. The current thesis utilised a novel, maximal drop-jump land and unanticipated cutting task to assess the lower limb biomechanics of uninjured male and females, and previously ACL injured subjects (ACLr). Discrete measures of lower limb kinematics and kinetics were firstly compared between uninjured males and females, and secondly between the previously injured (PI) leg of ACLr subjects and both the contralateral non-injured (NI) leg and an uninjured subject's control leg. The results show that females had increased hip internal rotation, the PI leg was not significantly different to the NI leg but was different to the control subject's leg with increased hip flexion, internal knee abduction moment and transverse plane knee ROM. Lower limb coordination was assessed in the ACLr subjects and both legs of the ACLr subjects had similar coordination patterns. The PI leg however showed different coordination patterns than the control subject's leg for a number of couplings. Movement and coordination variability were also utilised for a gender and ACLr – control comparison. The female subjects and the PI leg had lower levels of movement and coordination variability than males and the contralateral non-injured leg respectively. The PI leg however, had higher levels of movement and coordination variability than the control subject's leg. In conclusion, females and previously ACL injured subjects may be at an increased risk of initial ACL injury and the development of osteoarthritis on the PI leg respectively, due to lower levels of movement and coordination variability. Altered biomechanics at the hip were also highlighted as a potential mechanism increasing injury risk in females and ACLr subjects.

## **AUTHOR'S DECLARATION**

I hereby declare that the work contained in this thesis is my own and was completed with the counsel of my supervisors, Dr AJ Harrison and Dr IC Kenny of the Department of Physical Education and Sport Sciences, University of Limerick. This work has not been submitted to any other higher education establishment, or for any other academic award in this Institute.

Signed:

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Sarah Breen

22/03/2012

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Dr A.J. Harrison

22/03/2012

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Dr I.C. Kenny

22/03/2012

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- Jen, Kate, Dave & Paul, *"My home is not a place, it is people."*
- Ellie, Dribbles and Twix *"Animals are such agreeable friends — they ask no questions, they pass no criticisms."*

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## **LIST OF ABBREVIATIONS**

ACL – Anterior Cruciate Ligament

AFL - Australian Football League

Q-angle - Quadriceps femoris angle

OA – Osteoarthritis

ACLR – Individuals who have received ACL reconstruction surgery and finished rehabilitation

nACL – Individuals with no history of ACL injury

PI – Previously injured leg of ACLr subject

NI - Contralateral non-injured leg of ACLr subject

C – Control leg on nACL subject matched for dominance to PI leg of ACLr subject

ROM – Range of Motion

TD - Touchdown

Max - Maximum

Min – Minimum

3D – Three-dimensional

OA – Osteoarthritis

S.D. – Standard Deviation

AVG – Average

$\eta^2$  - Partial eta squared

PFP – Patellofemoral pain

## **CHAPTER 1**

### **INTRODUCTION TO THE THESIS**

## 1.1 Introduction

Anterior cruciate ligament (ACL) injury continues to be a constant adversary to the athletes of field sports with upwards of 250,000 ACL injuries occurring in the United States each year (Boden et al., 2000). The female athlete has been widely acknowledged as being at an increased risk of ACL injury, in comparison with males participating in the same sport (Roos et al., 1995b; Bjordal et al., 1997; Myklebust et al., 1997; Myklebust et al., 1998; Arendt et al., 1999; Messina et al., 1999; De Loes et al., 2000; Olsen et al., 2004; Agel et al., 2005; Junge et al., 2006; Mihata et al., 2006; Prodromos et al., 2007; Borowski et al., 2008; Flood and Harrison, 2009; Randazzo et al., 2010; Pappas et al., 2011). Reconstructive surgery is typically recommended after ACL injury, to restore the knee joint stability and function required for sports participation. Up to 80% of athletes who undergo surgery are unable to successfully return to their pre injury-level of sport participation and therefore, quit their sports (Söderman et al., 2002; Chong and Tan, 2004; Ardern et al., 2010). Athletes who are successful in rehabilitation after surgery and return to their sport have been shown to be at an increased risk of repeated ACL injury to both the previously reconstructed knee (Oates et al., 1999; Orchard et al., 2001; Kvist, 2004; Myklebust and Bahr, 2005; Salmon et al., 2005; Tanaka et al., 2010) and the contralateral knee (Shelbourne et al., 1998; Oates et al., 1999; Orchard et al., 2001; Salmon et al., 2005). Additionally up to 50% of ACLr individuals will display signs of osteoarthritis (OA) 10 years post injury (Roos et al., 1995a; Von Porat et al., 2004; Pinczewski et al., 2007; Meuffels et al., 2009; Holm et al., 2010; Øiestad et al., 2010).

There is a plethora of previous investigations comparing lower limb biomechanics of male and female athletes and rehabilitated ACL injured subjects and controls. These investigations have utilised a variety of sport specific movement tasks such as landing (Decker et al., 2002; Salci et al., 2004) and side cutting or pivoting (Fedie et al., 2010; Tsarouhas et al., 2010). Landing movements rarely occur in isolation during match situations, therefore, some investigations have utilised tasks combining both landing and cutting movements (McLean et al., 2005; Ristanis et al., 2005). Although these studies are interesting advancements on the isolated tasks of landing and cutting, they still lack the unanticipated and high intensity nature of match situations. No study has investigated landing from a jump followed by cutting to either the right or left in an unanticipated manner, as would happen in a game after landing from catching a ball when the player would then try to evade a tackle.

Previous investigations have reported altered lower limb joint kinematics and kinetics in females (Chappell et al., 2002; Hewett et al., 2005; McLean et al., 2005; McLean et al., 2007) and previously ACL injured subjects (Ristanis et al., 2003; Tashman et al., 2004; Ristanis et al., 2005; Deneweth et al., 2010; Scanlan et al., 2010; Webster et al., 2011). These joint mechanics are undoubtedly important in terms of injury risk, however the coordination of these

movements and the distribution of their associated loadings over numerous repetitions of a movement task are also issues worthy of further investigation.

## **1.2 Aims of the thesis**

The aims of this thesis were to firstly determine potential biomechanical risk factors that contribute to the disproportionate incidence of non-contact ACL injury in females and secondly determine potential biomechanical mechanisms that contribute to the increased risk of ACL re-injury and the development of osteoarthritis in rehabilitated ACL injured athletes. The mechanisms to be investigated include lower limb joint kinematics, lower limb joint moments, lower limb intralimb segment and joint coordination and the variability of joint kinematics and lower limb segment and joint coordination. Hip, knee and ankle kinematics and moments and lower limb intralimb segment and joint coordination were assessed during the performance of a maximal drop-jump land and unanticipated cutting task. The variability of the hip and knee kinematics and lower limb intralimb segment and joint coordination were also assessed during the performance of a maximal drop-jump land and unanticipated cutting task.

## **1.3 Thesis Structure**

The thesis contains a series of progressively linked studies, which are presented in a manner consistent with the Scandinavian Approach. Two streams of investigation were addressed; the ACL injury risk associated with gender (Chapters 3-4), the risks of repeated ACL injury and degenerative joint disease associated with previous ACL injury (Chapters 5-7).

- CHAPTER 2 contains a review of the literature on Anterior Cruciate Ligament (ACL) injury. The structure and function of the ACL is outlined followed by a review of the literature on the mechanisms and risks of receiving an ACL injury. The fate of the ACL injured patient post rehabilitation is also discussed. The current position on the role of variability and injury risk is evaluated based on research findings to date. The relevant experimental issues such as population, movement task and biomechanical model selection are identified. The aims of this research programme in relation to the current knowledge are then outlined.
- CHAPTER 3 describes an investigation utilising a novel maximal drop-jump land and unanticipated cutting task to compare the lower limb mechanics and of uninjured male and female athletes. The use of this novel movement task will further inform the current literature on gender differences in lower limb biomechanics.
- CHAPTER 4 further examines the data collected in Chapter 3 to investigate any gender differences in movement and coordination variability during a maximal drop-jump land and unanticipated cutting task. The examination of movement and coordination variability in this novel movement task will add to previous research in the area (Pollard et al., 2005).

- CHAPTER 5 describes an investigation utilising a novel maximal drop-jump land and unanticipated cutting task to examine the lower limb mechanics and of the previously injured leg of ACL injured athletes. The lower limb mechanics of the previously injured leg were compared against the contralateral non-injured leg and a matched control leg from a non-injured athlete. The use of this novel movement task will further inform the current literature on adaptations in lower limb biomechanics present in previously injured leg of ACL injured subjects.
- CHAPTER 6 further examines the data collected in Chapter 5 to investigate any differences in the coordination patterns of the previously injured leg of ACL injured athletes. The coordination patterns of the previously injured leg were compared against the contralateral non-injured leg and a matched control leg from a non-injured athlete. The use of this composite measure of coordination will further assess any adaptations in lower limb biomechanics present in previously injured leg of ACL injured subjects.
- CHAPTER 7 further examines the data collected in Chapter 5 and 6 to investigate any differences in movement and coordination variability between the previously injured leg, the contralateral non-injured leg and a matched control leg from a non-injured athlete. The relationship between previous ACL injury and movement or coordination variability is a highly under investigated one. Its potential as a risk factor for re-injury and development of degenerative joint disease strongly merits investigation.

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## **CHAPTER 2**

### **REVIEW OF THE LITERATURE**

## **2.1 Introduction**

This chapter will introduce the issue of Anterior Cruciate Ligament Injury (ACL) and critique previous research in the area. The structural and functional anatomy of the ACL will be outlined and details of the ACL injury mechanism will be reported from a number of research perspectives. Various risk factors for ACL injury will be covered and the issue of gender will be discussed in further detail. The treatment protocols following ACL injury will also be discussed as will the associated risks to athletes who return to sport. The role of movement variability in injury and specifically ACL injury will conclude the review. The aims of this programme of research which have been driven by this review will then be delineated.

## **2.2 Anterior Cruciate Ligament Injury Problem**

Knee injuries occur regularly in various sports and account for 11-17% of all sports injuries (Junge et al., 2006; Junge and Dvorak, 2007; Shankar et al., 2007; Hägglund et al., 2009). One of the most devastating injuries to the knee is the rupture of the anterior cruciate ligament (ACL) which requires surgical reconstruction and a lengthy rehabilitation period. For example, the incidence of Anterior cruciate ligament (ACL) injuries in the general population in Switzerland is approximately one anterior cruciate or posterior cruciate ligament injury per 5000 people, at a rate of 0.62 and 0.52 per 100,000 hours of sports participation for males and females respectively (De Loes et al., 2000). The highest risk of ACL injury is in individuals between the ages of 15-25 years who are involved in cutting and pivoting sports (Myklebust et al., 1998).

The incidence of ACL injury has been recorded in soccer, basketball, team handball, skiing, rugby union, Australian football league (AFL) and lacrosse; these are shown in Table 2-1.

Table 2-1 ACL Incidence in sports

	Per 1000 athlete exposures		Per 1000 athlete hours	
<b>Soccer</b>	0.1-0.4	(Arendt et al., 1999; Hewett et al., 1999; Gwinn et al., 2000; Harmon and Ireland, 2000; Agel et al., 2005; Mandelbaum et al., 2005; Mihata et al., 2006)	0.02-2	(Bjordal et al., 1997; De Loes et al., 2000; Faude et al., 2005; Giza et al., 2005)
<b>Basketball</b>	0.06-0.5	(Arendt et al., 1999; Hewett et al., 1999; Gwinn et al., 2000; Harmon and Ireland, 2000; Meeuwisse et al., 2003; Agel et al., 2005; Lombardo et al., 2005)	0.01-0.09	(Gomez et al., 1996; Messina et al., 1999; De Loes et al., 2000; Lombardo et al., 2005)
<b>Handball</b>			0.005-0.77	(Myklebust et al., 1998; De Loes et al., 2000; Myklebust et al., 2003; Olsen et al., 2003)
<b>Skiing</b>			0.005-0.044	(Viola et al., 1999; De Loes et al., 2000)
<b>Rugby Union</b>			0.01-0.42	(Dallalana et al., 2007)
<b>AFL</b>			1	(Orchard and Seward, 2002) <i>(in season match time hours)</i>
<b>Lacrosse</b>	0.17	(Mihata et al., 2006)		

Gaelic football, hurling, (Gaelic Athletic Association Medical Player Welfare Committee, 2011) American football (Shankar et al., 2007) and rugby league (Hoskins et al., 2006) have also been noted as sports inundated with ACL injuries. With increasing levels of participation in these sports, ACL injuries are becoming increasingly notorious.

ACL injuries are a burden to the individual through pain, inconvenience and time off work. There is also a significant burden on society, through direct and indirect costs (Flood and Harrison, 2009). Knee injuries encompass a large portion of the medical care expenditure for sports injuries (Arendt et al., 1999). Hospital costs can exceed €6000 (USD\$ 7964) in the Republic of Ireland, USD\$9,220 in the United States (Kao et al., 1995), and NZD\$11,154 (USD\$ 9132) in New Zealand for reconstructive surgery, rehabilitation physiotherapy and the initial

diagnostic MRI or arthroscopic surgery<sup>1</sup>. The annual costs in the US for the treatment of ACL injury are estimated to range between 1.5 and 2 billion dollars (Griffin et al., 2000; Childs, 2002; Yu and Garrett, 2007). Subsequent to the injury, there may also be loss of time from work, school, or sports (Lee and Garraway, 1996; Oates et al., 1999; Hägglund et al., 2009). Up to 80% of athletes fail to return to pre-injury levels of sports participation, post surgery and rehabilitation (Söderman et al., 2002; Chong and Tan, 2004; Ardern et al., 2010).

The ACL is the most frequently disrupted knee ligament, but isolated ACL tears or ruptures are uncommon. Associated injuries to the menisci, other ligaments, joint cartilage, and subchondral or cancellous bone are often observed (Lohmander et al., 2007). In addition to the initial trauma ACL injuries are associated with long term joint degeneration which is further amplified due to the presence of these associated injuries (Øiestad et al., 2010). Up to 50% of previously injured knees experience premature osteoarthritis (OA) 10- 14 years post injury (Roos et al., 1995a; Von Porat et al., 2004; Pinczewski et al., 2007; Meuffels et al., 2009; Holm et al., 2010; Øiestad et al., 2010) This early onset OA is associated with pain, functional limitations, and decreased quality of life between the age of 30 and 50 years (Lohmander et al., 2004; Von Porat et al., 2004) this be further discussed in Section 2.7.2 Risk of Degenerative Joint Disease.

There is no doubt that sports participation and regular physical activity is of benefit to its participant's through improvements in mood and reductions in the risk of; coronary heart disease, hypertension, colon cancer, obesity, osteoporosis, falling in the elderly and diabetes mellitus (National Center for Chronic Disease Prevention and Health Promotion, 2011). Life expectancy is longer in high-level athletes (Sarna et al., 1993), with a lower rate of hospital care for heart disease, respiratory disease, and cancer (Kujala et al., 1996). On the other hand these high-level athletes are at an increased risk of developing gonarthrosis (Kujala et al., 1995). Although sports participation has undeniable benefits, the associated injuries are a significant consequence.

### **2.3 Function and Structure of the Anterior Cruciate Ligament (ACL)**

*"The ACL is a keystone to controlled, fluid, and stable flexion and rotation of the normal knee"*  
(Goldblatt and Richmond, 2003).

The ACL serves as the primary restraint to anterior tibial translation and as a secondary stabiliser to resist internal tibial rotation (Fukubayashi et al., 1982; Matsumoto et al., 2001) and varus and valgus motion (Fu et al., 1993).

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<sup>1</sup> Conversion into USD\$ is accurate as of 21<sup>ST</sup> March 2012.

The ACL joins the femur to the tibia and is completely covered by synovium and is thus extrasynovial but intra-articular (Mérida-Velasco et al., 1997). From its femoral attachment, the ACL runs anteriorly, medially, and distally to the tibia. Its length ranges from 22 to 41 mm (average, 32 mm) and its width from 7 to 12 mm (Amis and Dawkins, 1991). The ACL is composed of collagen fibrils grouped to form 1-20  $\mu\text{m}$  fibres. These ligamentous fibres merge to form larger units that fan out into three functional bands, antero-medial, postero-lateral and intermediate (Amis and Dawkins, 1991).

## 2.4 ACL injury mechanism

ACL injuries are usually classified into four grades; intact, low-grade partial tear, high-grade partial tear, and complete tear (Hong et al., 2003). The classic patient injury history for an ACL injury, is of a "pop" that occurs while landing, side cutting or pivoting (Boden et al., 2000a) followed by intense pain. The knee usually swells within 12 hours, positive results on the Lachman test<sup>2</sup> and pivot shift manoeuvre<sup>3</sup> are diagnostic (Cameron et al., 2000). MRI is used most frequently as definitive method of diagnosis. The majority of ACL injuries occur in sporting situations with a high incidence in sports that involve sudden stops and starts, cutting and pivoting e.g. team handball, Australian rules football, soccer, rugby union, basketball, gaelic football, hurling, American football, rugby league and netball (Myklebust et al., 1997; Myklebust et al., 1998; Arendt et al., 1999; Messina et al., 1999; Orchard and Seward, 2002; Hoskins et al., 2006; Dallalana et al., 2007; Shankar et al., 2007; Flood and Harrison, 2009; Gaelic Athletic Association Medical Player Welfare Committee, 2011). They also occur outside of the sports arena, as a result of events such as falling down stairs, motorcycle crashes, slipping on ice, or twisting the leg or knee (Boden et al., 2000b).

Seventy five per cent of ACL tears in the sporting context occur during game time or competitions (Myklebust et al., 1997; Myklebust et al., 1998). The sporting movements associated with occurrence of an ACL injury are deceleration movements such as landing (Gray et al., 1985; Hutchinson and Ireland, 1995; Myklebust et al., 2003; Cochrane et al., 2007; Hashemi et al., 2011), sudden one-step stops (Hutchinson and Ireland, 1995; Cochrane et al., 2007) or changes of direction such as sidestep (Hutchinson and Ireland, 1995; Myklebust et al., 1997; Myklebust et al., 2003; Cochrane et al., 2007).

### 2.4.1 Research Approaches for Identifying ACL Injury Mechanisms

Various research approaches have been utilised to describe the inciting event of an ACL injury; athlete interviews, clinical studies, cadaver studies, in vivo studies, mathematical

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<sup>2</sup> A test to detect deficiency of the ACL; with the knee flexed 20–30° and patient supine, the tibia is displaced anteriorly relative to the femur; a soft endpoint or greater than 4 mm of displacement is positive (abnormal). (Cameron et al., 2000)

<sup>3</sup> A test to detect a deficiency of the ACL; when the knee is moved into a position near full extension, a subluxation of the lateral tibial condyle upon the distal femur is positive. (Cameron et al., 2000)



modelling, video-based, injuries during biomechanical experiments and motion analysis of non-injury simulations (Krosshaug et al., 2005). Athlete interviews and video analysis studies can be used to get an idea of the gross biomechanics occurring prior to and during an ACL injury. Clinical, in-vivo, mathematical modelling and motion analysis studies give us a more detailed biomechanical inspection of the athlete's movements and the associated ACL loading prior to and during the ACL injury.

#### 2.4.2 *Classification of ACL Injuries*

ACL injury can occur with varying amounts of force employed by certain movements, other players or external objects. Hewett has classified ACL injuries into three categories (Hewett et al., 2007).

- Noncontact ACL injury occurs when the forces applied to the knee at the time of injury are generated from the athlete's own movements.
- Indirect contact ACL injury occurs when external force is applied to the athlete but not directly to the injured knee, in this case the indirect force is involved in the injury process but is not the proximate cause of the injury.
- Direct contact ACL injury occurs when an external force is applied to the injured knee and was probably the proximate cause of the injury.

Fifty-six to ninety-five per cent of all ACL injuries occur in a non-contact setting in a situation without external bodily contact, in a movement the athlete would have done numerous times before (McNair et al., 1990; Strand et al., 1990; Myklebust et al., 1997; Cochrane et al., 2007; Krosshaug et al., 2007; Kobayashi et al., 2010). Few studies report the percentage of indirect contact and direct contact injuries, these categories have been summarised into one 'contact' category in publications prior to the proposal of this new classification system (Hewett et al., 2007). Cochrane and colleagues did report on indirect and direct contact ACL injuries with Australian rules football reporting 56% non-contact, 32% direct contact and 12% indirect contact injuries (Cochrane et al., 2007). Similarly Olsen and colleagues reported 32% of ACL injuries in team handball occurred due to indirect contact and 68% during non-contact (Olsen et al., 2004).

#### 2.4.3 *Gross biomechanical description of ACL injury mechanism*

Video-based studies and athlete interview studies retrospectively analyse video footage and athlete experiences of ACL injury occurrences, identifying body postures and movement patterns at the time of injury. These studies indicate a common body posture during non-contact injury in which the knee is near full extension (0-30°), the tibia is internally or externally rotated, the foot is planted, and a deceleration followed by an valgus collapse of the knee joint occurs (Arnold et al., 1979; McNair et al., 1990; Boden et al., 2000a; Olsen et al., 2004; Cochrane et al., 2007; Krosshaug et al., 2007; Kobayashi et al., 2010).

#### 2.4.4 Detailed biomechanical description of ACL injury mechanism

Motion of the knee joint occurs in three planes (sagittal, frontal, transverse) with six degrees of freedom (3 rotations, 3 translations, allowing 12 directional motions) between the femoral condyles and tibial plateau (Woo et al., 2006). The knee joint can rotate in the sagittal plane by flexion and extension, in the frontal plane by abduction and adduction, and in the transverse plane by internal and external rotation. The knee joint translation occurs in the sagittal plane anteriorly and posteriorly, in the frontal plane medially and laterally, and in the transverse plane via compression and distraction (Figure 2-1).

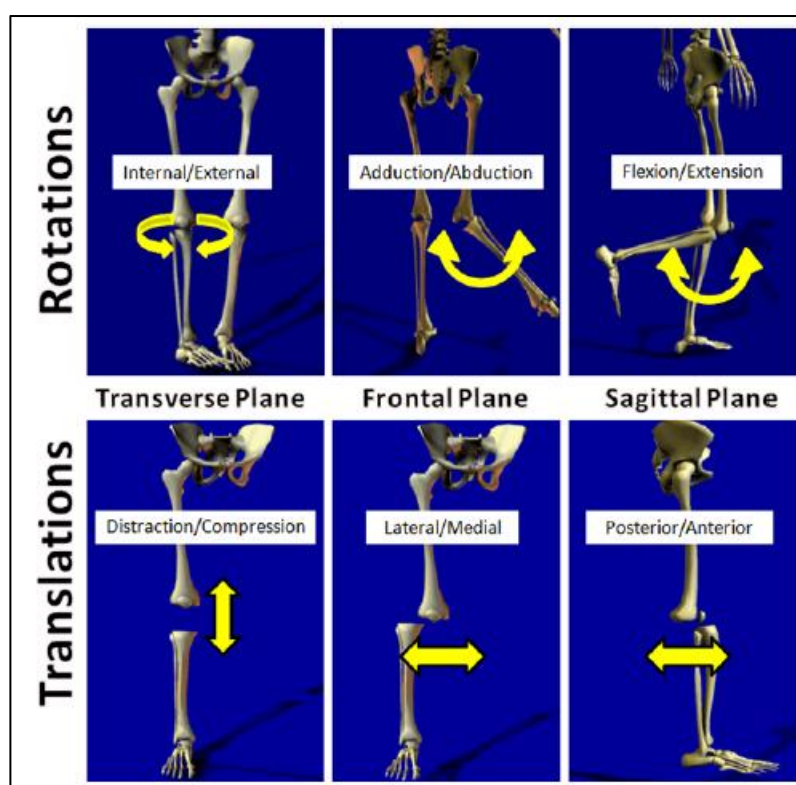


Figure 2-1. Rotations and translations of the knee joint, (Quatman, 2009)

Mechanically, ACL injury occurs when excessive loading is applied on the ACL. Anterior shear force<sup>4</sup> at the proximal end of the tibia is the most direct ACL loading mechanism. Quadriceps muscles are a major contributor to the anterior shear force at the proximal end of the tibia through the patella tendon (Beynnon et al., 1995; Yu and Garrett, 2007). Decreasing knee flexion angles also acts to increase the anterior shear force at the tibia (Arms et al., 1984; Markolf et al., 1995).

Lower limb joint kinematics and kinetics in the transverse plane have also been investigated in relation to ACL injury mechanism. Numerous studies in the non weight-bearing condition have reported higher ACL strains during internal tibial rotation, while only minimal increases

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<sup>4</sup> Shear Force, a force directed parallel or at a tangent to a surface, tends to cause one portion of an object to slide, displace, or shear with respect to another portion of the object. (Kent, 2006)

in strains during external rotation have been noted (Wascher et al., 1993b; Markolf et al., 1995; Fleming et al., 2001a). Similarly, this was also found to be the case with ACL strain due to internal rotation moments increasing as the knee flexion angle decreases (Wascher et al., 1993b; Markolf et al., 1995). Given this strong evidence for internal tibial rotations as an ACL injury mechanism rather than external rotations, this places into question the video –based studies that describe an externally rotated tibia during ACL injury (Krosshaug et al., 2007). It is therefore, interesting to note that studies in weight-bearing conditions show significant increases in the ACL strain during both internal and external moments (Fleming et al., 2001a). Meyer and Haut, (2008) have reported that joint motions can vary in magnitude and direction before and after failure of the ACL. Knee compression experiments showed there was internal rotation of the tibia in pre-failure tests, but external rotation of the tibia after the ACL had failed. Therefore, the video-based studies that document external rotation of the tibia as the mechanism of ACL injury, could in fact be documenting a movement caused by the ACL rupture itself. This is supported by Koga et al., (2010) who reported 3-dimensional knee joint kinematics in anterior cruciate ligament injury situations using a model-based image-matching technique on traditional video-based data.

Lower limb, frontal plane, joint kinematics and kinetics have also been widely associated with ACL injury. Increased knee abduction angles and moments are common gender differences during athletic movements and predictors of future ACL injury risk (Malinzak et al., 2001; Hewett et al., 2004; McLean et al., 2004a; Ford et al., 2005a; Hewett et al., 2005a; McLean et al., 2008). External knee valgus moment has also been shown to prospectively predict ACL injury with 73% sensitivity and 78% specificity (Hewett et al., 2005a). Athletes that went on to ACL injury were also reported to have 8° greater knee abduction angles compared to uninjured athletes (Hewett et al., 2005a). In cadaver knee torsion experiments, failure of the ACL occurred at 58±19° of internal tibial rotation, and valgus rotation of the femur was shown to increase significantly after ACL injury (Meyer and Haut, 2008). Therefore, the valgus motion identified as the mechanism of ACL injury could be similar to tibial external rotation in that it is caused by the ACL rupture itself.

The knee may experience increased loading in any plane, particularly in sporting movements such as landing, jumping, and cutting, all of which are conducted through multiple planes. Thus, it is unlikely that a non-contact ACL injury occurs in any of the above planes in isolation. Video-based studies indicate that non-contact ACL injuries can occur with a combination of knee extension and knee abduction/valgus or transverse knee rotation movements (Boden et al., 2000a; Olsen et al., 2004; Cochrane et al., 2007; Krosshaug et al., 2007). This was also reported to be the case in athlete interview studies (Arnold et al., 1979; McNair et al., 1990). Shimokochi and Shultz (2008), systematically reviewed these video-based and athlete interview studies and found that the principal mechanism appears to be a

result of multi-planar knee loading conditions. Cadaveric investigations have demonstrated that frontal and transverse plane moments, combined with an anterior tibial shear force can increase ACL strain. Markolf et al., (1995) and Berns et al., (1992) demonstrated that combined loading of an abduction or internal tibial rotation moment to an anterior tibial shear force under certain knee flexion conditions leads to additional generation of ACL strain compared to an isolated anterior tibial shear force, abduction or internal tibial rotation moment. In contrast, coupled external tibial and anterior tibial shear force appears to lower the ACL tensile force after 20° of knee flexion (Markolf et al., 1995). Fung and colleagues (2003) developed 3-dimensional models of knees to examine factors that could lead to ACL impingement on the intercondylar notch of the femur. They proposed an ACL injury mechanism where knee valgus with external rotation can lead to impingement of the ACL against the intercondylar notch.

#### *2.4.5 The role of the kinetic chain in ACL injury mechanism*

Motion and forces at any segment of the kinetic chain (foot, ankle, hip, trunk, and upper extremities) may influence knee joint mechanics. Ireland emphasised the role of the hip and pelvis and coined the term “position of no return” (Ireland, 1999). This position of hip flexion, adduction and internal rotation, combined with knee abduction and extension and tibial external rotation details how ineffective neuromuscular control at joints other than the knee can act to produce a multi-planar ACL injury mechanism.

Yu et al., (2006) also emphasised the role of the hip joint in ACL loading through analysis of lower limb joint angular velocities in landing. The hip joint angular velocity at the initial foot contact with the ground was shown to affect the proximal tibia anterior shear force more than the knee joint angular velocity. No such relationship was shown for initial joint angles at contact. Norcross et al., (2010) suggested that active joint motion following ground contact, may be more important than merely the joint positions at initial contact in altering both the magnitude and temporal characteristics of energy absorption. Hashemi et al., (2011) have theorized a new ACL injury mechanism which also includes the role of hip and knee flexion velocities. They propose that ACL injury occurs due to the coexistence of specific neuromuscular events, external loads due to ground contact, and certain anatomical features. The theorized mechanism involves the co-existence of following four elements: a delayed or slow co-activation of quadriceps and hamstrings muscles, a dynamic ground reaction force applied while the knee is near full extension, a shallow medial tibial plateau and a steep posterior tibial slope, and a stiff landing due to incompatible hip and knee flexion velocities (Hashemi et al., 2011).

#### 2.4.5 *Concluding comments*

Video-based, athlete interview and biomechanical investigations examining ACL injury mechanisms are unable to identify the exact moment of ACL injury, therefore, the strongest evidence is found in the simulation, cadaver and in-vivo investigations. These investigations identify the moment of injury and therefore, allow us to distinguish between pre- injury causal motions and motions seen post-injury due to the lack of the ACL. Several theories for the ACL injury mechanism are suggested, however, a lack of conclusive evidence for the existence of one definite mechanism is still a matter of controversy (Krosshaug et al., 2005). It is possible that a number of mechanisms may exist to cause an ACL rupture; these would include combinations of knee flexion, internal rotation and abduction to overload the ACL, or knee flexion, external rotation and abduction to impinge the ACL against the intercondylar notch (Fung and Zhang, 2003).

The neuromuscular control at joints other than the knee has not yet been adequately addressed in the literature. Delayed or slow muscular activation at the hip may force the knee into the unsafe positions mentioned above and force the knee to rely on passive support from structures such as the ligaments. This would be particularly dangerous during high risk deceleration movements. It is paramount that future studies further investigate the role of the hip joint during high risk movement tasks. The investigation of joint motions following ground contact, as opposed to joint alignments and moments at contact and at maximum and minimum values, may provide information on the level of joint control, immediately following landing.

In conclusion the ACL injury mechanism is complex a multi-planar mechanism, which is affected by the rate and magnitude of motion and forces, at any segment of the kinetic chain, during and immediately following ground contact.

## 2.5 **Risk factors for ACL injury**

The potential mechanisms for ACL injury have been outlined in the previous section. There are a number of factors that will increase an individual's risk of receiving an ACL injury. These factors can be subdivided into two distinct categories, factors which are external to the individual and those which are internal or inbuilt in the individual.

#### 2.5.1 *External Risk Factors*

External risk factors are typically risk factors arising from the environment where an athlete performs, the equipment used and human factors of the performance such as actions of teammates opponents etc.

The main risk factors associated with the performance environment are playing surface, level of participation/competition and the chosen sport. Uneven surfaces (Boden et al., 2000a),

high friction (Dowling et al., 2010), synthetic flooring (Olsen et al., 2004) and natural or rye grass (Scranton Jr et al., 1997; Meyers and Barnhill, 2004; Orchard et al., 2005), with limited rain fall (Orchard et al., 1999; Orchard et al., 2001; Orchard and Powell, 2003; Meyers and Barnhill, 2004) are also factors reported to increase the risk of receiving an ACL injury. It is reported that the incidence of general sports injuries (Seil et al., 1998; Emery et al., 2005) and ACL injuries (Scranton Jr et al., 1997; Bradley et al., 2002; Roi et al., 2006; Shankar et al., 2007) is greater in competition than in practice. Certain playing positions (midfielders (Roi et al., 2006)) and sports also have an increased risk of ACL injury. Lacrosse players for example are up to two times more prone to receiving an ACL injury than other high risk sports such as basketball and soccer (Mihata et al., 2006), this is thought to be due to increased levels of physical contact in lacrosse.

Equipment such as sports shoes and field boots, have been shown to significantly interact with underfoot conditions to increase ACL injury risk. Thread sole shoes (Wannop et al., 2010) and cleats with longer irregular cleats positioned at the periphery of the shoe and smaller pointed cleats positioned interiorly (Lambson et al., 1996) are two such examples.

Human factors associated with ACL injury risk include motion perturbations from opponents/teammates, and playing style. This can take the form of a physical perturbation such as being pushed off balance (Olsen et al., 2004) or distraction by external stimuli such as an opponent (McLean et al., 2004b). In-direct ACL injuries as previously described in Section 2.4.2, occur when an individual is perturbed to a body part other than the knee, just prior to or during the ACL injury inciting event. Playing style which although may be genetically based is externally influenced by training and coaching techniques and can also act to increase risk of ACL injury (Boden et al., 2000a; Boden et al., 2000b).

### 2.5.2 *Internal Risk Factors*

Below is a brief outline of each of risk factors that are internal to each individual include **N**euromuscular, **A**natomical, **H**ormonal, **G**ender, **E**thnicity, and **G**enetic risk factors.

**N**euromuscular factors that relate to an increased risk of ACL injuries include altered movement, muscle activation patterns (Boden et al., 2000b), altered lower limb coordination (Pollard et al., 2005) and neuro-cognitive function (Swanik et al., 2007).

Movement patterns that involve movements associated with an ACL injury mechanism are logical risk factors for ACL injury. As discussed in Section 2.4.4 such movements include decreased knee and hip flexion (Arnold et al., 1979; Olsen et al., 2004; Cochrane et al., 2007; Krosshaug et al., 2007; Kobayashi et al., 2010) and increased hip (Ireland, 1999) and tibial internal rotation (Wascher et al., 1993b; Fleming et al., 2001a; Krosshaug et al., 2007; Meyer and Haut, 2008; Koga et al., 2010). Whether the knee abduction/ valgus motion is involved in

the ACL injury mechanism or as a consequence of ACL injury is debatable (Meyer and Haut, 2008). Hewett et al., (2005a) however have reported increased knee valgus loading in athletes who later go on to rupture their ACL. Markolf et al., (1995) have also demonstrated that a quadriceps force applied in combination with a valgus load increased the ACL force compared to valgus loads without a quadriceps force. Markolf et al., (1995) reported similar increases for combinations including tibial internal rotation loads. Therefore, the combination of knee extension and tibia rotation or knee valgus would be considered risky movements. Although external tibial rotation and abduction does not load the ACL (Markolf et al., 1995) the ACL may be injured via its impingement against the lateral wall of the intercondylar notch this will be discussed in the subsequent sub section on anatomical risk factors.

Muscle activation patterns play a vital role in preventing the production of these altered movement patterns as a quadriceps dominant muscular contraction will increase knee flexion and also increase ACL anterior shear forces (Wascher et al., 1993b; Beynnon et al., 1995; Malinzak et al., 2001; DeMorat et al., 2004; Yu and Garrett, 2007) and inadequate muscle contraction will not provide the necessary muscular protection to control the knee (Granata et al., 2002a; Granata et al., 2002b; Wojtys et al., 2003; Griffin et al., 2006) and hip joints (Ireland, 1999; Norcross et al., 2010). Delays in muscular activation (Hashemi et al., 2011) and imbalanced muscle firing rates between the hip and knee flexors (Yu and Garrett, 2007) has also been suggested as a factor to increase the risk of an ACL injury. Adequate pre-activation of the lower limb muscles is essential as ACL injuries are reported to occur in the first 40ms of ground contact (Krosshaug et al., 2007; Koga et al., 2010). Zebis et al., (2009) have reported that reduced preactivation of the semitendinosis muscle (a knee flexor) and increased preactivation of the vastus lateralis muscle (a knee extensor) was shown in athletes who go onto injury their ACL. Increased pre-activation of the quadriceps has been shown to protect the ACL in landing and therefore, reduce ACL injury risk (Hashemi et al., 2010).

The majority of biomechanics literature assessing movement patterns during dynamic tasks, in terms of ACL injury risk, have reported solely on the kinematics of the lower extremity (Nichols et al., 1998; Hewett et al., 2004; Bahr and Krosshaug, 2005). Far fewer have assessed the interaction between the joints (Pollard et al., 2005), which has also been suggested as risk factor for ACL injury (Bates et al., 1978; Yu and Garrett, 2007; Norcross et al., 2010). It is proposed that if timing, and movement velocity discrepancies occurred between the lower limb joint actions the posterior ground reaction force may increase and the tibia may undergo antagonistic counter rotations, leading to excessive stress at the knee joint (Bates et al., 1978; Stergiou and Bates, 1997).

In addition cognitive function and fatigue may have a crucial effect on these aforementioned neuromuscular factors. Swanik et al., (2007) propose that decreased cognitive function may predispose an athlete to ACL injury through loss of neuromuscular control and coordination errors. Borotikar et al., (2008) reported increased initial contact hip extension and internal rotation and peak stance knee abduction and internal rotation angles in a fatigued performance of a side cut. These movement patterns were further exaggerated when an unanticipated cutting task was performed when fatigued. The integrative impact of fatigue and decision making may therefore, present the worst case scenario for high-risk dynamic landing and movement strategies. This links back to the idea the indirect ACL injury mechanism, as a perturbation may have an even greater effect in situations where the athlete is already fatigued. Research in the area of fatigue and ACL injury is inconclusive. Laboratory based investigations assessing movement patterns post fatigue have shown altered movement patterns (Borotikar et al., 2008), but epidemiological data show little increase in the incidence of injury in the final period of competition (Myklebust et al., 1997).

**A**natomical risk factors for ACL injury are widely based around the mechanical alignment of the lower extremity and ACL geometry which contributes to the overall stability of the knee joint. Conflicting evidence exists from a variety of study designs regarding ; Q angle (Boden et al., 2000b; Griffin et al., 2006), knee joint and general joint laxity (Boden et al., 2000b; Uhorchak et al., 2003), ACL geometry (Boden et al., 2000b; Griffin et al., 2006), and composition (Griffin et al., 2006), the width of the femoral notch (Boden et al., 2000b; Griffin et al., 2006), foot pronation (Griffin et al., 2006), tibial plateau geometry (Hashemi et al., 2011), and body mass index (BMI) (Uhorchak et al., 2003; Griffin et al., 2006). The latter of these anatomical risk factors can be considered gender neutral (notch width, BMI, tibial plateau geometry and foot pronation), however the areas of Q-angle, knee valgus, knee joint laxity and ACL geometry and composition are more widely associated with increased risk of ACL injury in the female population and therefore, will be dealt with in the subsequent section (2.6 The gender bias in ACL Injury).

Impingement of the ACL against the intercondylar notch has been proposed as a cause of ACL injuries (Kennedy et al., 1974; Harner et al., 1994; Fung and Zhang, 2003; Fung et al., 2007; Park et al., 2010). An ACL lodged in a narrow A-shaped notch, instead of a reverse U-shaped notch, may experience greater shearing forces against the bone (Boden et al., 2000b). Research on notch dimensions is difficult to interpret due to the lack of standardized methods to obtain the data (Griffin et al., 2006). Arendt, (2000) summarized the existing data on notch width and its relationship to ACL injury, it was noted that notch width (regardless of measurement method utilised) was greater for bilateral ACL-injured knees than that of unilateral ACL-injured knees and greater for bilateral and unilateral ACL-injured knees than notch width of normal controls, implying an association between notch width and injury.



Research suggests that over-pronation of the foot may contribute to the incidence of ACL tears by increasing internal tibial rotation (Woodford-Rogers et al., 1994; Loudon et al., 1996; Smith et al., 1997; Bonci, 1999; Allen and Glasoe, 2000). Studies have documented increased foot pronation or navicular drop in individuals with a history of an ACL tear (Woodford-Rogers et al., 1994; Loudon et al., 1996; Allen and Glasoe, 2000). It is theorised that increased foot pronation leads to increased subtalar pronation and therefore, increased knee joint laxity which may increase ACL injury risk. However, controversy exists as to whether this structural variable is a significant risk for ACL injury (Loudon et al., 1996; Allen and Glasoe, 2000) or simply a side effect in previously injured individuals. Myers et al., (2011) have shown that navicular drop was significantly correlated with anterior tibial translation during landing.

Increased BMI has been associated with landing patterns utilising a more extended lower extremity position with decreased knee flexion velocity, these altered movement patterns may increase risk of ACL injury (Brown et al., 2005). Epidemiological data is inconsistent however providing evidence supporting (Uhorchak et al., 2003) and refuting (Ostenberg and Roos, 2000; Knapik et al., 2001) the correlation between BMI and ACL injury.

Geometry of the tibial plateau has been suggested as a risk factor for ACL injury. Hashemi et al., (2011) suggests that a shallow medial tibial plateau and a steep posterior tibial slope increase the risk of ACL injury. The following studies have supported this showing subjects with previous ACL injuries display this tibial plateau geometry (Stijak et al., 2008; Simon et al., 2010; Todd et al., 2010; Khan et al., 2011).

Mc Lean et al., (2010) suggests that ACL injury risk may increase due to a combination of morphological factors such as ACL anatomy and laxity skeletal alignment and mechanical factors such as the high risk movements identified in Section 2.4.4. Mc Lean et al., (2010) suggest that future injury prevention programmes should be designed in an athlete specific manner catering to the individual morphological vulnerabilities.

**G**ender and its association with ACL injury has been widely discussed and researched, possible risk factors associated solely with females vary from hormonal risk factors to specific anatomical and neuromuscular risk factors. The issue of gender bias in ACL injury and its associated risk factors will be discussed in more detail in the subsequent section (2.6 The gender bias in ACL Injury).

**E**thnicity is also associated with an increased risk of ACL injury. White Europeans are six times more at risk than other ethnic groups and eleven times more at risk than African Americans in professional ladies basketball (Trojian and Collins, 2006). African Americans

have also been shown to have a significantly wider notch width than white Americans which may explain the racial associated differences in ACL injury risk (Shelbourne et al., 2007).

**G**enetic risk factors for ACL injury have been suggested due to a strong familial link in ACL injuries (Harner et al., 1994; Flynn et al., 2005). It has been suggested that the genetic sequence within the genes which code for the extracellular matrix of the ACL may cause this familial risk. The COL1A1, COL5A1 and COL12A1 are gene codes for sub-units of type I, V and XII collagen respectively, which act as components of the ACL. These 3 gene codes have been shown to be associated with ACL ruptures (Posthumus et al., 2009; Collins et al., 2010; Posthumus et al., 2010).

### 2.5.3 *Concluding comments*

The risk factors of most importance to this thesis are those that can be altered through some form of intervention, in order to reduce ACL injury incidence. As the only modifiable risk factor the category most suited to intervention is neuromuscular risk factors. The ability of these interventions to reduce injury risk is underpinned by the validity of the outlined ACL injury risk factors.

The movement patterns that are purported as risk factors for ACL injury risk stem from the literature on the ACL injury mechanism. As previously outlined there are limitations to some of these investigations due to the lack of exact timing for the ACL rupture. Knee joint extension and tibial internal rotation are supported through video based and ACL loading literature as risk factors for ACL injury. The impact of knee abduction and tibial external rotation on ACL loading is still questionable. Video based studies report tibial external rotation and valgus collapse during an ACL injury mechanism (Boden et al., 2000a; Olsen et al., 2004; Krosshaug et al., 2007). ACL loading and cadaver experiments are in disagreement however suggesting that external tibia rotation and knee abduction may be caused by the ACL rupture and not a causal factor in the injury (Meyer and Haut, 2008). Prospective investigations are a definite method of assessing if these risky movement patterns actually result in ACL injury. Hewett et al., (2005a) has confirmed that greater knee abduction motion and moments predict ACL injury in athletes. It is therefore, important that further prospective research is used to confirm if the proposed risky movement patterns result in ACL injury.

Muscular activation has not been measured during ACL injuries and therefore, assumptions of muscular activations during ACL injury are based on the ACL loading literature (Berns et al., 1992; Wascher et al., 1993a; Markolf et al., 1995; Fleming et al., 2001a) and the movements associated with certain muscle activations. Additional evidence informing the proposal of muscular activation risk factors stems from investigations assessing high risk groups e.g. females versus males.

The role of the quadriceps in increasing anterior shear force on the ACL is well documented and the altered hamstring to quadriceps ratio has also been widely suggested as an ACL injury risk factor. It is strongly linked to the knee extension involved in the ACL mechanism with increased quadriceps activation decreasing knee flexion.

The magnitude and timing of muscular activation and pre-activation have received some degree of research attention. Inadequate muscular strength and delayed muscular activation at the hip or knee joint are outlined as potential components of ACL injury mechanism, and are suggested to increase the risk of ACL injury. Limited pre-activation of the muscles of the lower limb is also put forward as a factor increasing risk of ACL injury. These factors are not strongly linked with a definite ACL injury mechanism; they have been however linked with the presentation of risky movement patterns at the hip and knee. Limited pre-activation has also been investigated prospectively to predict future ACL injuries (Zebis et al., 2009). The preparatory and initial activations of the lower limb muscles at ground contact play a vital role in ACL injury risk. Future research needs to prospectively investigate these risk factors and further evaluate the muscular activations in the period just prior to and shortly after ground contact.

As stated very few studies have investigated the interaction between the joints of the lower limb as potential risk factors for ACL injury. The timing and coordination of motions of the lower limb are vital in maintaining a balance that allows for a fluid and controlled movement. The role of movement perturbations in indirect contact ACL injury has been highlighted, an optimal interaction between joints of the lower limb is hypothesised to allow for an increased ability to deal with such perturbations. This perspective is further addressed in Section 2.8.1 Coordinative Structures. There are very few investigations assessing these joint interactions and their association with ACL injury risk, therefore, it is an area that strongly merits further research.

## **2.6 The gender bias in ACL injury**

Epidemiologic injury surveillance data has demonstrated and continues to show although more male athletes suffer ACL injuries compared with female athletes, the incidence rate of ACL tears is greater among females in comparison with males participating in the same sport (Roos et al., 1995b; Bjordal et al., 1997; Myklebust et al., 1997; Myklebust et al., 1998; Arendt et al., 1999; Messina et al., 1999; De Loes et al., 2000; Olsen et al., 2004; Agel et al., 2005; Junge et al., 2006; Mihata et al., 2006; Prodromos et al., 2007; Borowski et al., 2008; Flood and Harrison, 2009; Randazzo et al., 2010; Pappas et al., 2011), with females being 2-6 times more likely to receive the injury than males (Gray et al., 1985; Dick and Arendt, 1993; Roos et al., 1995b; Bjordal et al., 1997; Messina et al., 1999; Junge et al., 2006; Mihata et al., 2006; Borowski et al., 2008; Flood and Harrison, 2009). A number of gender-specific risk factors,

which may account for this alarming rate of ACL injuries in the female population, have been proposed. These include hormonal risk factors, ACL geometry and composition, anatomical risk factors and increased incidence of some of the previously mentioned neuromuscular risk factors all of which are outlined below. Elliot et al., (2010) has recently suggested the inclusion of factors such as burnout, diet, sleep/ fatigue, substance abuse and stress/ psychological issues. These factors are considered to be beyond the scope of this review but merit mentioning.

#### *2.6.1 Hormonal Risk Factors*

Female sex hormones (i.e. oestrogen, progesterone and relaxin) fluctuate radically during the menstrual cycle and are reported to increase ligamentous laxity and decrease neuromuscular performance. It is thought that these fluctuations may cause decreases in both passive and active knee stability in female athletes.

Receptors for the hormones oestrogen, progesterone and relaxin have been identified in the human ACL (Liu et al., 1996; Dragoo et al., 2003). This could have a direct regulatory effect on fibroblast function, collagen remodelling and, ultimately, alter the structural, material and mechanical properties of the ACL in vivo (Liu et al., 1997; Slauterbeck et al., 2002). Increased anterior motion at the knee has been reported during different phases of the menstrual cycle (Heitz et al., 1999; Deie et al., 2002; Shultz et al., 2004; Shultz et al., 2005; Park et al., 2009a; Park et al., 2009b; Shultz et al., 2010) but other reports refute this concept (Karageanes et al., 2000; Arnold et al., 2002; Romani et al., 2003; Van Lunen et al., 2003; Beynnon et al., 2005; Hertel et al., 2006; Eiling et al., 2007). The additional presence of oestrogen and progesterone receptors in portions of the brain associated with movement patterns (McEwen et al., 2001), has implications for neuromuscular performance, as oestrogen levels have been reported to decrease the performance of fine motor skills (Posthuma et al., 1987). Different phases of the menstrual cycle have been reported to result in significant slowing of muscle relaxation, increased muscle fatigability (Sarwar et al., 1996) and reduce balance, kinaesthesia and postural control (Fridén et al., 2003; Fridén et al., 2005). The actual effect of anterior knee laxity and the aforementioned decreases in neuromuscular function on ACL injury risk is still unclear.

Menstrual cycle phase and ACL injury risk have been analysed retrospectively, with increased incidence of noncontact ACL injuries during both peri-menstrual (Myklebust et al., 1998; Slauterbeck et al., 2002), and pre-ovulatory days of the menstrual cycle (Arendt et al., 2002; Wojtys et al., 2002; Beynnon et al., 2006). Women using oral contraceptives (which lower estrogen levels) have been shown to have a lower ACL injury rate than those who did not (Moller-Nielsen and Hammar, 1989). Only three of these studies measured actual hormone levels to assess menstrual phase (Slauterbeck et al., 2002; Wojtys et al., 2002; Beynnon et al.,

2006), and of these no conclusive menstrual phase was identified as when an ACL injury was most likely to occur. Many questions still remain on the issue of sex hormones and increased risk of ACL injury. It is accepted that a mounting body of evidence indicates that sex hormones mediate cyclic increases in knee laxity but further research is needed to determine the implications of these cyclic fluctuations on knee joint stability and corresponding injury risk.

### 2.6.2 *ACL composition*

The size of the ACL in females has been shown to be smaller in length, cross-sectional area, volume and mass when compared with males (Anderson et al., 2001; Chandrashekar et al., 2005). When the female ACL was mathematically modelled based on these structural qualities, female ACLs were reported to rupture in response to comparatively smaller external load applications (Chandrashekar et al., 2005; Mizuno et al., 2009). A positive correlation has also been shown between smaller ACLs reported in women and injury risk (Shelbourne and Kerr, 2001; Uhorchak et al., 2003; Chandrashekar et al., 2005). An interesting addition to possible differences between genders in ACL composition is the novel finding, that the *COL5A1* and *COL12A1* genes are underrepresented in female participants with anterior cruciate ligament ruptures (Posthumus et al., 2009; Posthumus et al., 2010). The *COL5A1* gene encodes for the  $\alpha 1$  chain in type V collagen, which is an important structural constituent of both ligaments and tendons (Hildebrand et al., 2004). The fact that the *COL5A1* gene has also been associated with a decreased risk of chronic achilles tendinopathy in men and women (Mokone et al., 2006; September et al., 2009) is very interesting to note and qualifies the proposition that an underrepresentation of this gene in females may predispose them to ACL injury. It may be said that ACL geometry and composition differs between genders but as yet there is no explanation as to how these factors may act to increase the susceptibility of females to ACL injury.

### 2.6.3 *Anatomical Risk Factors*

Various anatomical risk factors have been solely attributed to females such as Q-angle (Hvid et al., 1981; Horton and Hall, 1989; Livingston, 1998; Moul, 1998; McKeon and Hertel, 2009), genu recurvatum, greater anterior pelvic tilt, femoral anteversion (McKeon and Hertel, 2009) and knee joint laxity. Increased Q-angle or quadriceps femoris angle is proposed to alter lower extremity kinematics (Heiderscheit et al., 2000; Mizuno et al., 2001). Static Q-angle values in combination with measures of lower limb strength have been shown to predict dynamic knee valgus on landing (Buchanan, 2003) cited in (Griffin et al., 2006; Alentorn-Geli et al., 2009; Bush, 2010), with increased Q-angle producing larger valgus angles. Females have been shown to land with greater total valgus knee motion and greater maximum valgus knee angle in comparison to males (Ford et al., 2003), which has been shown to be predictive of future ACL injury (Hewett et al., 2004; Hewett et al., 2005a). Altered pelvic tilt, femoral

anteversion, and navicular drop have resulted in increases in hip internal rotation and knee external rotation (Nguyen et al., 2011); movements associated with ACL injury. Increased rotational knee joint laxity in females which can occur due to factors such as hormone levels and lack of muscular protection (Rozzi et al., 1999; Wojtys et al., 2003), has also been shown to produce movements associated with ACL injury (Shultz and Schmitz, 2009).

#### 2.6.4 *Biomechanical and Neuromuscular Risk Factors*

Many of the neuromuscular risk factors for ACL injury are shown to a greater extent in females. To recap these include, **J**oint mechanics, **M**uscle activation patterns and **C**oordination.

**J**oint mechanics have been shown to vary according to gender in numerous investigations. Females have been shown to demonstrate altered joint mechanics at the hip knee and ankle in comparison to males when completing dynamic tasks such as landing and cutting see Table 2-2.

Table 2-2. Gender differences in joint angles and external joint moments

	Landing Tasks	Cutting Tasks
Hip	< Flexion (Salci et al., 2004; Schmitz et al., 2007; Beutler et al., 2009)	(McLean et al., 2004b; McLean et al., 2005b; Landry et al., 2007b; Pollard et al., 2007; Fedie et al., 2010)
	> Int. Rot. <b>Angle</b> (Lephart et al., 2001; Lephart et al., 2002; Brown et al., 2009)	(Pollard et al., 2007)
	< Int. Rot. <b>Angle</b>	(McLean et al., 2004b)
	> Ext. Rot. <b>Angle</b>	(Landry et al., 2007b)
	> Adduction <b>Angle</b> (Lephart et al., 2001)	
	< Adduction <b>Moment</b>	(Pollard et al., 2004a)
	> Adduction <b>Moment</b>	(Pollard et al., 2007)
	< Extension <b>Moment</b>	(Pollard et al., 2007)
Knee	< Flexion <b>Angle</b> (Huston et al., 2001; Lephart et al., 2002; Decker et al., 2003; Salci et al., 2004; McLean et al., 2005b; Schmitz et al., 2007; Beutler et al., 2009)	(Malinzak et al., 2001; James et al., 2004; McLean et al., 2004b; McLean et al., 2005b; Fedie et al., 2010)
	> Flexion <b>Angle</b> (Fagenbaum and Darling, 2003)	
	> Abduction <b>Angle</b> (Ford et al., 2003; Hewett et al., 2004; Hewett et al., 2005a; Kernozek et al., 2005; Ford et al., 2006; Russell et al., 2006; McLean et al., 2007; Pappas et al., 2007; Hughes et al., 2008; Beutler et al., 2009; Gehring et al., 2009; Schmitz et al., 2009; Ford et al., 2010)	(Malinzak et al., 2001; Ford et al., 2005b; McLean et al., 2005b; Beaulieu et al., 2009)
	> Int. Rot. <b>Angle</b> (McLean et al., 2007; Brown et al., 2009)	
	< Int. Rot. <b>Angle</b>	(McLean et al., 2004b)
	> Int. Rot. <b>Angle</b> (Nagano et al., 2007; Kiriyaama et al., 2009)	
	< Int. Rot. <b>Angle</b>	(Lephart et al., 2001)
	< Flexion <b>Moment</b>	(Sigward and Powers, 2006)
	> Extension <b>Moment</b> (Chappell et al., 2002)	
	> Abduction <b>Moment</b> (Chappell et al., 2002; McLean et al., 2007)	
	> Adduction <b>Moment</b> (McLean et al., 2007)	(Sigward and Powers, 2006)
	> Abduction <b>Moment</b> (Kernozek et al., 2005)	
	> Int. Rot. <b>Moment</b> (McLean et al., 2007)	
Ankle	> Eversion <b>Angle</b> (Ford et al., 2006)	(Ford et al., 2005b)
	> Pronation <b>Angle</b> (Kernozek et al., 2005; McLean et al., 2007)	(McLean et al., 2004b)
	> Plantar Flexion <b>Angle</b> (Decker et al., 2003; Kernozek et al., 2005; McLean et al., 2007)	
	> Dorsi-Flexion <b>Angle</b> (Kernozek et al., 2005)	
	> Flexion ROM (Kernozek et al., 2005)	
	< Dorsi-flexion <b>Moment</b> (McLean et al., 2007)	

In addition to the above there are investigations that report no differences between genders for the above lower limb kinematics (McLean et al., 1999; Cowling and Steele, 2001; Pollard et al., 2004a; Sigward and Powers, 2006) and kinetics (Pollard et al., 2004a), and no single investigation has shown all of the above kinematics or kinetics to differ between gender for one population. The variety of research methodologies utilised to come to the above conclusions is a concern. The majority of these risk factors are reported during jumping and landing tasks, a smaller proportion are during cutting. The lack of match specificity in these movement tasks is also a concern with limited investigations utilising unanticipated (Ford et al., 2005b; Landry et al., 2007c; Beaulieu et al., 2009; Brown et al., 2009) tasks that replicate match intensity (Pollard et al., 2004a; McLean et al., 2005b) and conditions (McLean et al., 2004b; Fedie et al., 2010). Other inconsistencies within these investigations include the subject populations; skill level has been shown to impact on ACL injury incidence (Bjordal et al., 1997; Orchard et al., 2001), movement patterns in high risk tasks (McKinley and Pedotti, 1992) and movement variability (McLean et al., 1999) and should be controlled within investigations. There are a limited number of studies that investigate three dimensional hip joint kinematics (Lephart et al., 2001; McLean et al., 2005b; Yu et al., 2005; Sigward et al., 2007) and kinetics (McLean et al., 2004b; Pollard et al., 2004a; Kernozek et al., 2005; McLean et al., 2005a; Hewett et al., 2006; Landry et al., 2007b; McLean et al., 2007; Brown et al., 2009; Fedie et al., 2010), which are thought to play a vital role in control of knee joint motions. Investigation of knee joint kinetics is also limited to a small number of studies (McLean et al., 2004b; Hewett et al., 2005a; Kernozek et al., 2005; McLean et al., 2005b; Sigward and Powers, 2006; Landry et al., 2007b; McLean et al., 2007; Sigward et al., 2007; Brown et al., 2009; Fedie et al., 2010; Sigward et al., 2010). Few investigations assess risk for both legs according to dominance and it has been more common to solely assess the dominant leg as a representation of the lower limb movement (Ford et al., 2003; Hewett et al., 2004; Ford et al., 2005b; Hewett et al., 2005a; Ford et al., 2006; Ford et al., 2010). Limb dominance has been suggested as a factor affecting ACL injury and has been reported to differ according to gender (Negrete et al., 2007; Brophy et al., 2010); therefore, the investigation of both dominant and non-dominant legs is recommended in future research. One final limitation to the research identifying various joint mechanics as ACL injury risk factors is the overreliance on discrete measures. All except one of the cited studies reports discrete measures such as joint alignments at initial contact, maxima and minima, time to peak to name just a few. There needs to be more research utilising continuous methods such as principal component analysis (PCA) (Landry et al., 2007b). Further inspection of joint motions in the initial period following ground contact as opposed to joint motion at contact has also been suggested provide information on joint control on landing (Norcross et al., 2010).



**M**uscular activation pattern alterations, in all three axes of motion (proximal-distal, anterior-posterior, medial-lateral) may contribute to the gender bias in anterior cruciate ligament (ACL) injury rates.

Decreased activation of the musculature at the proximal end of the kinetic chain, e.g. hip and trunk, may lead to lower extremity malalignment as outlined above. Neuromuscular control<sup>5</sup> at the hip in dynamic running tasks (Ferber et al., 2003) and single leg squats (Zeller et al., 2003) has been suggested to be functional as a means of countering injury inducing valgus loads (Lloyd and Buchanan, 2001; Zhang and Wang, 2001; Besier et al., 2003). Females are reported to have lower levels of activation in the gluteal (Zazulak et al., 2005) and core musculature (Zazulak et al., 2007; Brophy et al., 2009). Zazulak et al., (2007) report that core stability predicts the risk of athletic knee, ligament, and ACL injuries with high sensitivity and moderate specificity in females. Chimera et al., (2004) assessed the effects of a plyometric training programme on muscular activation in landing. They reported increased activation of the hip musculature post training intervention, strongly supporting the role of hip muscle activation in dynamic restraint and control of lower extremity alignment. Pollard et al., (2010) also state that females with limited hip flexion had increased knee valgus movement and risk of ACL injury. Athletic females have also been reported to have increased activation of the gastrocnemius and soleus muscles (Padua et al., 2005; Landry et al., 2007a; Landry et al., 2009). Contraction of the gastrocnemius muscle produces ACL strain at low knee flexion angles which is magnified when combined with quadriceps contraction (Fleming et al., 2001b). This highlights the role of the proximal end of the kinetic chain as a possible contributor to ACL injuries in females.

Females demonstrate muscle activations that are quadriceps dominant with less involvement from the hamstrings (Hewett et al., 1996; Huston and Wojtys, 1996; Wojtys et al., 1996; Lephart et al., 2001; Malinzak et al., 2001; Padua et al., 2005; Zazulak et al., 2005; Ahmad et al., 2006; Padua et al., 2006; Nagano et al., 2007; Beaulieu et al., 2008; Hanson et al., 2008; Myer et al., 2009; Ebben et al., 2010) which will produce increased knee extension, and concurrently increase anterior shear force due to the forceful eccentric quadriceps contraction on landing (Wascher et al., 1993a; Beynnon et al., 1995; Markolf et al., 1995; DeMorat et al., 2004; Myer et al., 2005; Yu and Garrett, 2007).

Mediolateral muscular imbalances have also been reported in females. An increase in lateral gastrocnemius (Landry et al., 2007b; Beaulieu et al., 2009) and decrease in hip adduction activation (Brophy et al., 2010) will result in increased hip and knee abduction and increased ACL injury risk.

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<sup>5</sup> "Neuromuscular control is defined as the unconscious efferent response to an afferent signal regarding dynamic joint stability" (Mandelbaum et al., 2005)

Coordination and timing of the above muscular activations are also very important and have been shown to differ by gender. Cowling and Steele, (2001) describe how male hamstring muscular activation is timed to coincide with peak anterior shear forces in order to better protect the ACL via increased joint compression and posterior tibial drawer. This was not replicated in females, which may increase their susceptibility to ACL injury. Gehring et al., (2009) report that females have un-synchronous timing of mediolateral hamstring and quadriceps musculature. This balance which was present in male subjects is necessary to avoid excess knee abduction and ACL injury risk (Hewett et al., 2005b; Myer et al., 2005). Females have been shown to display increased peak knee abduction and knee abduction velocity during the deceleration phase of landing in comparison to the acceleration phase for males (Joseph et al., 2011), this rapid valgus collapse in the initial deceleration phase of landing may be due to a lack of muscular pre-activation (Beaulieu et al., 2008; Medina et al., 2008; Gehring et al., 2009) and protective stiffness around the knee joint (Wojtys et al., 1996; Granata et al., 2002a; Granata et al., 2002b) which is common in females.

## **2.7 The Fate of the ACL injured patient**

Following diagnosis of an ACL injury, patients have two treatment options, repair the ACL surgically or rehabilitate the knee conservatively. Conservative rehabilitation, involves muscular strengthening, agility skill training and sport-specific skill training. It is recommended for patients who are willing to avoid high risk activities/ sports involving following rehabilitation. Surgical rehabilitation on the other hand is recommended for patients who intend to return to sports participation at a high risk level. There are two graft types (allograft<sup>6</sup> or autograft<sup>7</sup>) used to replace the torn ACL; hamstring or patella tendon. Following surgery patients are required to complete a 6-9 month rehabilitation programme, including strength, balance, agility, sports specific skills and plyometric activities.

The purpose of ACL reconstruction and rehabilitation is to restore normal functionality of the knee joint (Chmielewski et al., 2002). Surgery's ability to achieve this normal functionality is assessed with either static measures (i.e., KT-1000, pivot- shift test) or questionnaires (i.e., Lysholm score) or combinations of both (i.e., International Knee Documentation Committee score) (Decker et al., 2011). The ability of these measures to assess the dynamic functionality of the reconstructed knee in sporting activities is limited. A review by Ardern et al., (2010) found that 85-90% of patients demonstrated successful rehabilitation when tested using the above methods, but rate of return to pre-injury sports participation was only 63%. Inadequate dynamic functionality in sports activities or fear of re-injury (Kvist et al., 2005)

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<sup>6</sup> An allograft is a graft of tissue obtained from a donor genetically different from, though of the same species as the recipient.

<sup>7</sup> An autograft is tissue that is taken from one part of a person's body and transplanted to a different part of the same person.

may account for this decreased rate of return. Those who are successful in returning to sports are however at an increased risk of re-injury and degenerative joint disease.

### *2.7.1 Risk of Re-Injury*

Prior injury is one of the single best predictors of future injury risk (Ekstrand and Gillquist, 1983; Tropp et al., 1984; Moller-Nielsen and Hammar, 1989; DuRant et al., 1992; Arnason et al., 1996; Dvorak et al., 2000; Hewett et al., 2005a; Hägglund et al., 2006). Previously ACL rehabilitated (ACLR) individuals have an increased risk of re-rupturing their ACL graft, and their contralateral non-injured knee.

Re-rupture of the ACL graft occurs at a rate of 0-18% (Kvist, 2004; Myklebust and Bahr, 2005). ACLR individuals are 3.2 times more likely to re-injure their previously injured knee (Oates et al., 1999) than an individual with no risk of ACL injury (nACL). This risk of graft failure is greatest in the first year after returning to sport (Orchard et al., 2001; Salmon et al., 2005; Tanaka et al., 2010). Trauma, poor surgical technique, concurrent knee injury, failed incorporation of the graft, high activity levels on return to sport and use of an allografts are suggested as possible reasons for graft failure (George et al., 2006; Borchers et al., 2009; Swärd et al., 2010; Kaeding et al., 2011).

The contralateral non-injured knees of ACLR individuals are at increased risk of ACL injury in comparison to nACL knees (Shelbourne et al., 1998; Orchard et al., 2001). The contralateral knee of ACLR individuals is 6.2 times more likely than an nACL individual's knees (Oates et al., 1999). This risk is greatest two to three years after returning to sport (Salmon et al., 2005). Receiving the initial ACL injury at a younger age (Dunn et al., 2004; Roi et al., 2006) and having a narrower notch width (Souryal et al., 1988; Shelbourne et al., 1998) are suggested as factors to increase risk of contralateral ACL injury in ACLR individuals.

### *2.7.2 Risk of Degenerative Joint Disease*

Up to 50% of ACLR individuals will display signs of osteoarthritis (OA) 10 years post injury (Roos et al., 1995a; Von Porat et al., 2004; Pinczewski et al., 2007; Meuffels et al., 2009; Holm et al., 2010; Øiestad et al., 2010). Factors thought to increase the chances of developing OA are the increased incidence the knee giving way (Jomha et al., 1999), meniscectomy and chondral damage (Keays et al., 2010). There has been no proven link between graft choice and the development of OA (Holm et al., 2010). ACL injuries in isolation contain a significant risk of OA (Gillquist and Messner, 1999), and it may be that the initial injury itself is the important determinant of the development of OA, no matter what treatment is used or how the knee is loaded during subsequent years.

### 2.7.3 *Neuromuscular risk factors for re-injury and degenerative joint disease.*

Altered biomechanics and neuromuscular function as a result of the initial ACL injury, affecting both the injured and the contralateral leg, are likely to increase the risk of a repeated ACL injury (Swärd et al., 2010) and degenerative joint disease (Deneweth et al., 2010).

ACLR individuals have shown altered joint mechanics in tasks such as hopping, pivoting and walking for both the previously injured and contralateral leg (Ristanis et al., 2003; Deneweth et al., 2010; Gao and Zheng, 2010; Orishimo et al., 2010). When compared to the contralateral knee the previously injured knee of ACLR individuals executes hopping movements with greater knee extension and external tibial rotation on contact, and greater maximum knee flexion, external tibial rotation and anterior tibial translation (Deneweth et al., 2010). When ACLR individuals were compared to nACL subjects during a pivoting manoeuvre after stair descent, tibial rotation was not restored to normal levels following surgery and rehabilitation (Ristanis et al., 2003). This was replicated in walking gait (Gao and Zheng, 2010). These kinematic alterations have previously been referred to as risk factors for initial ACL injury; therefore, it is likely that they may also act as risk factors for re-injury in ACLR individuals. Regular repetition of these altered kinematics and associated abnormal loading cycles during daily activity could exaggerate the risky biomechanical factors and progressively lead to cartilage degeneration and premature osteoarthritis in ACLR knees (Gao and Zheng, 2010). Altered levels of movement variability in ACLR individuals, has also been suggested as a risk factor for this increase in injury risk and development of degenerative joint disease. See Section 2.8, for a further explanation.

It is essential that future research on ACLR knees utilise tasks that more closely replicate sport specific demands that ACLR individuals are likely to encounter on their return to sport. The concept of movement variability in ACLR individuals has limited previous investigations and strongly merits further study. This future work will provide new information regarding possible mechanisms for development of degenerative joint disease and increased risk of re-injury.

## **2.8 Movement variability and injury risk**

From a dynamical systems perspective, human movement is a process utilising a highly intricate network of inter-reliant sub-systems (e.g. circulatory, respiratory, perceptual, nervous and musculoskeletal), comprised of many of interacting components (e.g. blood cells, oxygen molecules, muscle tissue, metabolic enzymes, connective tissue and bone) (Glazier et al., 2003). Every human movement therefore, exhibits considerable movement variability in all movement parameters.

### *2.8.1 Dynamical Systems Theory*

Movement variability can be described as variations in motor performance across multiple repetitions of a task (James et al., 2000). It is linked to the complex interactions between the performer and the environment, the nervous system and the movement pattern apparatus, and the disproportionate number of degrees of freedom emerging from these interactions (Lin et al., 1997; Robertson and Miall, 1997; Desmurget et al., 1998). Movement variability has traditionally been viewed as dysfunctional and a reflection of noise in the central nervous system (Newell and Corcos, 1993), error in the planning and execution of movement (Schmidt et al., 1979; van Emmerik and van Wegen, 2000) which is detrimental to skilled coordinated performance. In light of this a 'skilled performer' was one who was thought to demonstrate an ideal, invariant performance of movement patterns (Slifkin and Newell, 1998). In contrast dynamical systems theory suggests that movement variability is an intrinsic feature of skilled motor performance, with variability as an emergent property of self-organizing behaviour within the neuromuscular system (van Emmerik et al., 2005), helping individuals explore their environment to pick up information for actions (Davids et al., 2003). Healthy and adaptive biological systems depend on variability to ensure optimal performance (Collins, 1999; van Emmerik and van Wegen, 2000). This variability allows for the development of coordination patterns that turn the associated joints, segments, muscles, and motor units into controllable systems (Bernstein, 1967).

### *2.8.1 Coordinative structures*

In terms of a biomechanical system, the high number of available degrees of freedom is reduced through the formation of coordinative structures. Coordinative structures can be defined as muscle synergies, often spanning several joints, which are functionally linked to satisfy the task demands (Turvey, 1990). Coordinative structures enable the organism to achieve (a) the same outcome by using different degrees of freedom (e.g. muscles, joints) and (b) use the same degrees of freedom to attain different outcomes (Hamill et al., 1999). Variations in the coupling of these degrees of freedom in a coordinative structure provide a rationale for situations when repeated attempts of the same task show deviations in both kinematic and kinetic parameters even when environmental factors remain constant and the movement is highly trained (Müller and Loosch, 1999; Kurz and Stergiou, 2004). This provides a dimension of flexibility which allows adaptations to complex dynamic sport environments (Kelso and Tuller, 1984; Scholz, 1990; Turvey, 1990; Williams and Davids, 1999; Mullineaux and Uhl, 2010).

### 2.8.1 *Coordination Variability and Injury Risk*

The most interesting application of coordination variability has been its potential role in susceptibility to injury. Various investigations have shown individuals who have received lower limb injuries demonstrate differing levels of coordination variability in comparison to injury free individuals (Stergiou and Bates, 1997; Hamill et al., 1999; James et al., 2000; Heiderscheit et al., 2002; Hamill et al., 2005; Lewek et al., 2006; Kiefer et al., 2008; Moraiti et al., 2010) but it is unclear if this was symptomatic of the existing injury or present prior to injury and acted as a causal factor. Interesting relationships have also been uncovered between gender, previous ACL injury and movement variability.

Anatomical changes associated with maturation in females such as flexibility and skeletal alignment may limit individuals to a narrow range of lower-limb postures and therefore, also limit the number of movement solutions or coordination patterns athletes can utilize in various sporting tasks. Mc Lean et al., (1999; 2005b) both showed increased variability in hip internal rotation and frontal knee joint laxity. Findings presented by (Pollard et al., 2005; O'Connor and Bottum, 2009) were in contrast to this as they show decreased variability in female athletes. There is a distinctive difference between the investigations of McLean et al., (1999; 2005b) and those of O'Connor and Bottum, (2009) and Pollard et al., (2005). McLean considers variability from discrete joint variables whereas O'Connor and Bottum, (2009) and Pollard et al., (2005) utilised continuous measures of variability i.e. Principal Component Analysis<sup>8</sup> showing decreased levels of sagittal and frontal plane knee movement and Continuous Relative Phase<sup>9</sup> showing decreased levels of coordination variability for thigh/leg rotation and hip rotation/knee abduction adduction couplings in females.

All of the above investigations reported on gender differences during cutting tasks. Future research looking at other high risk tasks such as landing, are recommended to further inform this debate. The use of continuous measures of variability is recommended as they will provide a more comprehensive description on the variability demonstrated by both genders.

Individuals with previous ACL injury have also been reported to have altered levels of movement variability. As outlined in the previous section (2.7 The Fate of the ACL Injured patient) those who receive an ACL injury are at increased risk of both re-injury and degenerative joint disease such as osteoarthritis. Individuals with a deficient and reconstructed ACL have been reported to have altered levels of movement variability

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<sup>8</sup> Principal component analysis is used when you have obtained measures on a number of observed variables and wish to develop a smaller number of artificial variables (called principal components) that will account for most of the variance in the observed variables. The principal components may then be used as predictor or criterion variables in subsequent analyses. It may also be used to extract common modes of variability from a data set and can partition variability into random and deterministic components (Daffertshofer et al. 2004).

<sup>9</sup> Continuous relative phase is a method of quantifying interlimb coordination (Hamill et al. 1999)

(Stergiou et al., 2004; Georgoulis et al., 2006; Moraiti et al., 2007; Moraiti et al., 2010; Zampeli et al., 2010).

ACL deficient (ACLD) individuals were reported to exhibit a less variable and rigid behaviour in their knee movements in comparison to their contralateral non-injured leg (Stergiou et al., 2004; Georgoulis et al., 2006) and to matched nACL controls (Moraiti et al., 2007; Zampeli et al., 2010) during walking gait. On the other hand, after ACL reconstruction the knee demonstrates a more variable and unpredictable behaviour (Moraiti et al., 2010) in comparison to matched nACL subjects. All of above investigations assessed kinematic variability using non-linear measures, Lyapunov Exponent<sup>10</sup> or Approximate Entropy<sup>11</sup> for the times series' of knee flexion-extension. Kiefer et al., (2008) have addressed this relationship measuring postural coordination in balancing tasks.

ACL rehabilitated individuals in Kiefer et al.,'s (2008) study demonstrated a more variable hip-ankle coordination during a slow balance task in comparison to control subjects, similar to Moraiti et al., (2010). Tim Hewett a member of the research team involved in Kiefer et al.,'s (2008) study, expanded on these results in a roundtable discussion published in BioMechanics Magazine (Foster, 2009). A small cohort of the ACLr subjects tested in Kiefer's study went on to receive a second injury. During the slow balance task, these ACL re-injury subjects demonstrate very high variability—higher than the other ACLr subjects. During the fast balance task, these ACL re-injury subjects locked down the lower extremity and limited the degrees of freedom due to increased task difficulty, and their variability decreased. All these responses were in opposition to what was demonstrated in the nACL control subjects. When ACLr subjects were exposed to a demanding task they exhibited decreased variability similar to that demonstrated by the ACLd subjects tested by Stergiou et al., (2004). It may be hypothesised that the variability reported may be functional to the demands of the task. ACLr individuals feel “secure” enough to increase and add extra movement during gait however since the proper proprioceptive channels are not there, the temporal structure of the stride-to-stride variations of the knee is not restored to normative levels. On the contrary, the rigidity found in the ACLd knees signifies that they are more “careful” in the way they walk trying to eliminate any extra movements (Decker et al., 2011). These “careful” rigid movements are replicated by the ACLr individuals in the more difficult balancing task similar as would be expected when an individual is introduced to a new skill (Bernstein, 1967). Stergiou and Decker, (2011) suggest an optimal level of variability for various movements

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<sup>10</sup> A non-linear measure that estimates the underlying structure of variability during movement, it is particularly useful for periodic movements (Moraiti et al., 2010).

<sup>11</sup> A non-linear measure that evaluates the regularity or predictability of a system (Moraiti et al., 2010).

where repetitive execution of the movement with levels of variability above or below an optimal level would increase injury risk.

The relationship between previous ACL injury and movement variability is an area lacking investigation. Its potential as a risk factor for re-injury and development of degenerative joint disease has been outlined and strongly merits further investigation. The task demands presented to ACLr individuals in future testing should utilise tasks that have a risk of ACL injury and replicate sport's demands, as past investigations have only looked at gait and balance tasks. Due to the fact that movement variability may be specific to the task demands for each individual, it is essential that the subject population utilised are fully rehabilitated and have returned to full sports participation.

If these proposed altered levels of variability are in fact present in individuals such as female or ACLr athletes, there is substantial theoretical reasoning to suggest this may increase their risk of injury, re-injury or future pathology. Decreased levels of variability in complex tasks may limit the ability of an individual to adapt to perturbations such as uneven playing surface (Boden et al., 2000b), being off balance, being pushed or held by an opponent or trying to evade a collision with another player (Krosshaug et al., 2007). Due to the fact that >50% of ACL injuries occur closely following an induced perturbation (Krosshaug et al., 2007) the role of decreased variability is especially relevant for ACL injury and worthy of further investigation.

## **2.9 Experimental Design issues**

Existing research examining potential ACL injury risk factors, and comparing ACL injury risk across **P**opulations, has measured a variety of **P**arameters during a range of **M**ovement tasks, utilising various **B**iomechanical models. This has led to some inconsistent and conflicting conclusions.

**P**opulations such as females or ACLr individuals are widely utilised in ACL injury risk studies. The criteria used in the subject selection for these populations are vital. The level of competition (e.g. recreational, amateur, elite/professional), sport, playing position, gender, height and weight have all been shown to affect ACL injury incidence (Bjordal et al., 1997; Orchard et al., 2001), movement patterns in high risk tasks (McKinley and Pedotti, 1992) and movement variability (McLean et al., 1999). Therefore, these factors should be considered and matched between subject groups where appropriate. In terms of ACLr individuals, the above factors should be accounted for when selecting nACL control subjects. Time since; initial injury, surgery and return to sport should be also considered when selecting control subjects due to the effect of these time lapses have on risk of ACL re-injury, the prevalence of OA and the individual athlete's perception of task demands.



**P**arameters measured in biomechanical experiments, should be closely aligned with the ACL loading patterns, injury mechanisms and neuromuscular risk factors which have previously been reported (2.4 ACL Injury Mechanism, 2.5.2 Internal Risk Factors) in order to adequately define injury risk. These include:

- Multi-planar kinematics of the trunk, thigh, shank, and joints of lower extremities.
  - Their timing, velocity, and coordination.
- Multi-planar kinetics of the joints of lower extremities.
- Muscular activations of the lower extremities.
- Variability of the above joint mechanics and their coordination.

It is important that future investigations into injury risk in females and ACLr individuals move away from the overreliance on discrete measures. The use of continuous methods in the measurement of movement variability and movement patterns is essential to provide new information on the risk factors for ACL injury in certain populations.

Previous investigations have provided limited information on the three dimensional mechanics of the lower limb joints other than the knee, therefore, further investigation of the contribution of the hip and ankle is merited.

Data on lower limb joint kinetics has also been limited in previous research and merits future inclusion. The combined assessment of, three dimensional joint kinetics and kinematics of the lower limb is essential in clearly identifying risky movements and their associated loading on the knee joint.

Previous measurement of these parameters on both dominant and non-dominant legs has been limited. Due to altered injury rates between legs for nACL and ACLr individuals analysis and comparison of both legs is essential to allow for a comprehensive conclusion on an individual's risk of ACL injury during high risk tasks.

**B**iochemical models and associated marker sets chosen need to allow accurate three dimensional measurement of these parameters. Three dimensional measurement of joint and segment kinematics requires the development of a global and local/ joint or segment coordinate system (Hamill and Selbie, 2004). Definitions of local coordinate systems require calculation joint centres and rotations in all three planes of motion. A variety of marker sets can be employed to define the local coordinate system; these will vary according to location on anatomical landmarks and the use of marker clusters/ arrays<sup>12</sup>. An array of three non collinear markers on a bone pin is the gold standard for definition of segment rotations

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<sup>12</sup> Marker clusters are a group of three to four markers attached to the body segment on a rigid surface.

(Fuller et al., 1997; Reinschmidt et al., 1997; Benoit et al., 2006). The least accurate method uses three non-collinear markers placed on the skin, as the markers move independently of each other (Fuller et al., 1997; Reinschmidt et al., 1997; Karlsson and Tranberg, 1999). Reinschmidt et al., (1997) reported good agreement between skin and bone pin markers in the sagittal plane but excessive disagreement in the transverse and frontal plane during running. Errors of 21%, 63% and 70% were reported for all three planes respectively. An emerging solution to this problem involves the use of a marker array which attached to the segment (McClay and Manal, 1999; Manal et al., 2000; Manal et al., 2002). Because the markers are attached to the array they do not move independently of each other and therefore, improve measurement accuracy. Deviations of  $\pm 2^\circ$  have been shown in the frontal and sagittal plane, and  $\pm 4^\circ$  in the transverse plane (Manal et al., 2000). Marker arrays have successfully been used in previous ACL injury research (Pollard et al., 2004b; Pollard et al., 2005; Joseph et al., 2008) and they are suggested as the most optimal non-invasive method of estimating segment rotations (Angeloni et al., 1993).

**M**ovement tasks chosen to replicate high risk sporting movements are essential in producing ecologically valid experiments. Firstly these tasks need to closely align with ACL injury indirect and non-contact mechanisms (2.4 ACL Injury Mechanism). These movements include landing, one step stops, and changes of direction such as a side step or cutting movement. Unanticipated (Besier et al., 2001a; Besier et al., 2001b; Besier et al., 2003; Pollard et al., 2004a; Ford et al., 2005b; Borotikar et al., 2008; O'Connor et al., 2009), high intensity performance (Besier et al., 2003; McLean et al., 2004b; Pollard et al., 2004a; O'Connor et al., 2009) of these tasks in combination with match specific conditions such as opponents (McLean et al., 2004b) and equipment (Chaudhari et al., 2005; Ford et al., 2005a) have been shown to increase the magnitude of risky joint motions such as knee flexion and abduction. This suggests that these tasks may be more representative of high risk ACL injury movements in sports.

## **2.10 Aims of this programme of research**

ACL injury has been identified as a common injury occurring in sports that involve landing and pivoting movements. Treatment for active athletes following ACL injury, involves surgical reconstruction and rehabilitation, leading to at least six months lost from playing time. After returning to their sport, athletes are at an even greater risk of additional ACL injury and development of degenerative joint disease. Female athletes have been identified as a group with exceptionally high rates of ACL injury, in comparison to males participating in similar sports.

ACL injury risk factors for groups, such as females and ACLr individuals need further validation, utilising ecologically valid movement tasks, as described in the previous section

(2.9 Experimental Design Issues). This information will further inform preventative and rehabilitation programmes for prevention of initial and repeated ACL injury. The increased risk of ACLr individuals developing degenerative joint disease or OA also requires further investigation. The assessment of the ACLr knee during sports specific tasks in comparison to the contralateral uninjured knee and a nACL control knee may indicate factors linked with OA development. This information will further inform surgical and rehabilitation protocols, to aid in the reduction of OA development. The role of movement variability in ACL injury risk and OA development also merits further investigation as previous research highlight its potential role in injury risk and pathology.

Therefore, the overall aims of this programme of research are to:

- Compare males and females in the performance of a sports specific movement task.
  - Utilising a traditional comparison with discrete parameters.
  - Utilising a dynamic systems comparison with kinematic and coordination variability.
- Compare ACLr knees with contralateral and nACL knees in the performance of a sports specific movement task.
  - Utilising a traditional method with discrete parameters.
  - Utilising a composite measure of lower limb coordination.
  - Utilising a dynamic systems method with kinematic and coordination variability.

## 2.11 References

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## **CHAPTER 3**

### **THE ROLE OF GENDER ON JOINT KINEMATICS AND KINETICS IN UNINJURED ATHLETES DURING A MATCH SPECIFIC LAND-CUT TASK**

### 3.1 Abstract

This study compared the lower limb mechanics of male (n=10) and female (n=10) subjects. Three dimensional hip, knee and ankle joint moments and angles were calculated during the performance of a maximal drop jump land and unanticipated cutting task (n=20). Females demonstrated increased hip internal rotation angles during the landing component of the task, when compared to males. Hip internal rotation alone is unlikely to increase the risk of ACL injury, but this finding does provide support for the premise that altered proximal control may influence the incidence of ACL injuries in female athletes.

### 3.2 Introduction

Anterior Cruciate Ligament (ACL) injury continues to be a constant adversary to the athletes of landing and cutting sports such as basketball (Ireland and Wall, 1990; Gomez et al., 1996; Messina et al., 1999; Agel et al., 2005), soccer (Bjordal et al., 1997; Agel et al., 2005; Mihata et al., 2006), gaelic football (Gaelic Athletic Association Medical Player Welfare Committee, 2011) and Australian rules football (Orchard and Seward, 2002). The female athlete has been widely acknowledged as being at an increased risk of ACL injury, in comparison with males participating in the same sport (Roos et al., 1995; Bjordal et al., 1997; Myklebust et al., 1997; Myklebust et al., 1998; Arendt et al., 1999; Messina et al., 1999; De Loes et al., 2000; Olsen et al., 2004; Agel et al., 2005; Junge et al., 2006; Mihata et al., 2006; Prodromos et al., 2007; Borowski et al., 2008; Flood and Harrison, 2009; Randazzo et al., 2010; Pappas et al., 2011).

Although the exact mechanism for indirect and non-contact ACL injury is unknown most authors agree that it occurs in movements that involve rapid deceleration and changes of direction when the athlete's attention is focused on the task at hand (Hutchinson and Ireland, 1995; Myklebust et al., 1997; Myklebust et al., 2003; McLean et al., 2004; Chaudhari et al., 2005; Cochrane et al., 2007; Fedie et al., 2010; Hashemi et al., 2011). Joint loading and associated movement patterns of the lower limbs have been shown to differ by gender during these movement tasks.

There is a plethora of previous investigations comparing lower limb biomechanics of male and female athletes during landing tasks (Huston et al., 2001; Lephart et al., 2002; Decker et al., 2003; Ford et al., 2003; Hewett et al., 2004; Salci et al., 2004; Kernozek et al., 2005; Ford et al., 2006; Russell et al., 2006; McLean et al., 2007; Nagano et al., 2007; Pappas et al., 2007; Hughes et al., 2008; Beutler et al., 2009; Brown et al., 2009; Gehring et al., 2009; Kiriya et al., 2009; Schmitz et al., 2009; Ford et al., 2010). Fewer investigations are published utilising a cutting task (James et al., 2004; McLean et al., 2004; Pollard et al., 2004a; Pollard et al., 2004b; McLean et al., 2005b; Sigward and Powers, 2006; Landry et al., 2007; Beaulieu et al., 2009; Fedie et al., 2010). These investigations utilising a cutting task, were anticipated (James et al., 2004; McLean et al., 2004; Sigward and Powers, 2006; Pollard et al., 2007; Fedie et al., 2010),

and unanticipated in nature (Pollard et al., 2004a; McLean et al., 2005b; Landry et al., 2007; Beaulieu et al., 2009), and conducted on the dominant leg. The addition of alternative tasks such as a run through, stop jump etc. was used to make cutting tasks unanticipated, while still confining the analysis to the dominant leg.

Landing movements rarely occur in isolation during match situations, therefore, some investigations have utilised tasks combining both landing and cutting movements. McLean et al., (2005b) utilised an anticipated side-jump task which involved running towards a force plate, initiating a jump from 2 m before the force plate and upon landing completed a jump to the left (with the right leg) in a direction perpendicular to the initial running direction. Females were reported to have greater peak knee abduction angle and greater knee abduction angle at initial contact during this side-jump task. Brown et. al., (2009) utilised unanticipated and anticipated forward jump and cut maneuverers which involved jumping forward from a standing position 2 m behind the force plate and upon landing cut aggressively to either the right or left. Brown et al.'s, (2009) use of an unanticipated task and analysis of both dominant and non-dominant legs was an advancement on the previous work of McLean et al., (2005b). At initial ground contact females demonstrated greater hip flexion, adduction and internal rotation and knee abduction angles (Brown et al., 2009). During the initial landing period females demonstrated greater hip internal rotation angles, smaller knee internal rotation angles and smaller hip flexion and abduction external moments (Brown et al., 2009). No significant dominance related differences were reported for females between legs. Although these studies are interesting advancements on the isolated tasks of landing and cutting, no investigation has studied landing from a jump followed by cutting to either the right or left, as would happen in a game after landing from catching a ball when the player would then try to evade a tackle.

### *3.2.1 Aims and significance of the study*

The aim of this investigation was to compare the lower limb mechanics of male and female athletes during the performance of a maximal drop-jump land and unanticipated cutting task. In line with previous developments in movement task design, the task was unanticipated by means of a random cutting direction. 3D kinematics and kinetics of the hip, knee and ankle joints were measured for both legs during the land, and for the push off leg during cutting. This will allow for comparison between genders for both dominant and non-dominant legs during the landing and cutting components of the task. The assessment of lower limb mechanics during this novel drop-jump land and cut task will provide new information on the differences in lower limb joint mechanics between genders during high risk movement tasks.

Based on previous research on ACL loading (Wascher et al., 1993; Markolf et al., 1995) and gender comparisons (Section 2.6.4), it was hypothesised that the gender differences in Table



3-1 would be present for joint kinematics and kinetics in the sagittal, frontal and transverse plane:

Table 3-1 Hypothesised gender differences for the present investigation

Hypothesised female joint mechanics			Landing	Cutting
Sagittal Plane	Angles	Hip: Knee: Ankle:	< Flexion < Flexion > Peak Dorsi-flexion > Flexion ROM > Plantar Flexion @ Touchdown	< Flexion < Flexion <b>NO DIFFERENCE</b>
	Internal Moments	Hip: Knee: Ankle:	> Extension < Flexion > Plantar Flexion	> Extension > Extension <b>NO DIFFERENCE</b>
Frontal Plane	Angles	Hip: Knee:	> Adduction > Abduction	<b>NO DIFFERENCE</b> > Abduction
	Internal Moments	Hip: Knee:	< Abduction < Adduction	< Abduction < Adduction
Transverse Plane	Angles	Hip: Knee: Ankle:	> Internal Rotation > Internal Rotation > Eversion	> Internal Rotation > Internal Rotation > Eversion
	Internal Moments	Hip: Knee: Ankle:	< External Rotation < External Rotation < Inversion	< External Rotation < External Rotation < Inversion

### 3.3 Methods

#### 3.3.1 Subjects

Ten males (height  $1.80 \pm 0.09$  m, mass  $79.2 \pm 6.4$  kg) and ten females (height  $1.67 \pm 0.06$  m, weight  $64.7 \pm 6.4$  kg) between the ages of 18 and 26 years (males  $22.4 \pm 2.7$ , females  $21.8 \pm 2.3$ ), were recruited for participation in this investigation. All subjects played at a standard of intermediate/senior club, intercollegiate, inter-county or above and were categorized as highly experienced in their sport. Approval for the participation of human subjects in this investigation was granted by the University of Limerick Research Ethics Committee. Subjects had no history of serious lower limb injury and were injury free for a period of six months prior to testing.

### 3.3.2 *Experimental set-up*

Kinetic and kinematic data were collected during a dynamic drop-jump land and cut task. Kinetic data were recorded using two AMTI force platforms (one leg to each platform) sampling at 1000 Hz. Kinematic data was recorded via 6 Eagle infrared Motion Analysis Corporation cameras sampling at 500 Hz. Kinetic and kinematic data were simultaneously recorded through Cortex software (Motion Analysis Corporation, v5.0, Santa Rosa, CA).

The cameras were positioned so that each marker was visible by at least 2 cameras throughout the drop-jump, landing and cutting movements. A right-handed lab coordinate system was defined using a rigid L-frame containing four markers of known locations. A three-marker wand of known length was used within the calibration volume to scale the individual camera views.

The maximal drop-jump land and unanticipated cutting task assessment was designed to replicate high demand match situations in a lab environment. A pre injury perturbation, via the introduction of a static defender in cutting manoeuvres (McLean et al., 2004) or using an overhead goal (Ford et al., 2005a) in vertical jump, has been shown to significantly alter the lower limb biomechanics. Unanticipated manoeuvres have also been suggested to more accurately replicate a true game-like situations (Besier et al., 2001a; Besier et al., 2003). The task designed for the current investigation involves dropping from a 30 cm bench, and performing an immediate drop-jump up to reach and touch a target which is suspended at a previously recorded maximum drop-jump height. The 30 cm bench height is similar to drop heights used in previous investigations (Huston et al., 2001). The suspended target acts as a trigger for a directional cueing system which indicates to the subject on landing which direction they must run to. The run/cutting manoeuvre is completed on a pathway orientated at a 45° angle in either the right or left direction. Angle of cut was required to be greater than 30° as little difference in joint kinetics have been demonstrated between 0 and 30° (Kaila, 2007). It has been shown that ACL injuries are most likely to occur between cutting ranges of 15-45° and 45-75° (Cochrane et al., 2007), several previous investigations have utilised cutting angles of 45° (Pollard et al., 2004b; Pollard et al., 2005; Sigward et al., 2007; O'Connor and Bottum, 2009) See Figure 3-1 for a pictorial representation of the set-up and Appendix A1 for a video recording of the maximal drop-jump land and unanticipated cutting task.

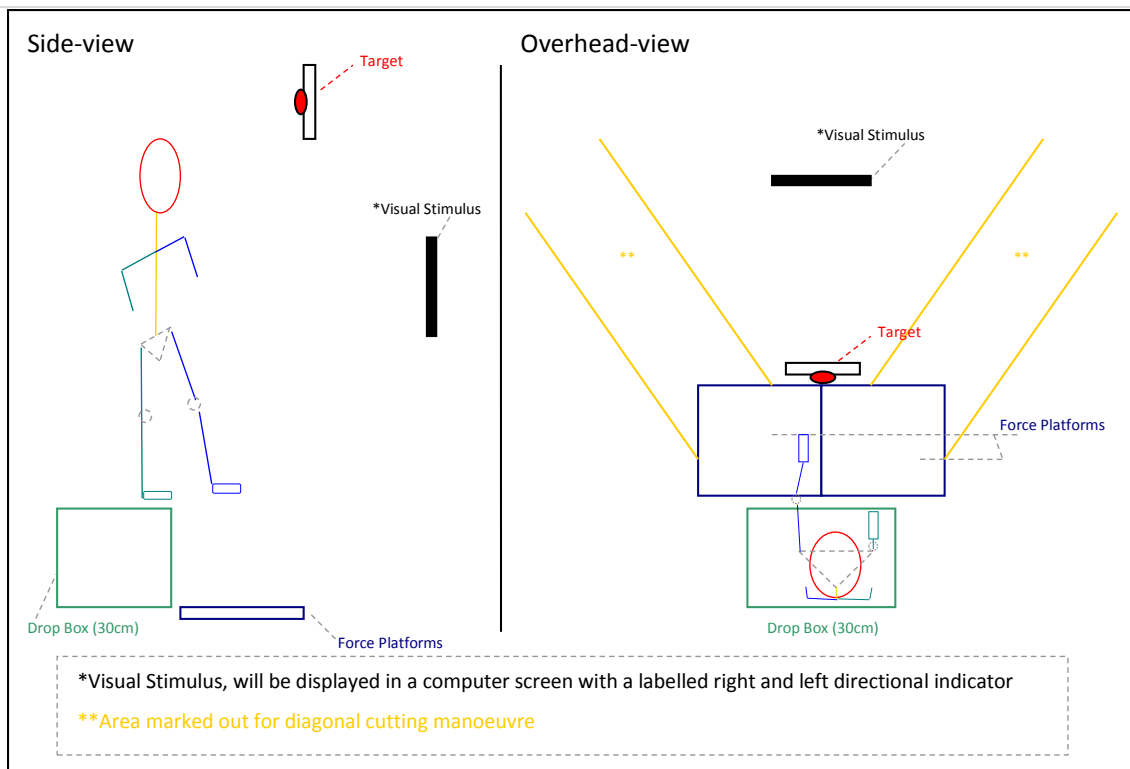
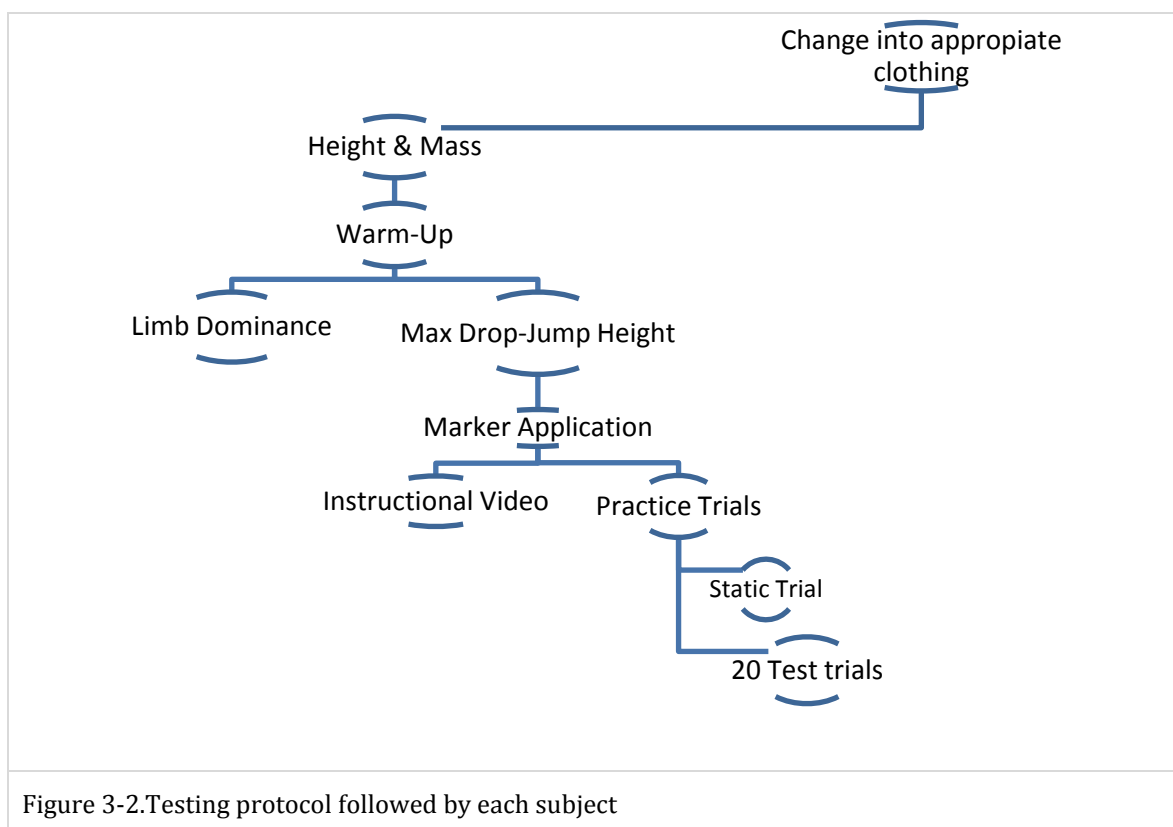


Figure 3-1. Illustration of the Drop-Jump-Land and Cut Task Set-Up

### 3.3.5 Protocol

Prior to initiation of the testing protocol each subject completed an informed consent form, a pre-test questionnaire to screen for any potential health risks and an injury history and sporting experience questionnaire to ensure all subjects were injury free and adequately experienced in their sport. See Appendix A3 for copies of all above documentation.

Testing protocol was completed as shown in Figure 3-2.

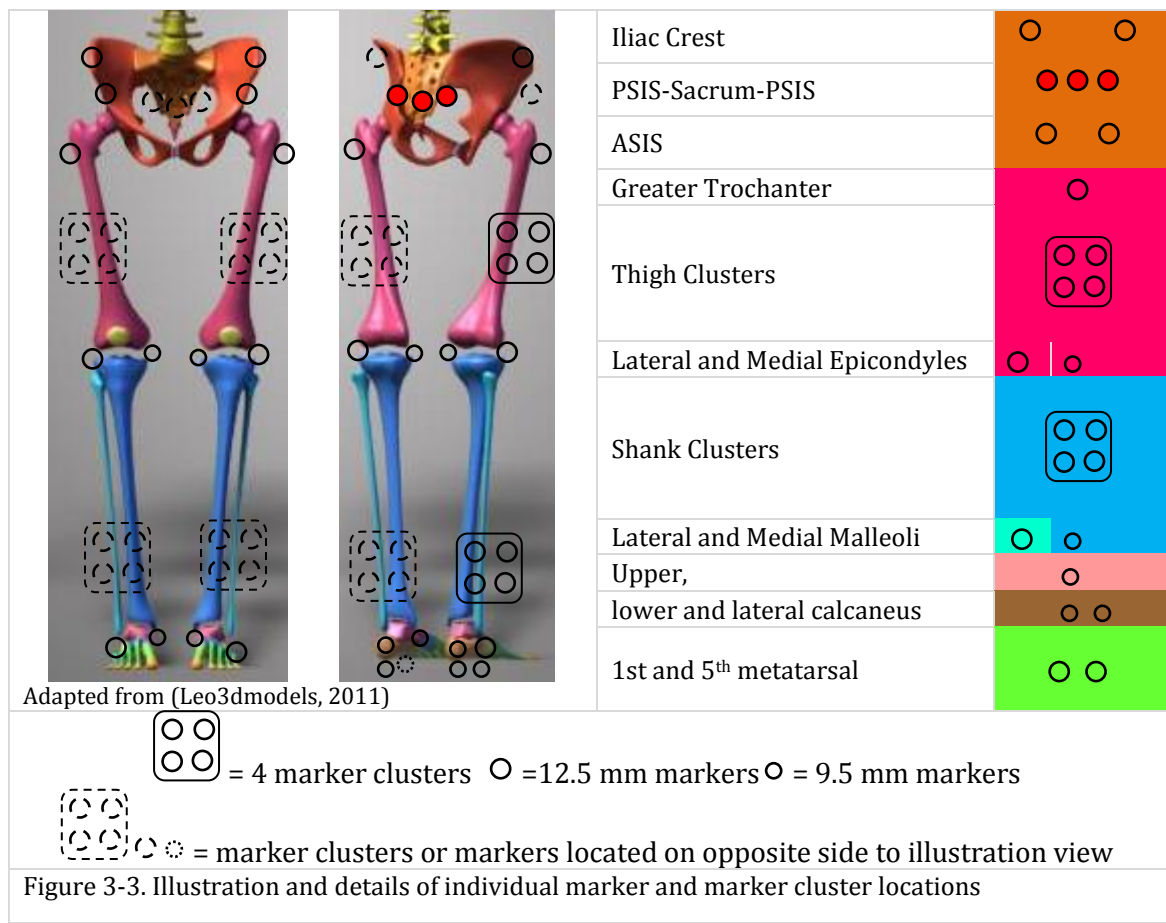


Appropriate clothing consisted of their own running shoes (those which they used for indoor training), tight fitting top and high-cut running shorts. Following this height and mass were measured. The warm-up consisted of a ten minute jog followed by light stretching.

Limb dominance was assessed using three independent tests. The leg used to kick a ball as far as possible (Wojtys et al., 1996; Simoneau et al., 1998; Ford et al., 2003; Atnip and McCrory, 2004; Ford et al., 2005a; McLean et al., 2007; Brown et al., 2009; Nagano et al., 2009), the leg used to land from a jump on one leg and the leg used to regain balance when push slightly off balance from behind (Hoffman et al., 1998). Golden et al., (2009) has utilised three similar tests for assessing limb dominance, the inclusion of three separate assessments allows for a conclusive decision, with the limb utilised in at least two of the three tasks chosen as the dominant limb. Maximum drop-jump height was assessed by a chalk mark imprinted on a wall from the subject's chalked palm following a maximum drop-jump from a 30 cm bench and was used in setting the demands of the dynamic task.

A total of 45 reflective markers, including four rigid four-marker clusters, and the remaining individual calibration and tracking markers were placed on each subject as shown in Figure 3-3. The rigid marker clusters were placed on the thigh and shank, and marker trios on the pelvis (left PSIS, sacrum and right PSIS) and calcaneus (upper, lower and lateral calcaneus), are used to define segment rotations. The remaining markers define joint centres (greater trochanters, epicondyles and malleoli) and act as tracking markers. The same individual

placed markers on all subjects. Similar marker sets have been previously utilised for side cutting manoeuvres (Pollard et al., 2004b; Pollard et al., 2005; Joseph et al., 2008) and are suggested as the most optimal non-invasive method of estimating segment rotations (Angeloni et al., 1993)



All subjects viewed an instructional video which consisted of key instructions and a demonstration of the task (See Appendix A1). Following this they completed a number of practice trials until they felt comfortable completing the task.

Following a static trial subjects completed a minimum of 20 trials of the dynamic task with a random cutting direction. Ten successful trials in each direction were required from each subject. Successful trials required the subject to run in the correct cutting direction as directed by the visual cue, through the mapped out pathway orientated at 45°. Both feet were also required to land on their respective force plates during the jump land as was the appropriate cutting leg (e.g. the right leg when cutting in left direction) during the cut. Subjects received 1 minute rest between trials (Pollard et al., 2004a; Pollard et al., 2005; Sigward and Powers, 2006) to prevent the potential effects of fatigue.

### 3.3.8 Data reduction

The following data reduction procedures were performed on data for all subjects. Cortex (Motion Analysis Corporation, Santa Rosa, CA, USA) was used to track and export raw 3D

marker coordinate data. The raw coordinate and ground reaction force data were low-pass filtered with a fourth-order Butterworth filter with a 12 Hz and 50 Hz cut off frequency respectively. These cut off frequencies were selected based on similar values utilised in previous investigations utilising similar movement tasks (Besier et al., 2001b; Malinzak et al., 2001; Ferber et al., 2003). The thigh, shank and foot segments were modeled as an assembly of cones, and the pelvis was modeled as a cylinder in Visual 3D™ (C-Motion, Rockville, MD, USA) as illustrated in Figure 3-4 (B). The local coordinate system and joint centers of these segments were defined from the static trial.

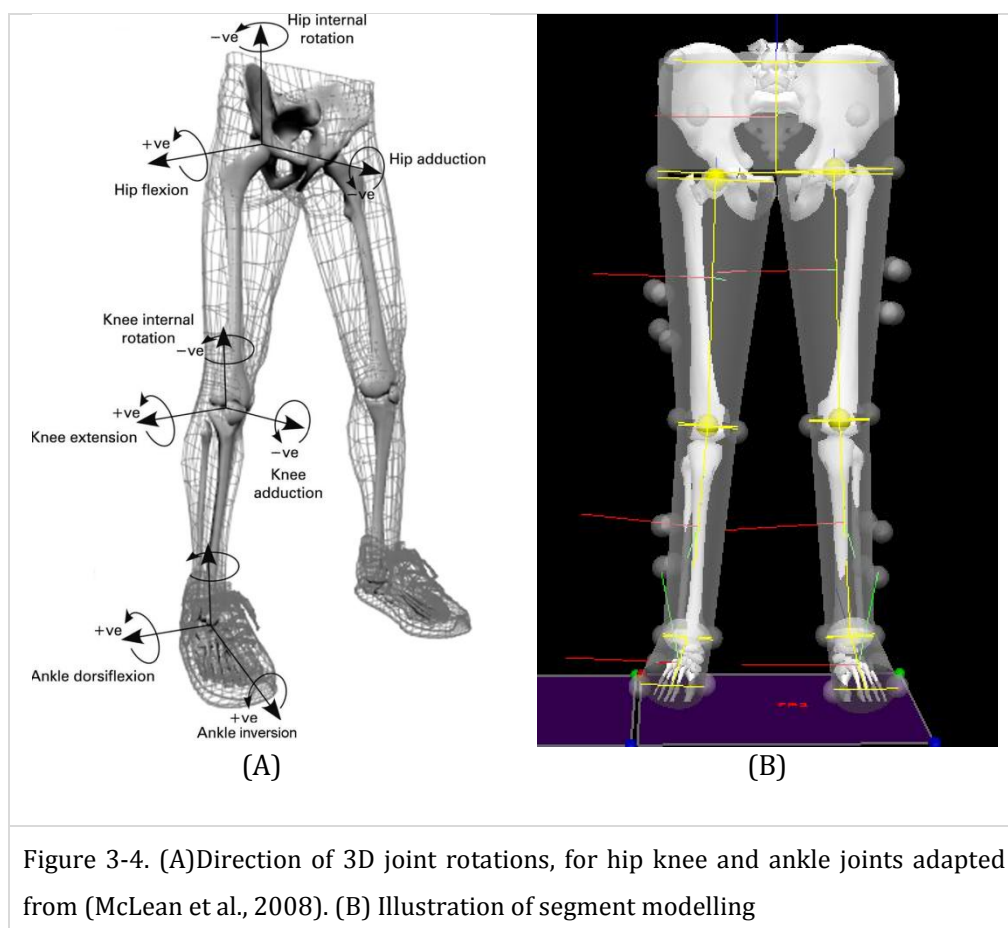


Figure 3-4. (A) Direction of 3D joint rotations, for hip knee and ankle joints adapted from (McLean et al., 2008). (B) Illustration of segment modelling

Right-handed cartesian local coordinate systems for the pelvis, thigh, shank and foot segments of the left leg were defined to describe position and orientation of each segment as illustrated in Figure 3-4 (A), this was mirrored in the frontal and transverse planes for the right leg to ensure consistent identification of anatomical movements (e.g. abduction and internal rotation) for both legs. Three-dimensional ankle, knee, and hip angles were calculated using a joint coordinate system approach (Grood and Suntay, 1983). Joint centres were denoted by the midpoint between the medial and lateral calibration markers for the knee and ankle joints (Grood et al., 1984; Wu et al., 2002) and one quarter the distance between the greater trochanter markers in the medial direction (Weinhandl and O'Connor, 2010) for the hip. Body segment parameters were estimated from (Dempster, 1955), and joint moments were represented in the joint coordinate system and resolved relative to the

distal segment reference frame (Schache and Baker, 2007). The joint moments referred to in this investigation are the internal resultant moments, similar to previous investigations (Decker et al., 2003; Madigan and Pidcoe, 2003; Salci et al., 2004; Kernozek et al., 2005; Pollard et al., 2007) normalised to body mass and height (Nm/kg.m). For example, an internal knee extension moment will tend to resist an external knee flexion load. Initial ground contact and take off were determined by the instant when the vertical ground reaction force exceeded or fell below 10 N respectively. The end of the jump landing phase and the beginning of the cutting movement were defined specific to each trial by the pattern of the vertical ground reaction force on the cutting leg. The landing and cutting components of the task were separated. The land was defined from touchdown to the end of landing, and the cut from the start of cutting to take off on the push off leg. These two components were cropped and time normalized to 1001 data points.

Various discrete measures calculated during the different phases of the task (i.e. the first 40 ms of the land, the entire landing stance phase, the entire cutting stance phase). Touchdown (TD), peak angles, peak angular excursion range of motion (ROMmax<sup>13</sup>) and peak moments of the hip, knee and ankle<sup>14</sup> were reported in all planes during the entire landing stance phase. Peak angles, angular range of motion (ROM<sup>15</sup>), and peak moments of the hip, knee and ankle were reported in all planes during the first 40 ms of the land and the entire cutting stance phase. ACL injury is reported to occur during the first 40 ms of the landing phase (Krosshaug et al., 2007; Koga et al., 2010) hence it was utilised for additional analysis.

### 3.3.9 Statistical Analysis

Differences between genders for the each of the above discrete measures for hip, knee and ankle joint angles and moments were analysed using a repeated measures ANOVA, with gender and trial as a between and within subject factors respectively. These methods were utilised for the data from first 40 ms of the land, the entire land stance phase and the entire cut stance phase. Significance for all tests was set at  $p < .05$ , except in cases where a moderate - large effect size was present with a p-value  $< 0.1$ , in such cases significance was set at  $p < 0.1$ . All statistical analyses were performed using SPSS (PASW v18.0, IBM Inc. Armonk, NY). Partial eta<sup>2</sup> ( $\eta^2$ ) was also reported as a measure of effect size (E.S.). It was calculated using the formula:  $\eta^2 = SS_{\text{effect}} / (SS_{\text{effect}} + SS_{\text{error}})$ , where  $SS_{\text{effect}}$  = effect variance and  $SS_{\text{error}}$  = error variance. Interpretation of effect size was based on the scale for effect size classification of Hopkins (2000). This scale is based on  $f$ -values for effect size and these were converted to  $\eta^2$  using the formula:  $f = (\eta^2 / (1 - \eta^2))^{0.5}$ . Consequently, the scale for classification of  $\eta^2$  was

<sup>13</sup> ROMmax was calculated during landing from touchdown to peak angular excursion

<sup>14</sup> Angles and moments at the ankle joint were limited to the sagittal and transverse plane

<sup>15</sup> ROM was calculated by subtracting the maximum and minimum values

< 0.04 = trivial, 0.041 to 0.249 = small, 0.25 to 0.549 = medium, 0.55 to 0.799 = large, and >0.8 = very large.

### 3.4 Results

#### 3.4.1 *Jump Landing*

Table 3-2 presents a summary of the significant differences present at the hip and knee joint in the transverse plane joint angles during landing. It is evident that individuals within each gender grouping demonstrate different touchdown and peak hip and knee transverse plane kinematics during the first 40 ms and the entire landing stance phase. Therefore, individual data for variables with a red asterisk (\*) are plotted as bar graphs in Figure 3-5 to further explore the results.

No gender differences in sagittal or frontal plane joint angles were found for the hip, knee or ankle joint during landing. Males and females both landed with knee flexion angles which were on average > 20° (Males  $-21 \pm 5^\circ$ , Females  $-23 \pm 4^\circ$ ) at touchdown. Males and females also landed with similar touchdown and peak hip flexion angles. There were also no gender differences in touchdown or peak hip adduction or knee abduction. The majority of gender differences were seen in the transverse plane. Females demonstrated increased hip internal rotation at touchdown and smaller minimum external-internal rotation during the initial 40 ms of the land. This was reported in the dominant leg during both the initial 40 ms and the entire landing stance. Females were also shown to have decreased peak external rotation at the hip on the dominant leg, when compared to males during both the initial 40 ms and the entire landing stance respectively. It was hypothesized that females would demonstrate increased internal rotation at the knee; this was not the case with the male subjects demonstrating smaller minimum external-internal rotation at the knee during the stance phase of the land on the non-dominant leg. Males also demonstrated increased transverse plane ROM at the knee in both legs during the first 40 ms of the land. There were no gender differences in transverse plane ankle angle variables during landing. Figure 3-6 presents the 3D joint angles over the entire stance phase of landing for the hip and knee angles that showed significant gender differences.

The knee rotation ROM values for individual males and females are presented in Figure 3-7 to identify the orientation (internal or external rotation) and magnitude of ROM for each subject.



Table 3-2. Significant gender differences in 3D Joint Angles of Hip, Ankle and Knee during Landing.

Average angles (°) for both legs of males (♂) and females (♀) are shown. Group differences (°), Partial eta2 ( $\eta^2$ ) and p-values are also presented. The scale for classification of  $\eta^2$  was < 0.04 = trivial, 0.041 to 0.249 = small, 0.25 to 0.549 = medium, 0.55 to 0.799 = large, and > 0.8 = very large.

Jump Land		Dominant Leg					Non-Dominant Leg				
1st 40 ms		♂	♀	Diff	$\eta^2$	p-value	♂	♀	Diff	$\eta^2$	p-value
Hip	Max Ext-Internal Rotation*	4.66°	-1.95°	6.61°	0.267	0.02	NON-SIGNIFICANT				
	Min Ext-Internal Rotation*	0.50°	-6.89°	7.39°	0.268	0.019					
		<i>Subjects with maximum angles &lt;0, females n=10 (average 12 trials/subject, at -4.63°), males n=7 (average 8 trials/subject, at -5.16°)</i> <i>Subjects with minimum angles &gt;0, females n=6 (average 4 trials/subject, at 2.66°), males n=8 (average 8 trials/subject, at 5.74°)</i>									
Knee	Ext-Internal Rotation ROM	4.37°	2.36°	2.01°	0.199	0.049	4.39°	2.46°	1.93°	0.307	0.011
Stance Phase		♂	♀	Diff (°)	$\eta^2$	p-value	♂	♀	Diff	$\eta^2$	p-value
Hip	Touchdown External-Internal Rotation*	4.29	-2.33	6.62	0.264	0.021	NON-SIGNIFICANT				
	Max Ext-Int Rotation*	6.52°	-0.18°	6.70°	0.281	0.016					
	Max Int Rotation	-5.81°	-12.63°	6.82°	0.311	0.011					
		<i>Subjects with touchdown angles &lt;0, females n=10 (average 13 trials/subject, at -4.96°), males n=7 (average 9 trials/subject, at -3.02°)</i> <i>Subjects with maximum angles &lt;0, females n=10 (average 10 trials/subject, at -3.66°), males n=7 (average 6 trials/subject, at -0.92°)</i>									
Knee	Min Ext-Int Rotation*	NON-SIGNIFICANT					-0.142°	5.53°	5.39°	0.207	0.044
							<i>Subjects with minimum angles &lt;0, females n=2 (average 3 trials/subject, at -1.09°), males n=7 (average 14 trials/subject, at -5.3°)</i>				

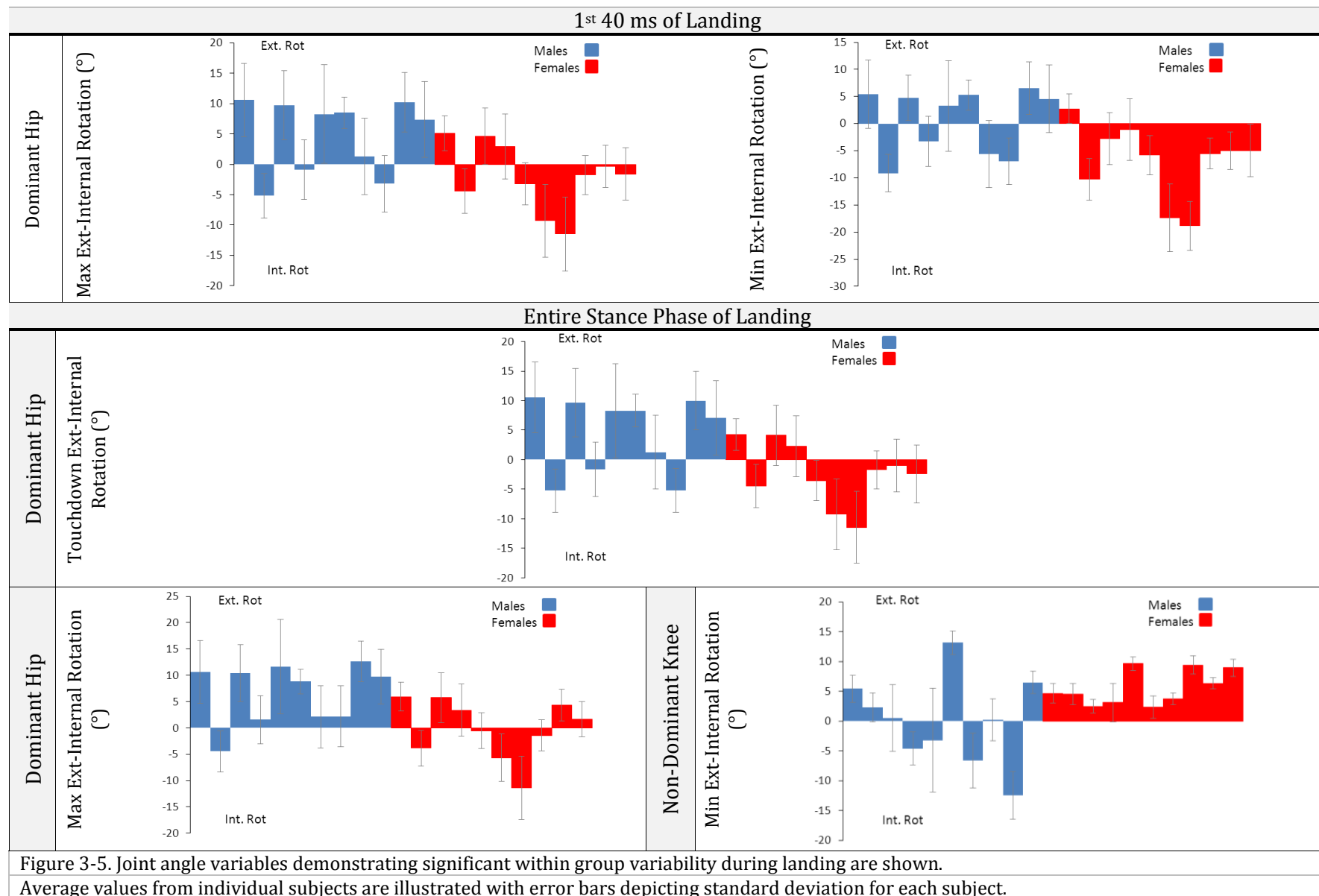


Figure 3-5. Joint angle variables demonstrating significant within group variability during landing are shown.

Average values from individual subjects are illustrated with error bars depicting standard deviation for each subject.

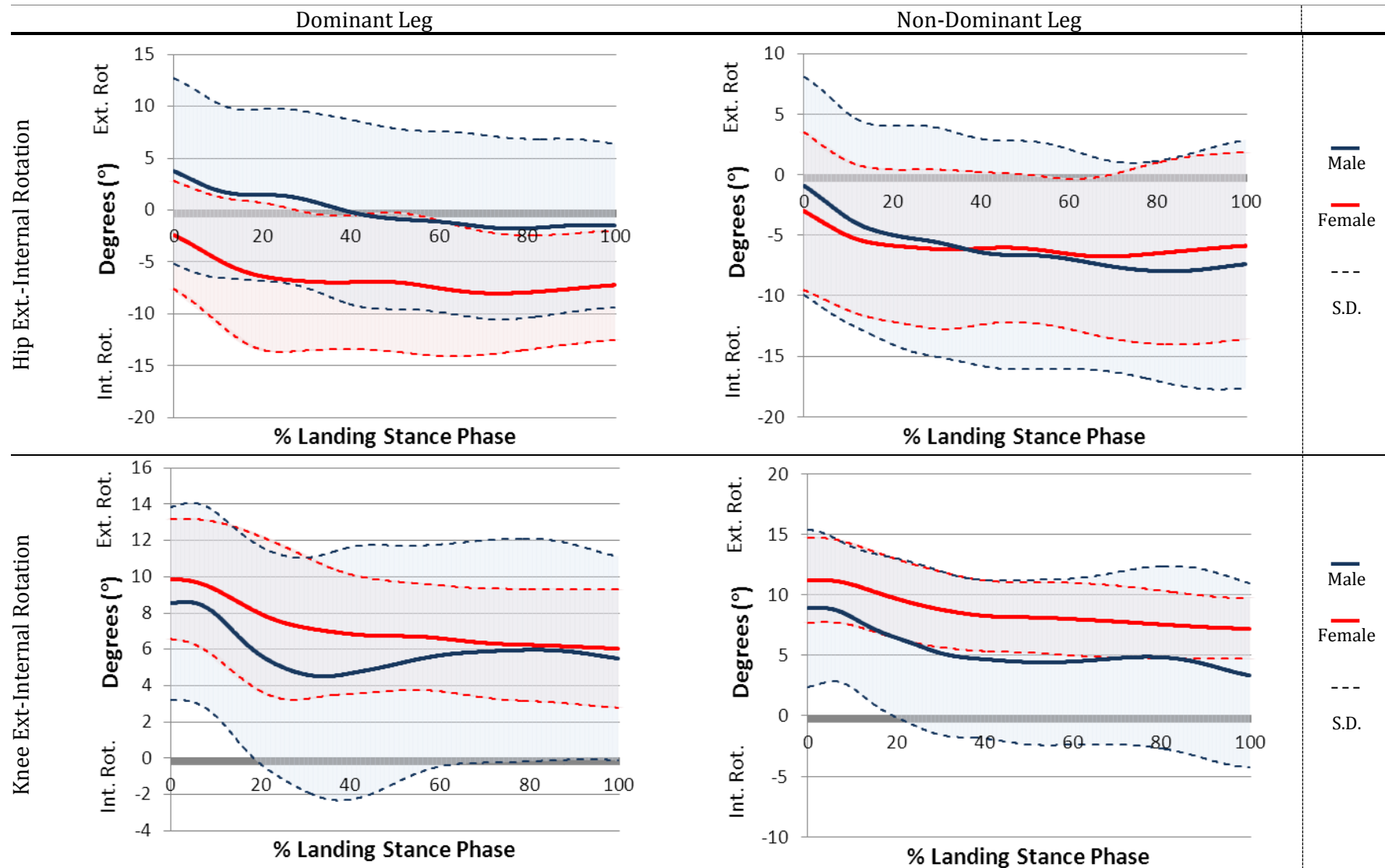
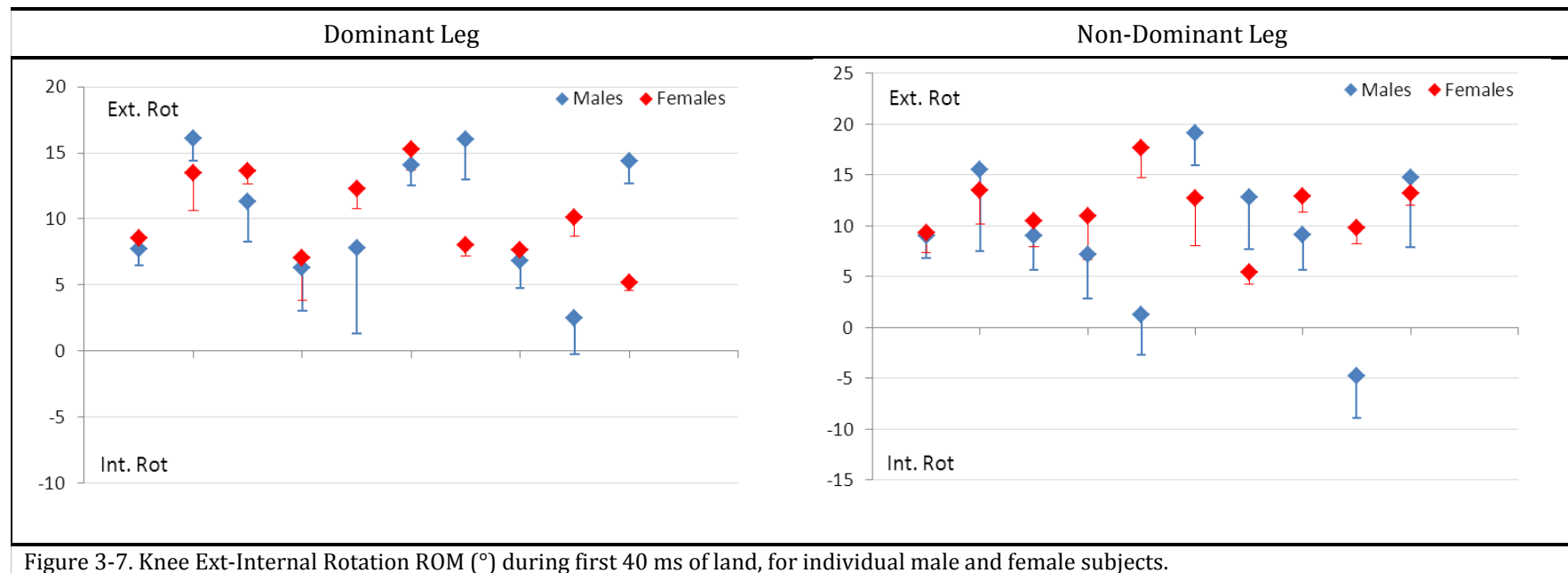


Figure 3-6. 3D joint angles of hip and knee with significant gender differences during landing.



As a post analysis, transverse rotation velocities at the hip were reported in Table 3-3 to allow evaluation alongside the work of McLean et al (2008).

Table 3-3. Average Hip Ext-Internal Rotation Angular Velocity ( $^{\circ}/\text{sec}$ ) during the first 40 ms of the land.

Average angular velocities ( $^{\circ}/\text{sec}$ ), for both legs of males ( $\sigma$ ) and females ( $\varphi$ ) are shown for the first 40 ms of the land. Group differences ( $^{\circ}/\text{sec}$ ), partial  $\eta^2$  ( $\eta p^2$ ) and p-values are also presented. The scale for classification of  $\eta p^2$  was  $< 0.04$  = trivial, 0.041 to 0.249 = small, 0.25 to 0.549 = medium, 0.55 to 0.799 = large, and  $> 0.8$  = very large.

Dominant Leg					Non-Dominant Leg				
$\sigma$	$\varphi$	Diff	$\eta p^2$	p-value	$\sigma$	$\varphi$	Diff	$\eta p^2$	p-value
36.82	<b>50.14</b>	13.32	<b>0.046</b>	0.363	<b>56.57</b>	37.33	19.24	<b>0.096</b>	0.183

Table 3-4 presents a summary of the significant differences present at the hip, knee and ankle joint in the 3D joint moments during landing.

Table 3-4. Significant gender differences in Peak 3D Normalised Internal Joint Moments of Hip, Ankle and Knee during Landing.

Average peak moments (Nm/kg.m), for both legs of males ( $\sigma$ ) and females ( $\varphi$ ) during the first 40 ms and entire stance phase are shown. Group differences (Nm/kg.m), partial  $\eta^2$  ( $\eta p^2$ ) and p-values are also presented. The scale for classification of p- $\eta p^2$  was  $< 0.04$  = trivial, 0.041 to 0.249 = small, 0.25 to 0.549 = medium, 0.55 to 0.799 = large, and  $> 0.8$  = very large.

Jump Land		Dominant Leg				
1st 40 ms		$\sigma$	$\varphi$	Diff	$\eta p^2$	p-value
Hip	Max Adduction	<b>-0.358</b>	-0.210	0.134	<b>0.275</b>	0.018
Knee	Max Flexion	<b>-0.43</b>	-0.30	0.13	<b>0.318</b>	0.010
Ankle	Max Plantar Flexion	<b>-0.811</b>	-0.573	0.24	<b>0.241</b>	0.028
Stance Phase		$\sigma$	$\varphi$	Diff ( $^{\circ}$ )	$\eta p^2$	p-value
Hip	Max Adduction	<b>-0.421</b>	-0.233	0.19	<b>0.334</b>	0.008
Knee	Max Flexion	<b>-0.434</b>	-0.309	0.125	<b>0.31</b>	0.011
Ankle	TD Dorsi-Plantar Flexion	<b>-0.012</b>	-0.002	-0.010	<b>0.199</b>	0.049
	Max Dorsi-flexion	-1.34	<b>-0.992</b>	-0.349	<b>0.214</b>	0.040

All gender differences in 3D normalized internal joint moments were shown on the dominant leg. No gender differences were found for hip extension moments. Females were shown to have decreased knee flexion moment during the initial 40 ms and entire stance period of the land. At the ankle, females had decreased plantar-flexion moment during the first 40 ms of the land. Increased dorsi-flexion moment was also shown at the ankle in females when

compared to males during the entire land. In the frontal plane females had greater hip adduction moment during the first 40 ms and entire landing stance period. In the transverse plane, all differences were seen at the hip. Females had greater hip internal rotation moment at touchdown. Figure 3-8 presents the entire landing phase of 3D hip, knee and ankle joint moments which had significant gender differences.

## Dominant Leg

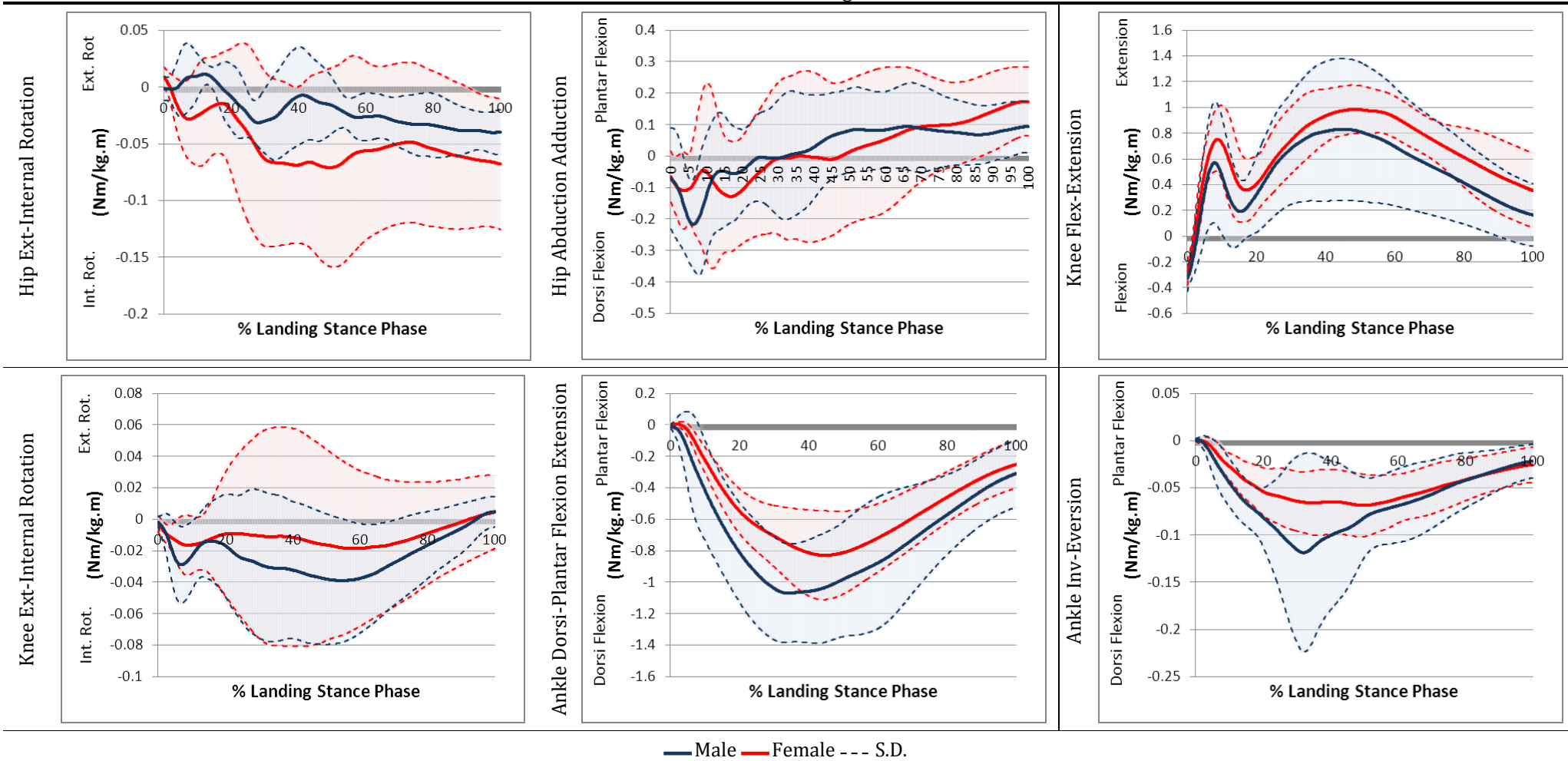


Figure 3-8. 3D normalised internal joint moments at the hip knee and ankle joints with significant genders differences during landing.

### 3.4.2 Cutting Manoeuvre

Table 3-5 presents a summary of the significant differences present at the knee and ankle joint in the 3D joint angles during cutting.

Table 3-5. Significant gender differences in 3D Joint Angles of Hip, Ankle and Knee during cutting.											
Average angles (°), for both legs of males (♂) and females (♀) during the first 40 ms and entire stance phase are shown. Group differences (°), partial eta2 (ηp2) and p-values are also presented. The scale for classification of ηp2 was < 0.04 = trivial, 0.041 to 0.249 = small, 0.25 to 0.549 = medium, 0.55 to 0.799 = large, and >0.8 = very large.											
Cut		Dominant Leg					Non-Dominant Leg				
		♂	♀	Diff	ηp²	p-value	♂	♀	Diff	ηp²	p-value
Knee	Max Abd-Adduction*	NON-SIGNIFICANT					-0.94°	3.20°	4.07°	0.234	0.031
	Abduction Adduction ROM	7.14	9.53	2.39	0.217	0.039	NON-SIGNIFICANT				
Ankle	Min Dorsi-flexion	61.62°	47.48°	14.14	0.479	0.001	NON-SIGNIFICANT				
	Dorsi-Flexion ROM	37.01	48.31	11.30	0.428	0.002	NON-SIGNIFICANT				

Again it is evident that individuals within each gender grouping are demonstrating varied maximum knee abduction angles during the cutting stance phase. Therefore, individual data for this variable (\*) are plotted as bar graphs in Figure 3-9 to further explore the results.



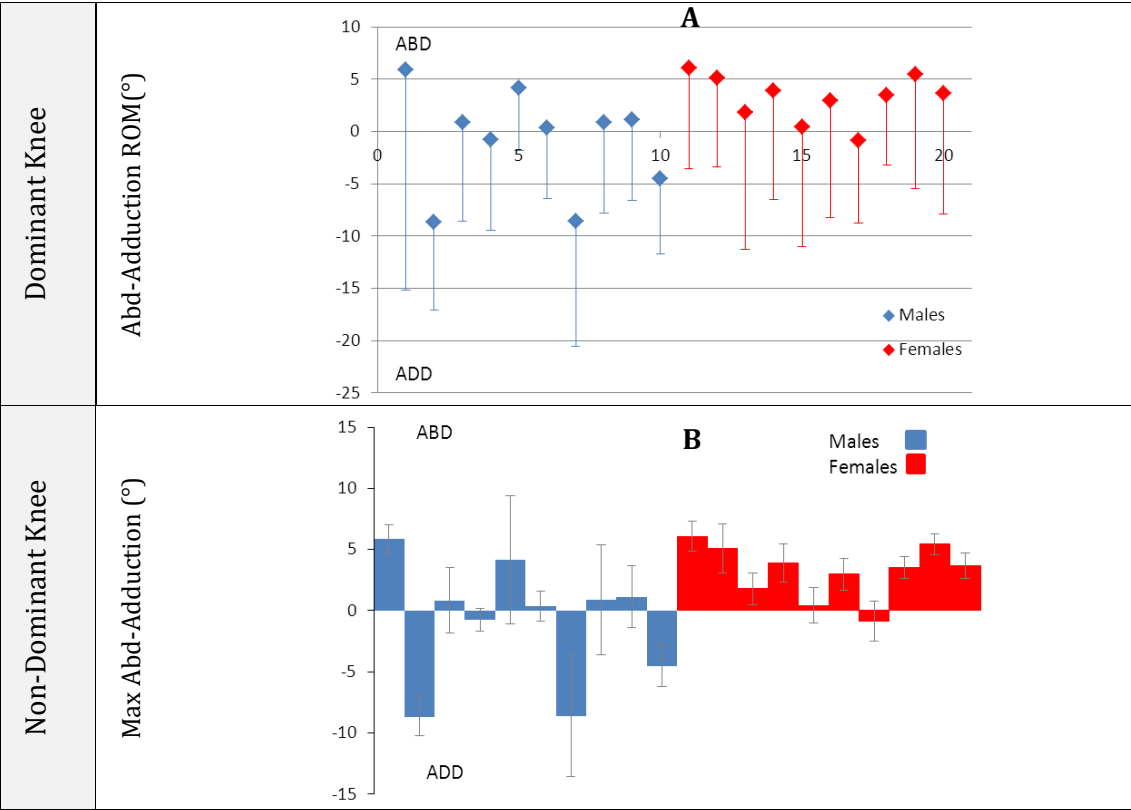


Figure 3-9. A) Knee Abd-Adduction ROM during the cut, for individual male and female subjects. B) Maximum Abd-Adduction values for each subjects during cutting are shown. Average values from individual subjects are illustrated with error bars depicting S.D. for each subject.

There were no gender differences in hip or knee flexion variables. At the ankle females had smaller minimum dorsi-flexion in the non-dominant leg and larger dorsi-flexion ROM on the non-dominant leg. There was no gender difference in hip adduction angles, and females had increased maximum knee abduction in the non-dominant leg as hypothesized. There were no gender differences in transverse plane joint angles at the hip, knee or ankle during cutting. Figure 3-10 show the entire cutting stance phase of the 3D hip, knee and ankle joint angles which had significant gender differences.

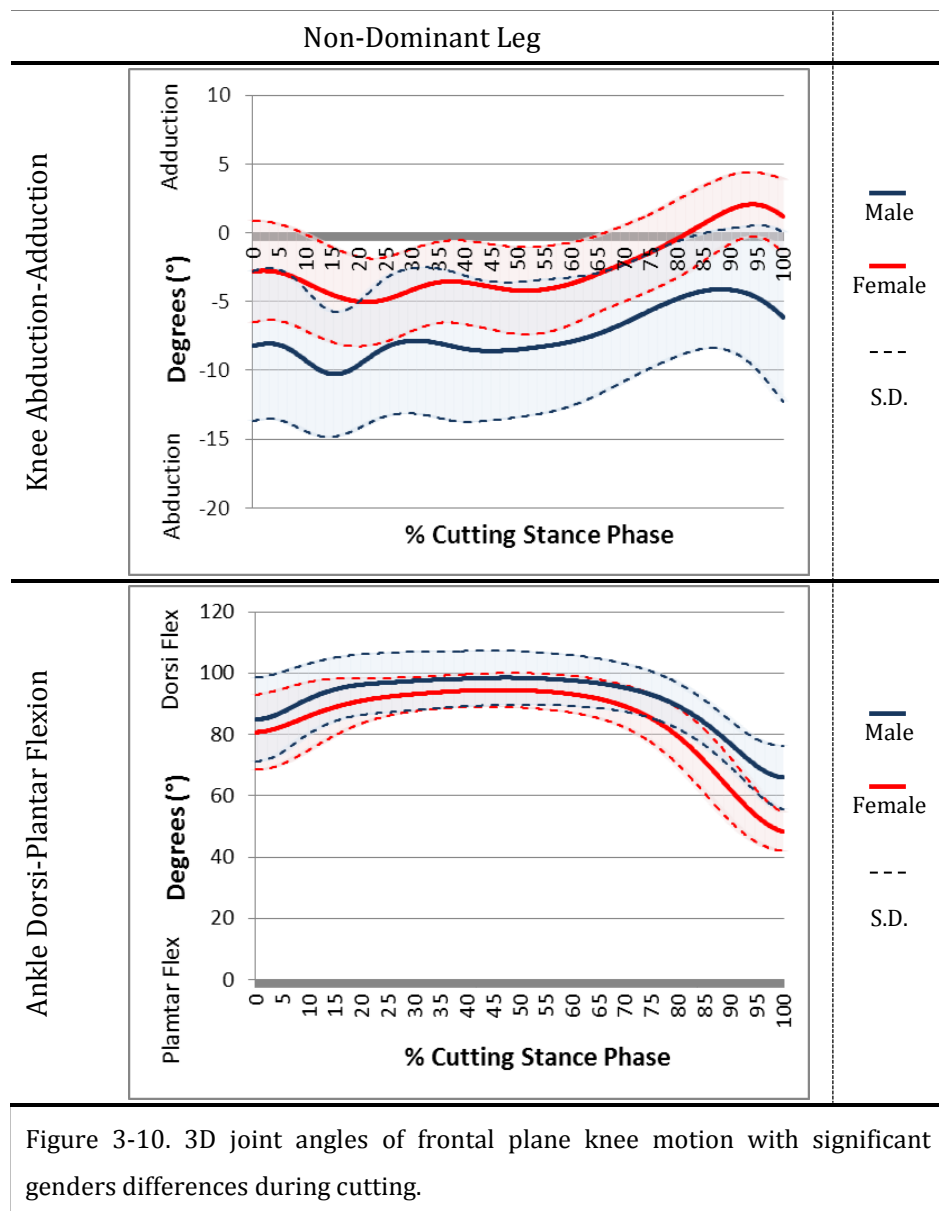


Table 3-6 shows the significant differences present in the hip, knee and ankle joint moments in all planes during the cutting component of the movement. At the hip there were no gender differences in extension moment, but females did have less flexion moment. At the ankle females demonstrated smaller maximum dorsi-flexion moments on both legs. Females had increased hip and knee abduction moments in the dominant leg. Females were hypothesized to have smaller hip and knee external rotation moments; females had larger internal rotation moments at the hip and larger external rotation moments at the knee. No gender differences were seen in the transverse plane moments at the ankle. Figure 3-11 and Figure 3-12 show the entire cutting stance phase of the 3D hip, knee and ankle joint moments which had significant gender differences.

Table 3-6. Gender differences in peak 3D normalised internal joint moments of hip, knee and ankle during cutting (Nm/Kg.m).

Average peak moments (Nm/kg.m), for both legs of males (♂) and females (♀) during the cutting stance phase are shown. Group differences (Nm/kg.m), partial eta<sup>2</sup> ( $\eta^2$ ) and p-values are also presented. The scale for classification of  $\eta^2$  was < 0.04 = **trivial**, 0.041 to 0.249 = **small**, 0.25 to 0.549 = **medium**, 0.55 to 0.799 = **large**, and >0.8 = **very large**.

Cut		Dominant Leg					Non-Dominant Leg				
		♂	♀	Diff	$\eta^2$	p-value	♂	♀	Diff	$\eta^2$	p-value
Hip	Flexion	<b>0.883</b>	0.605	0.28	<b>0.338</b>	0.007	<b>0.891</b>	0.548	0.34	<b>0.456</b>	0.001
	Abduction	0.425	<b>0.799</b>	0.374	<b>0.505</b>	<0.001	NON-SIGNIFICANT				
	Internal Rotation	-0.147	<b>-0.255</b>	0.108	<b>0.378</b>	0.004	NON-SIGNIFICANT				
Knee	Abduction	0.177	<b>0.360</b>	0.180	<b>0.390</b>	0.003	NON-SIGNIFICANT				
	External Rotation	0.069	<b>0.133</b>	0.064	<b>0.248</b>	0.026	NON-SIGNIFICANT				
Ankle	Max Dorsi-Plantar Flexion	<b>0.011</b>	-0.005	0.017	<b>0.330</b>	0.008	<b>0.017</b>	-1*10 <sup>-4</sup>	0.017	<b>0.317</b>	0.010

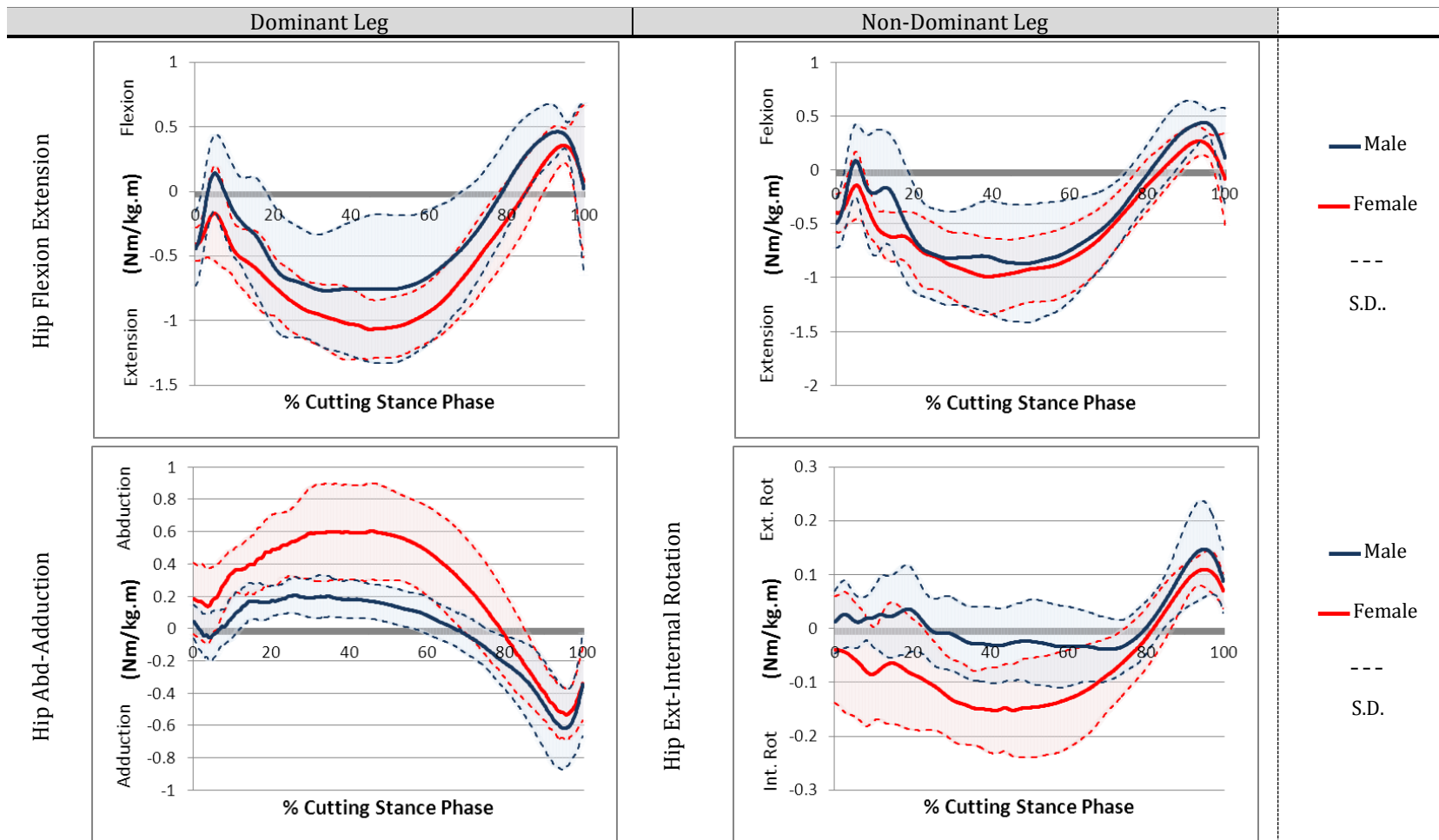


Figure 3-11 3D normalised internal joint moments of hip with significant genders, differences during cutting.

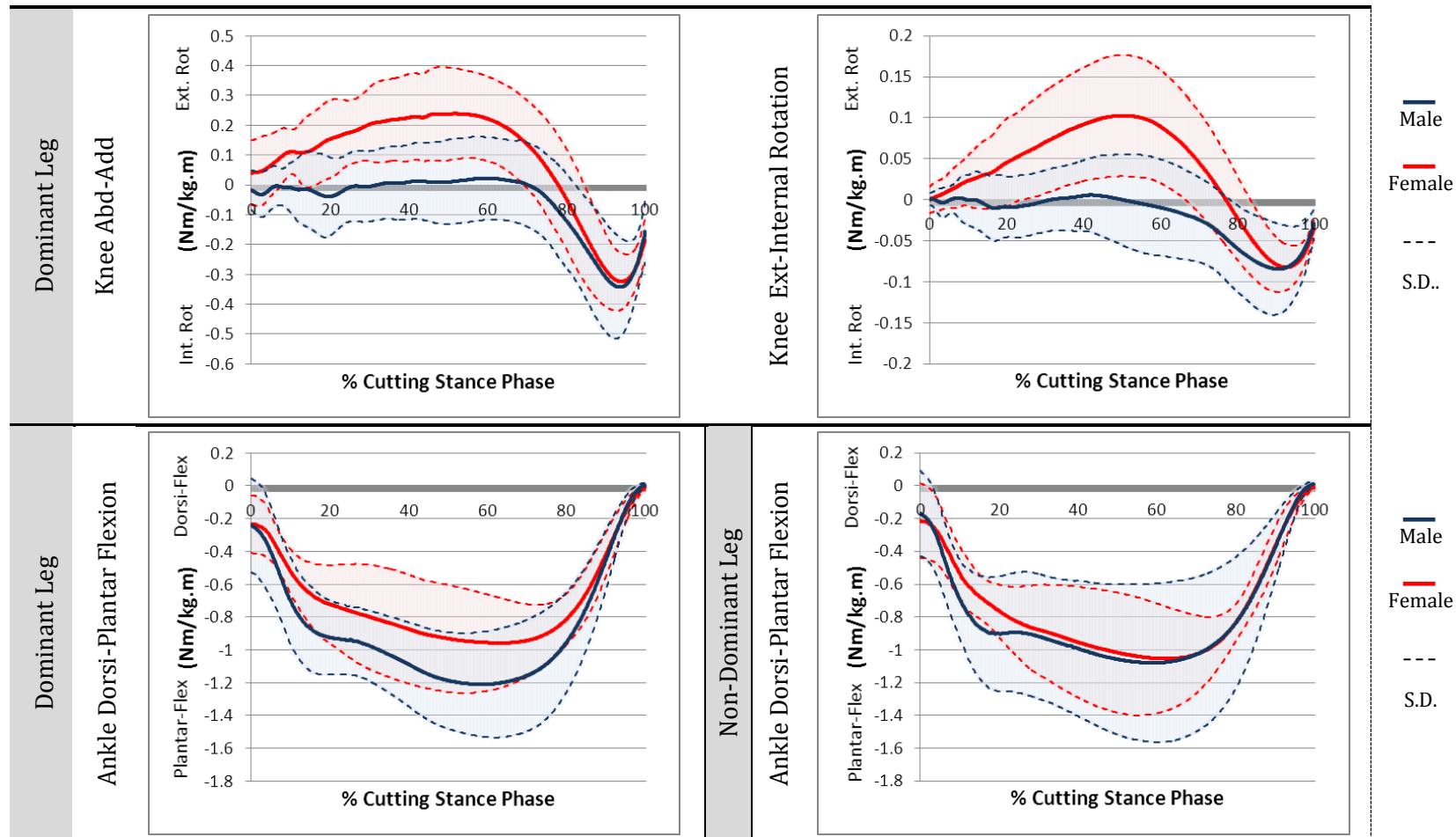


Figure 3-12 3D normalised internal joint moments of knee and ankle with significant genders, differences during cutting.

### 3.5 Discussion

In the present study, lower limb mechanics were compared between male and female athletes during a novel jump-land and cut task. Both landing from a jump and changes of direction such as side-cutting have been previously identified as high risk movements for ACL injury. The analysis of this manoeuvre is beneficial, providing further information on lower limb mechanics that may predispose females to ACL injury.

#### 3.5.1 *Gender differences in joint mechanics*

At the hip joint males and females had similar touchdown and peak flexion-extension and abduction-adduction joint angles. The similar hip flexion angles between genders is in contrast with previous findings showing decreased hip flexion in females during landing (Salci et al., 2004; Schmitz et al., 2007; Beutler et al., 2009) and cutting tasks (McLean et al., 2004; McLean et al., 2005b; Landry et al., 2007; Pollard et al., 2007; Fedie et al., 2010). Brown et al., (2009) who used a horizontal land and cut task also reported gender differences in hip flexion angles at touchdown and during the first 50% of the land-cut movement, females demonstrating increased flexion at touchdown and decreased flexion during the initial 50% of the land-cut stance. Interestingly the hip flexion angles in previous investigations cutting and landing investigations (McLean et al., 2004; Salci et al., 2004; McLean et al., 2005b; Pollard et al., 2007; Fedie et al., 2010) were 10-20° greater than the hip flexion values in the present investigation (range of ~ 10-35°). Initial contact hip flexion values of the present investigation (Males:  $11.15 \pm 7.40^\circ$ , Females:  $13.99 \pm 9.74^\circ$ ) were similar to that of Brown et al., (2009), therefore, the combination of a land and cut may induce a more extended hip posture on landing. The reduced hip flexion in both genders may increase loading on the ACL by increased knee extensor torques, which if combined with large ground reaction forces, may accelerate the tibia anteriorly (Ball et al., 1999). There were no gender differences in hip extension moments, females were expected to demonstrate increased internal hip extension moment, providing more resistance to hip flexion. The similar hip joint extension moments between genders is not surprising given the similar hip flexion angles. This replicates previous investigations such as McLean et al., (2007) and Decker et al., (2003), who both reported no gender difference in hip flexion angles and external flexion moment and internal extension moment during landing. It does contrast with Landry et al., (2009), who reported decreased external hip flexion moment in females during cutting. Females did however have decreased maximum hip flexion moment on both legs when compared to males during the cutting component of the task, providing decreased resistance to hip extension. These differences had medium effect sizes on the both legs. Peak or maximum hip flexion moment occurs in the final stages of the cutting movement, where hip extension is utilized in the push off. Increased hip extension loading at this point is not thought to increase loading on the ACL.

Hip abduction-adduction angles were similar in males and females during the landing and cutting components of the task. This is in agreement with previous landing (Kernozek et al., 2005; Ford et al., 2006; McLean et al., 2007; Pappas et al., 2007; Brown et al., 2009), cutting (McLean et al., 2004; Pollard et al., 2004a; McLean et al., 2005b; Sigward and Powers, 2006; Pollard et al., 2007; Beaulieu et al., 2009; Fedie et al., 2010) and combined landing and cutting investigations (Brown et al., 2009). This similarity in hip abduction-adduction angles between genders contradicts the views of (Ferber et al., 2003) who proposed increased hip adduction in female runners could increase knee abduction and predispose them to ACL injury risk. The lack of gender differences in hip adduction was inevitable as both males and females landed in an abducted position and remained there throughout the landing and cutting stance phase. Interestingly, analysis of hip abduction also failed to show gender differences. Hip joint adduction moments did however differentiate between the male and female subjects during the landing component of the task with a medium effect size. Males landed with greater hip adduction moment than females, therefore, the female population had decreased resistance to abduction loading in comparison to the males. This increased hip abduction load is not thought to be a risk factor for ACL injury. The decreased hip adduction moment in females may however indicate poor neuromuscular control about the hip joint that may be linked to the altered transverse plane kinematics at the hip joint in females during landing, which is discussed in further detail in the next paragraph. Females were reported to have increased hip abduction moment during the cutting component of the task, providing increased resistance to hip adduction loading in comparison to males. Increases in external hip adduction moment have been previously demonstrated in females during a cutting movement (Pollard et al., 2007). Similar to this investigation, gender differences in frontal plane hip kinetics occurred in the absence of gender differences in frontal plane hip kinematics. Pollard et al., (2007) speculated with qualitative video evidence that this occurred due to increased trunk lean. It is unclear in the current investigation why males would present with decreased resistance to hip adduction loads when compare to females. Increased ACL loading due to the decreased internal hip abduction moment in males is unlikely due to the abducted hip position adopted by both genders throughout the cutting stance.

Gender differences were present in transverse plane hip kinematics on the dominant leg during the landing component of the task. Females demonstrated increased hip internal rotation at touchdown and smaller minimum external-internal rotation on the dominant leg during the initial 40 ms and entire period of the land with medium effect sizes. Female subjects also had less maximal external rotation at the hip during the initial 40 ms and entire landing period, with the average female hip remaining in an internally rotated posture. This is in agreement with previous landing and land-cut investigations (Lephart et al., 2002; Brown

et al., 2009), and other investigations looking at walking and running (Ferber et al., 2003; Hurd et al., 2004). Hip internal rotation values are also within ranges of previous investigations (Lephart et al., 2002; Brown et al., 2009). Internal rotation at the hip has also been reported to occur during ACL injury, via video based studies of ACL injury occurrence (Boden et al., 2000). McLean et al., (2005a) have reported initial contact hip internal rotation to be associated with peak knee valgus moment during a cutting movement. It was proposed that the internally rotated position may compromise the activation of the medial muscle groups, previously reported to support the valgus loads at the knee (Besier et al., 2003). This idea has been further addressed by McLean et al., (2008) by modelling a side step with a forward dynamics model. It was concluded that increases in hip internal rotation velocities would act to decrease the potential for valgus-induced ACL injury. As a post analysis hip rotation velocities were analysed during the first 40 ms of landing stance and reported in Table 3-3. Both males and females reported similar external hip rotation velocities, which were reported by McLean et al., (2008) to increase risk of ACL injury. The exact mechanism for this increased ACL injury risk was not defined and has not been substantiated with further research. These similar external rotation velocities indicate no gender associated ACL injury risk but rather the potential for increased risk in both genders. Gender differences were also reported in the joint moments in the transverse plane. Transverse plane hip moments have not been previously compared between genders in landing or cutting tasks. Females demonstrated increased internal rotation moment on the dominant leg, during the cutting component of that task. This increase in hip internal rotation moment provides increased resistance to external rotation loads. This internal rotation moment coincides with a tendency for hip external rotation at the end of the cutting stance just prior to push off. ACL injuries rarely occur in this period of the cutting stance as ground reaction forces and ACL load decreases. Therefore, this increased hip internal rotation moment in females is not considered as a factor associated with increased ACL injury risk.

There were no gender differences in sagittal plane knee joint kinematics during the landing or cutting components of the task. Males and females both landed with knee flexion angles not considered to be near extension; on average  $>20^\circ$ . ACL loading literature has found that ACL strain is greatest when the knee is loaded near full extension ( $< 20^\circ$ ), and reduces with increased flexion (Arms et al., 1984; Renström et al., 1986; Markolf et al., 1995). Decreased knee flexion has been widely proposed as a potential risk factor for ACL injury (Shultz et al., 2010). Previous investigations have shown decreased knee flexion in females when compared to males during landing (Huston et al., 2001; Lephart et al., 2002; Decker et al., 2003; Salci et al., 2004; McLean et al., 2005b; Beutler et al., 2009), and cutting tasks (Malinzak et al., 2001; James et al., 2004; McLean et al., 2004; McLean et al., 2005b; Fedie et al., 2010). Brown et al., (2009) also reported similar knee flexion angles in both genders, as was the case



for the hip flexion angles reported by Brown et al., (2009) and the current investigation. It is likely that a task involving a combination of landing and cutting alters the sagittal plane kinematics of the hip and knee in comparison to a landing or cutting task alone. The internal knee joint moments did show some gender differences during the first 40 ms of the land. During this period, females demonstrated decreased knee flexion moments, providing decreased resistance to extension moments at the knee. This flexion moment occurred in the first 10% of the landing phase demonstrating a different deceleration strategy employed by the females, which is likely to involve increased quadriceps and decreased hamstring activation as previously demonstrated in females (Hewett et al., 1996; Huston and Wojtys, 1996; Wojtys et al., 1996; Lephart et al., 2001; Malinzak et al., 2001; Padua et al., 2005; Zazulak et al., 2005; Ahmad et al., 2006; Padua et al., 2006; Nagano et al., 2007; Beaulieu et al., 2008; Hanson et al., 2008; Myer et al., 2009; Ebben et al., 2010). Muscular activations were not measured in the current investigation but if increased quadriceps and decreased hamstring activations were present this would increase the anterior shear strain on the ACL and associated risk of ACL injury (Wascher et al., 1993; Beynnon et al., 1995; Markolf et al., 1995; DeMorat et al., 2004; Myer et al., 2005; Yu and Garrett, 2007).

No gender differences were present for maximum or touchdown frontal plane knee angles during the landing component of the task. Females did however show increased knee abduction angles and frontal plane ROM in the cutting component of the task. This was demonstrated in the non-dominant leg and dominant leg respectively. As presented in Figure 3-9 the maximum abduction adduction angle had some variability between gender groupings. Females consistently (apart from one subject) demonstrated average maximum abduction adduction angles that were greater than zero (in abduction). Males however had a mixture of subjects with maximum abduction adduction angles above and below zero. This increased knee abduction in the female group has been previously identified in females during cutting tasks (Malinzak et al., 2001; Ford et al., 2005b; McLean et al., 2005b; Beaulieu et al., 2009). Hewett et al., (2005) also showed that females athletes who demonstrated greater knee abduction angles in landing went onto receive an ACL injury. Even though there is a definite increase in knee abduction in females ( $4.07^\circ$  with medium effect size) the magnitude of the knee abduction angles demonstrated are still relatively small ( $3.20 \pm 2.2^\circ$ ) and within ranges for values demonstrated by male subjects in previous investigations (Malinzak et al., 2001; McLean et al., 2005b; Beaulieu et al., 2009), knee abduction values reported for females who received future ACL were  $\sim 9^\circ$  again well above the values presented in the current investigation. This increase in knee abduction demonstrated by the female subjects is unlikely to lead to increase in ACL strain and associated injury risk. From Figure 3-9 it is noticeable that only one subject is near this range, one male. Females were expected to demonstrate decreased knee adduction moments, providing less resistance to abduction

loads at the knee joint. Males and females however reported similar knee adduction moments. This result is in contrast to previous investigations reporting increased external abduction moment in females during landing (Chappell et al., 2002; McLean et al., 2007) and cutting (Sigward and Powers, 2006) and decreased internal adduction moments in females (Kernozek et al., 2005) during landing. The similar knee abduction moments in males and females during landing, is interesting considering the increased internal rotation at the hip in females during this component of the task. Previous research has suggested increased valgus moment in the presence of increased hip internal rotation (McLean et al., 2005a). This may be due to the similar hip internal rotation angles at touchdown in males and females and the similar knee abduction angles during landing. It is interesting to compare the internal knee adduction moments reported by Kernozek et al., (2005) (after normalisation to height: Males  $0.904 \pm 0.404$ , Females  $0.574 \pm 0.426$  Nm/kg.m) to the values in the present investigation during the land (Males  $-0.269 \pm 0.131$ , Females  $-0.232 \pm 0.109$  Nm/kg.m). The internal knee adduction moments in the present investigation for both genders are small and provide limited resistance to knee abduction loads at the knee. Internal knee adduction moments during the cutting component of the task are similar (Males  $-0.387 \pm 0.16$ , Females  $-0.399 \pm 0.10$  Nm/kg.m). Given the absence of large knee abduction angles in the majority of subjects, this decreased resistance to knee abduction load should not increase ACL load to the extent to increase injury risk. Females did demonstrate increased knee abduction moment during the cutting component of the movement, providing increased resistance to knee adduction loads at the knee when compared to males. This increase in knee abduction moment coincides with an adducted knee posture in the female subjects throughout the first 70% of the cutting stance phase. This combination is considered relatively safe position in terms of ACL loading.

Females were expected to have increased internal rotation at the knee during the landing and cutting components of the task. This was not the case with males demonstrating smaller minimum knee ext-internal rotation on the non-dominant leg during landing, with no gender differences present during cutting. As illustrated in Figure 3-5 a greater number of male subjects move through an internally rotated knee position than females. This is in contrast to McLean et al., (2007) who reported increased internal rotation at the knee in females during landing. Females have also been reported to have decreased internal rotation at the knee in comparison to males (McLean et al., 1999; McLean et al., 2004). This result is counter intuitive as females are reported to have increased risk of ACL injury and internal tibial rotation has been shown to contribute to ACL loading (Markolf et al., 1995). In terms of a group difference, the smaller minimum ext-internal rotation at the knee in males has a small effect size which may be due to the large between subject variability present in the male group. On inspection of Figure 3-5, it is evident that the majority of subjects have minimum knee ext-internal rotation values greater than zero (in external rotation), four males have

average minimum knee ext-internal rotation values in internal rotation. This finding is therefore, not a true reflection of increased magnitude of internal rotation in males, more of the increased tendency for males to have internally rotated non-dominant knee in landing in comparison to female's externally rotated knee. This internal rotation at the knee demonstrated by the 4 male subjects could increase strain on the ACL (Markolf et al., 1995), internal tibial rotation alone is unlikely to cause sufficient ACL strain to incur injury and is greatest in combination with knee flexion angles near extension. Males were also shown to have twice the knee transverse plane ROM of females during the initial 40 ms of landing. This increased transverse plane ROM as shown in Figure 3-7 indicates that males may utilise knee external rotation more than females when decelerating from a land. Given that males and females both have flexion angles  $> 20^{\circ}$  the increased risk associated with this tendency for internal rotation of the knee in males is minimal.

Females were expected to have decreased external rotation knee moments, providing less resistance to internal rotation. Given the internally rotated posture of the male subjects during landing it is not surprising that the females have increased external rotation moment during both the cutting component of the task. This increased external rotation moment in females, provides increased resistance to internal rotation and may explain the externally rotated position in females during the cutting component of the task. This is in contrast to the findings of McLean et al., (2007), who reported increased knee internal rotation angle and external internal rotation moment in females. The differences in task demands between the present investigation and that of McLean et al., (2007) may explain the difference in transverse plane knee mechanics. McLean et al., (2007) utilised a drop land from a 50 cm bench, where this investigation utilised the land from a maximal drop-jump. Landing following a jump involves more flight time prior to landing than landing from a drop from a bench. The extra flight time involved in landing from a jump may allow for external rotation of the tibia prior to landing which does not occur in a drop from a bench.

At the ankle females were expected to demonstrate a landing strategy linked with increased energy absorption at the ankle (Decker et al., 2003; Kernozek et al., 2005; McLean et al., 2007). This was not the case with similar sagittal plane ankle kinematics between genders during landing. It is possible that the combination of a jump and cut utilised less ankle flexion than previous landing investigations due to the concentration on the visual cue for direction and side cutting movement after landing. McLean et al., (2005b) utilised a horizontal jump-land followed by a perpendicular push off, and also showed similar ankle kinematics between genders. During the cutting component of the movement in the present investigation, females were reported to have smaller minimum dorsi-flexion and more dorsi-flexion ROM on the dominant leg. Minimum dorsi-flexion occurred during the push-off on the cutting movement. This decreased minimum dorsi-flexion and increased dorsi-flexion ROM in females with

medium effect sizes indicating that females utilised more plantar flexion in the push off than males. This is combined with non-significant increases in extension at the knee (Males  $-35.02 \pm 9.34^\circ$ , Females  $-28.22 \pm 6.99^\circ$ ,  $\eta^2=-0.159$ ,  $p=0.082$ ) and hip (Males  $6.66 \pm 9.03^\circ$ , Females  $5.75 \pm 9.48^\circ$ ,  $\eta^2=0.003$ ,  $p=0.829$ ). It is possible that the females in this group are utilising a triple extension strategy to produce maximal speed and power during the push off. Moments demonstrated by both genders at the final stage of the cut during push off also differ. Females pushed off with a plantar-flexion moment, which would aid the triple extension and ankle dominated push off. Even though the ankle kinematics do not support the theory of an altered female landing strategy linked with increased energy absorption at the ankle (Decker et al., 2003; Kernozek et al., 2005; McLean et al., 2007), the ankle joint moments do provide some support. Males had greater plantar flexion moment at touchdown and during the first 40 ms of the land and females had greater dorsi-flexion moment during the entire land. Greater plantar flexion moment, provides increased resistance to dorsi-flexion loads in the males. Greater dorsi-flexion moment, provides increased resistance to plantar flexion in the females. These loading patterns indicate the potential for increased dorsiflexion loading, and potentially increased energy absorption at the ankle in females. It is unclear from the current data whether this would directly reduce energy absorption at the knee and therefore, ACL injury risk. This altered ankle loading strategy during landing in females does however merit future investigation into the associated risk to ACL injury.

Females were expected to have greater eversion at the ankle which would contribute to increased knee valgus. In this investigation males and females remained in an inverted ankle position throughout the landing and cutting stance phases. This is in contrast with previous work (Ford et al., 2006), reporting increased ankle eversion in females when compared to males during single leg landings. Increased ankle eversion has been suggested as a causal factor in increased knee abduction loads and therefore, ACL strain (Nyland et al., 1999). Task design may have an influence on ankle eversion especially when bi-lateral landings are compared with single leg landings. It is likely that increased frontal plane stability associated with a bi-lateral landing may lead to a more inverted ankle position, or certainly produce less eversion. Cutting studies have also shown differences in ankle eversion (Ford et al., 2005b), which again contrasts with the findings of the present study. The subjects utilised in Ford et al.'s, (2005b) investigation were adolescents, and it is likely that differences in ankle inversion-eversion motion may be linked with increased control at the ankle joint in the highly experienced adult subjects utilised in the current investigation. No gender differences have been previously reported in inversion/eversion moments at the ankle during landing or cutting. Females are reported to have increased inversion moments on the dominant leg during landing, providing increased resistance to eversion loads. This would be considered a safe loading pattern and combined with the inversion angles present at the

ankle females do not seem to demonstrate any risk of increased ACL loading from distal ankle eversion angles or loadings.

### 3.5.3 *Directions for future Research*

Based on previous research identifying a female specific landing strategy, utilising increased energy absorption at the ankle and the support for this from the current investigation, future research into the association between this strategy and ACL injury risk is merited.

There are large amounts of between and within subject variability as presented in Figure 3-5 and Figure 3-9. Within subject variability seems to be larger in males which has been previously suggested as a factor for decreased injury risk in males (Pollard et al., 2004b). Further investigation of kinematic, kinetic and coordination variability during this novel drop-jump land and cut task may provide further evidence for the gender disparity in movement variability.

Hip internal rotation was the only notable ACL injury risk associated solely with the female population, therefore, future research investigating the impact of hip rotation postures on ACL loading is merited. Investigation into the reasons for demonstration of increased hip internal rotation are also merited, i.e. the role of muscular weaknesses and maturation.

## 3.6 **Conclusions**

On the basis of the research outcomes obtained for the population tested, the following conclusions can be drawn:

- There were no gender differences at the hip, knee or ankle in sagittal plane joint angles or internal moments that will increase ACL shear loading and injury risk during a drop-jump land and cut task.
- Males and females both perform a drop-jump land and cut task with small hip flexion angles in comparison to previous investigations.
- Females demonstrated increased maximum knee abduction angles during the cutting component of the drop-jump land and cut task, when compared to men. These abduction angles were however not considered large enough to increase ACL injury risk.
- Males and females both performed a drop-jump land and cut task with small adduction moments in comparison to previous investigations.
- Females demonstrated increased hip internal rotation angles during the landing component of the drop-jump land and cut task, when compared to men.
- Males and females both demonstrated external rotation angular velocity at the hip during the first 40 ms of the landing component of the drop-jump land and cut task.

The only notable ACL injury risk associated solely with the female population was increased internal rotation on the dominant hip during landing. Hip internal rotation alone is unlikely to increase the risk of ACL injury, but this finding does provide support for the premise that altered proximal control may influence the incidence of ACL Injuries in female athletes.

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## **CHAPTER 4**

# **THE ROLE OF GENDER ON THE VARIABILITY OF JOINT KINEMATICS AND KINETICS IN UNINJURED ATHLETES DURING A MATCH SPECIFIC LAND-CUT TASK**

## 4.1 Abstract

This study compared the movement and coordination variability of male (n=10) and female (n=10) subjects. The variability of lower limb joint angles and couplings were calculated during a maximal drop jump land and unanticipated cutting task (n=20). Females were reported to have reduced hip and knee angle variability and less coordination variability for several joint/segment couplings. This may explain the gender bias in ACL injury incidence, the reduced movement variability may act to induce repeated micro trauma on the ACL increasing its susceptibility to injury. Reduced coordination variability, could result in less flexible coordination patterns during match situations, with a decreased ability to adapt to perturbations experienced in game play. Poor adaptation to these perturbations may result in unsafe movement response in females which are associated with the occurrence of an ACL injury.

## 4.2 Introduction

Knee injuries occur regularly in various sports (Junge et al., 2006; Junge and Dvorak, 2007; Shankar et al., 2007; Häggglund et al., 2009) and one of the most devastating of these injuries is the rupture of the anterior cruciate ligament (ACL) which requires surgical reconstruction and a lengthy rehabilitation period. Female athletes have been highlighted as being at increased risk of ACL injury in comparison to their male counterparts (Roos et al., 1995; Bjordal et al., 1997; Myklebust et al., 1997; Myklebust et al., 1998; Arendt et al., 1999; Messina et al., 1999; De Loes et al., 2000; Olsen et al., 2004; Agel et al., 2005; Junge et al., 2006; Mihata et al., 2006; Prodromos et al., 2007; Borowski et al., 2008; Flood and Harrison, 2009; Randazzo et al., 2010; Pappas et al., 2011), with females being 2-6 times more likely to receive the injury than males (Gray et al., 1985; Dick and Arendt, 1993; Roos et al., 1995; Bjordal et al., 1997; Messina et al., 1999; Junge et al., 2006; Mihata et al., 2006; Borowski et al., 2008; Flood and Harrison, 2009).

Joint mechanics of the lower-limb have been associated with increased load on the ACL and an associated increase in the risk of ACL injury. Markolf et al., (1995) reported the combination of knee extension and internal tibial rotation or knee abduction load increase strain on the ACL. The risk of ACL injury is greatest during sporting tasks that involve deceleration such as landing (Gray et al., 1985; Hutchinson and Ireland, 1995; Myklebust et al., 2003; Cochrane et al., 2007; Hashemi et al., 2011), and changes of direction such as side-cutting (Hutchinson and Ireland, 1995; Myklebust et al., 1997; Myklebust et al., 2003; Cochrane et al., 2007). Past research investigating the gender bias in ACL injury incidence have utilised dynamic tasks such as landing and cutting. These investigations report altered lower limb joint kinematics and kinetics such as increased knee extension, internal tibial rotation and knee abduction moment in females when compared to males (Chappell et al., 2002; McLean et al., 2005b; McLean et al., 2007). Prospective studies have also identified

knee abduction moment as predictive of ACL injury in females (Hewett et al., 2005). These findings give strong evidence for increased risk of ACL injury in females due to altered joint mechanics such as knee abduction moment, during dynamic movement tasks. These joint mechanics and associated ACL loading are undoubtedly important in terms of ACL injury risk, however the distribution of these loads over numerous repetitions of a movement task that would occur during a game, or even number of seasons are also proposed as a factor with potential to influence injury risk.

Variations in motor performance across multiple repetitions of a task can be described as movement variability (James et al., 2000). Movement variability has traditionally been viewed as dysfunctional and a reflection of error in the planning and execution of movement (Schmidt et al., 1979; van Emmerik and van Wegen, 2000). Healthy and adaptive biological systems however depend on variability to ensure optimal performance (Collins, 1999; van Emmerik and van Wegen, 2000). This variability allows for the development of coordination patterns that turn the associated joints, segments, muscles, and motor units into controllable systems (Bernstein, 1967). Higher levels of movement or coordination variability provide a dimension of flexibility which allow adaptations to complex dynamic sport environments (Kelso and Tuller, 1984; Scholz, 1990; Turvey, 1990; Williams and Davids, 1999; Mullineaux and Uhl, 2010). Decreased levels of variability in complex tasks may limit the ability of an individual to adapt to perturbations such as uneven playing surface (Boden et al., 2000), being off balance, being pushed or held by an opponent or trying to evade a collision with another player (Krosshaug et al., 2007). Since more than 50% of ACL injuries occur closely following an induced perturbation (Krosshaug et al., 2007) the role of decreased variability is especially relevant for ACL injury and worthy of further investigation.

Various investigations have shown individuals who receive lower limb injuries demonstrate different levels of coordination variability than injury free individuals (Stergiou and Bates, 1997; Hamill et al., 1999; James et al., 2000; Heiderscheit et al., 2002; Hamill et al., 2005; Lewek et al., 2006; Kiefer et al., 2008; Moraiti et al., 2010). There is a dearth of research investigating gender related differences in movement or coordination variability. O Connor and Bottum (2009) and Pollard et al., (2005) have both addressed this issue. Pollard et al., (2005) assessed the gender differences in lower extremity joint and segment coupling variability during a cutting task. O Connor and Bottum (2009) also utilised a cutting task to compare the variability of knee joint kinematics and kinetics in males and females, using principal component analysis. Decreased variability was reported in the female population for sagittal and frontal plane knee kinetics (O'Connor and Bottum, 2009) and in four lower extremity joint and segment couplings (Pollard et al., 2005). These findings give some evidence for decreased movement and coordination variability in females when compared to



males during a cutting task. These findings require further investigation utilising other tasks such as landing and combinations of both landing and cutting.

#### *4.2.5 Aims and significance of the study*

The aim of this study was to compare the variability of hip and knee joint kinematics (movement variability), and lower extremity joint and segment coupling (coordination variability), between males and females during a dynamic drop-jump land and cut task. The hip and knee joint kinematics analysed were; flexion-extension, abduction-adduction and internal-external rotation. Specific joint / segment couplings were chosen based on loading patterns that are thought to increase ACL load as identified in Section 2.4.4. The joint and segment couplings analysed were; thigh abduction-adduction leg abduction-adduction (thigh-abad\_leg-abad), thigh rotation leg rotation (thigh-rot\_leg-rot), hip abduction-adduction knee rotation (hip-abad\_knee-rot), hip rotation knee abduction-adduction (hip-rot\_knee-abad), and knee rotation knee abduction-adduction (knee-rot\_knee-abad). Based on the variability hypothesis the female subjects were hypothesised to have decreased variability in all 3D joint angles and joint and segment couplings when compared to males. This investigation adds to the current literature investigating a gender comparison in the joint coupling variability of a cutting task. To the authors knowledge these analysis methods have not been employed on data from a drop-jump land and cut task, which adds new information to this research area.

### **4.3 Methods**

#### *4.3.1 Subjects*

Ten males (height  $1.80 \pm 0.09$  m, mass  $79.2 \pm 6.4$  kg) and ten females (height  $1.67 \pm 0.06$  m, weight  $64.7 \pm 6.4$  kg) between the ages of 18 and 26 years (males  $22.4 \pm 2.7$ , females  $21.8 \pm 2.3$ ), were recruited for participation in this investigation. All subjects played at a standard of intermediate/senior club, intercollegiate, inter-county or above and were categorized as highly experienced in their sport. Approval for the participation of human subjects in this investigation was granted by the University of Limerick Research Ethics Committee. Subjects had no history of serious lower limb injury and were injury free for a period of six months prior to testing.

#### *4.3.2 Experimental set-up*

Kinematic data were recorded during a dynamic land and cut task via 6 Eagle infrared Motion Analysis Corporation cameras sampling at 500 Hz, through Cortex software (Motion Analysis Corporation, v5.0, Santa Rosa, CA). Kinetic data were recorded using two AMTI force platforms (one leg to each platform) sampling at 1000 Hz. The cameras were positioned so that each marker was visible by at least 2 cameras throughout the landing and cutting

movement. Kinetic and kinematic data were simultaneously recorded through Cortex software (Motion Analysis Corporation, v5.0, Santa Rosa, CA).

A right-handed lab coordinate system was defined using a rigid L-frame containing four markers of known locations. A three-marker wand of known length was used within the calibration volume to scale the individual camera views.

The dynamic task assessment was designed to replicate high demand match situations in a lab environment, as explained in Chapter 3.

#### *4.3.5 Protocol*

Prior to initiation of the testing protocol each subject completed an informed consent form, a pre-test questionnaire to screen for any potential health risks and an injury history and sporting experience questionnaire to ensure all subjects were injury free and adequately experienced in their sport. Testing protocol was completed as described in Chapter 3. A total of 45 reflective markers, including four rigid four-marker clusters, and the remaining individual calibration and tracking markers were placed on each subject as outlined in Chapter 3. Following a static trial, subjects completed a minimum of 20 trials of the dynamic task with a random cutting direction. Ten successful trials in each direction were required from each subject. Successful trials required the subject to run in the correct cutting direction as directed by the visual cue, on the mapped out pathway orientated at 45°. Both feet were also required to land on their respective force plates during the jump land as was the appropriate cutting leg (e.g. the right leg when cutting in left direction) during the cut. Subjects received 1 minute rest between trials (Pollard et al., 2004; Pollard et al., 2005; Sigward and Powers, 2006) to prevent the potential effects of fatigue.

#### *4.3.3 Data treatment and analysis*

The following data reduction procedures were performed on data for all subjects. Cortex (Motion Analysis Corporation, Santa Rosa, CA, USA) was used to track and export raw 3D marker coordinate data. The raw coordinate and ground reaction force data were low-pass filtered with a fourth-order Butterworth filter with a 12 Hz and 50 Hz cut off frequency respectively. These cut off frequencies were selected based on similar values utilised in previous investigations utilising similar movement tasks (Besier et al., 2001; Malinzak et al., 2001; Ferber et al., 2003). The thigh, shank and foot segments were modeled as an assembly of cones, and the pelvis was modeled as a cylinder in Visual 3D™ (C-Motion, Rockville, MD, USA). Further detail on the specifications of these segment models are located in Chapter 3. The local coordinate system and joint centers of these segments were defined from the static trial.

Right-handed cartesian local coordinate systems for the pelvis, thigh, shank and foot segments of the left leg were defined to describe position and orientation of each segment; this was mirrored in the frontal and transverse planes for the right leg to ensure consistent identification of anatomical movements (e.g. abduction and internal rotation) for both legs. Three-dimensional ankle, knee, and hip angles, and thigh and leg segment rotations were calculated using a joint coordinate system approach (Grood and Suntay, 1983). Joint centres were denoted by the midpoint between the medial and lateral calibration markers for the knee and ankle joints (Grood et al., 1984; Wu et al., 2002) and one quarter the distance between the greater trochanter markers in the medial direction (Weinhandl and O'Connor, 2010) for the hip. Initial ground contact and take off were determined by the instant when the vertical ground reaction force exceeded or fell below 10 N respectively. The end of the jump landing phase and the beginning of the cutting movement were defined specific to each trial by the pattern of the vertical ground reaction force on the cutting leg. The landing and cutting components of the task were separated. The land was defined from touchdown to the end of landing, and the cut from the start of cutting to take off on the push off leg. These two components were cropped and time normalized to 1001 data points.

Intralimb couplings were created for the segments and joints listed in Section 4.2.5, from the time normalised three-dimensional lower limb segment rotations and joint angles. Using angle-angle plots quantification of joint coordination was obtained using a modification of the vector coding technique advocated by Sparrow (1987) consistent with previous work (Heiderscheit et al., 2002; Pollard et al., 2005). Coupling angles were calculated using the orientation to the right horizontal of the resultant vector between two adjacent data points in a timeseries as illustrated in Figure 4-1.

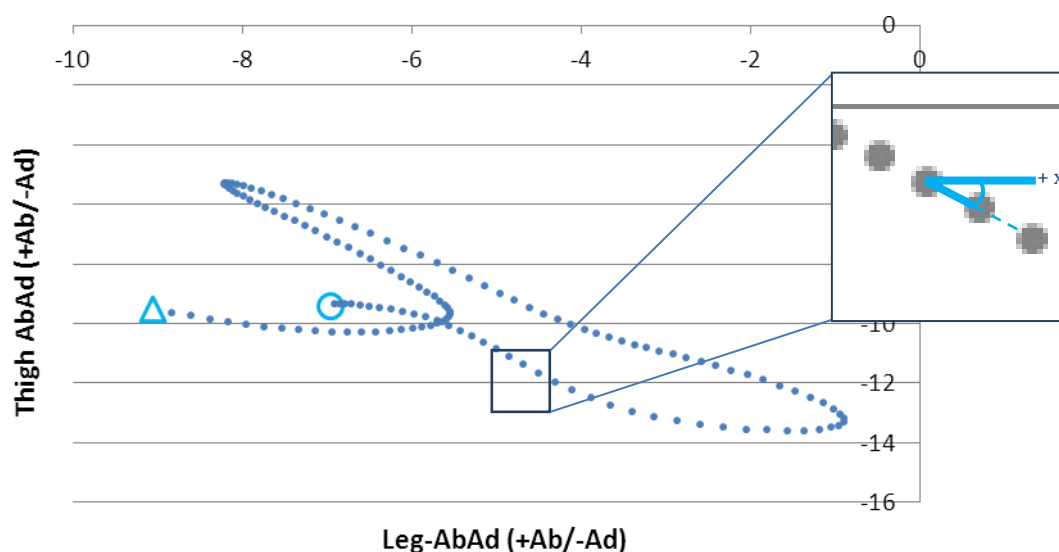
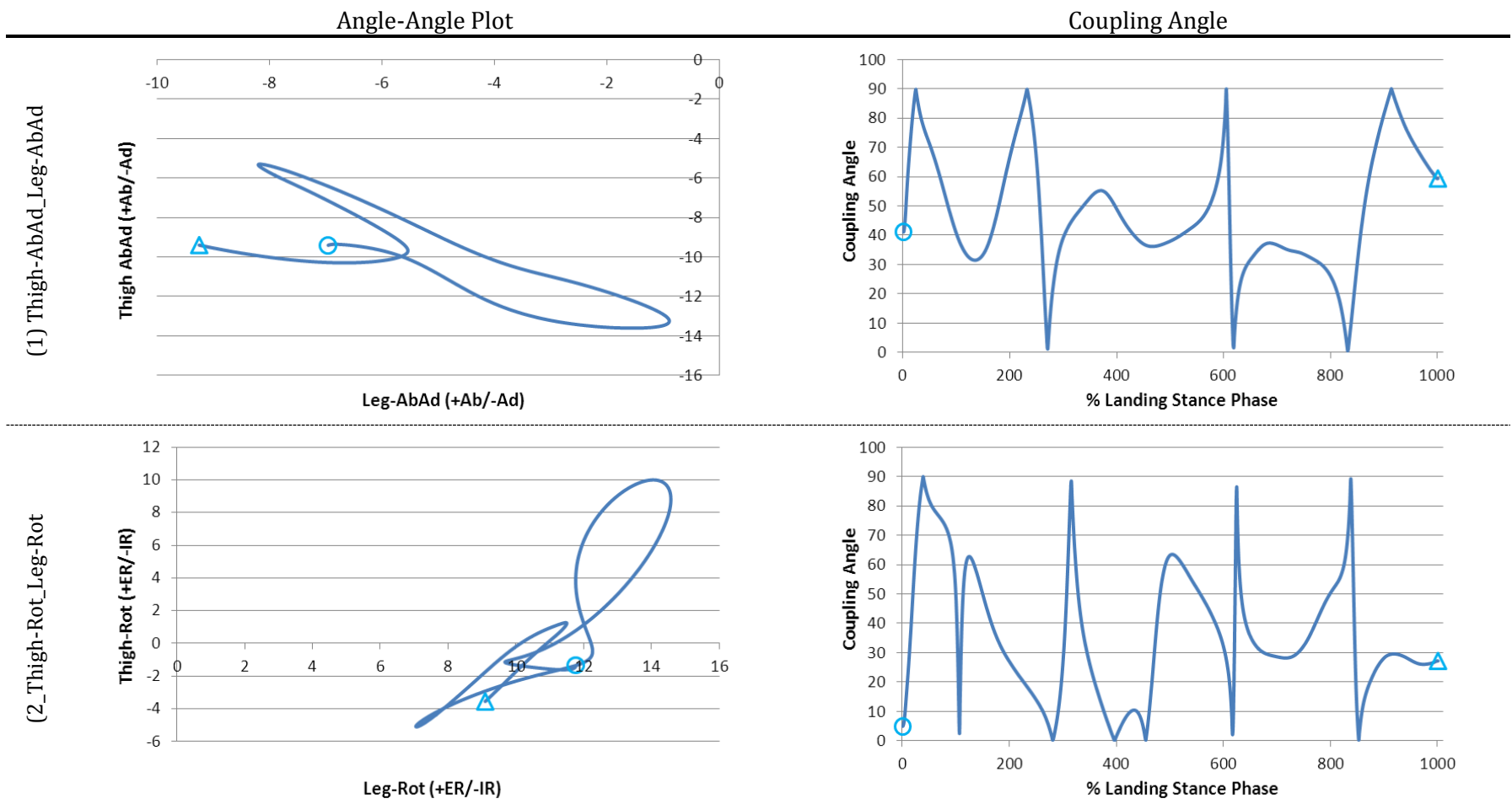


Figure 4-1. Illustration of vector coding technique to calculate coupling angle

Following conversion from radians to degrees, the resulting coupling angles were between 0 - 180°. These values were then converted into a range between 0-90°. Therefore, a coupling angle of 45°, 90° or 0° would indicate equal contribution from both segments/joints or symmetrical motion. Alternatively a coupling angle of 90° or 0° would indicate no contribution from the proximal or distal segment/ joint respectively or an asymmetrical motion. Some examples of angle-angle plots and their respective coupling angles are shown in Figure 4-2, Figure 4-3 and Figure 4-4.



Angle-Angle Plot

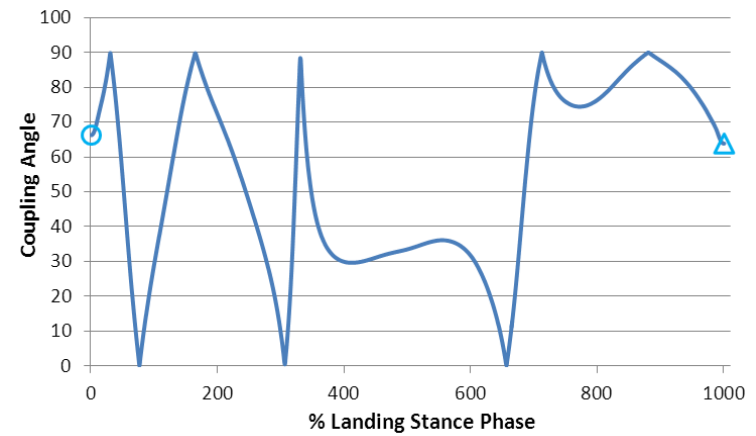
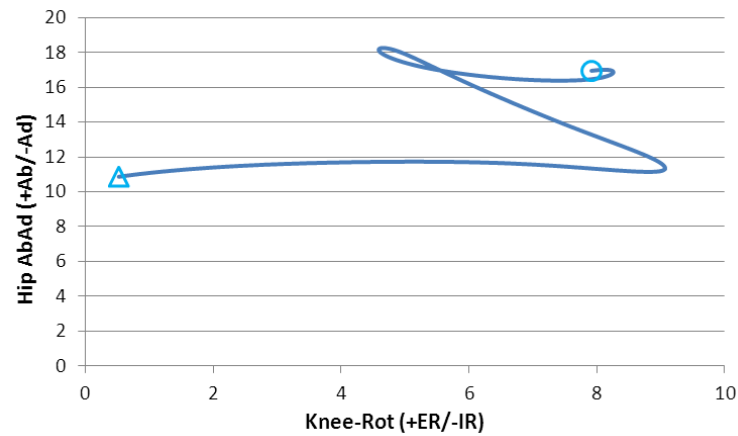
(2) Thigh-Rot\_Leg-Rot

Coupling Angle

Figure 4-2 Example of one subject's dominant leg angle-angle plots (°) and coupling angle timeseries (°) for each coupling during the landing component of the task.

(○ = Touchdown, △ = End of Land)

(3) Hip-AbAd\_Knee-Rot



(4) Hip-Rot\_Knee-AbAd

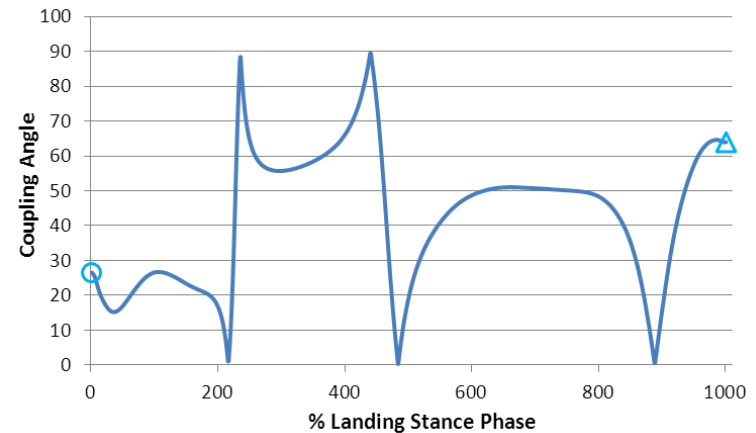
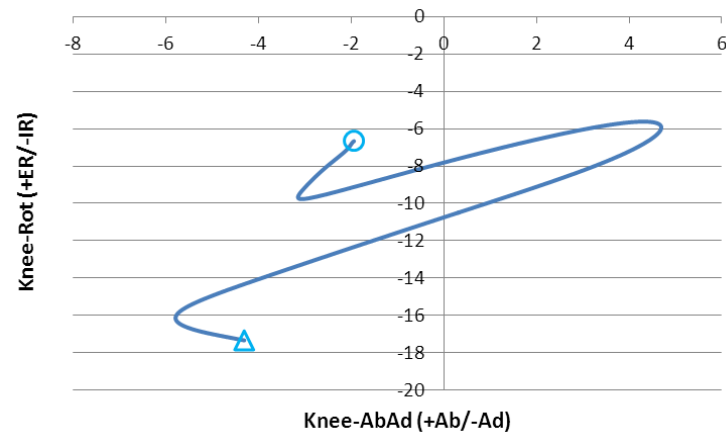


Figure 4-3 Example of one subject's dominant leg angle-angle plots ( $^{\circ}$ ) and coupling angle timeseries ( $^{\circ}$ ) for each coupling during the landing component of the task.

( $\circ$  = Touchdown,  $\triangle$  = End of Land)

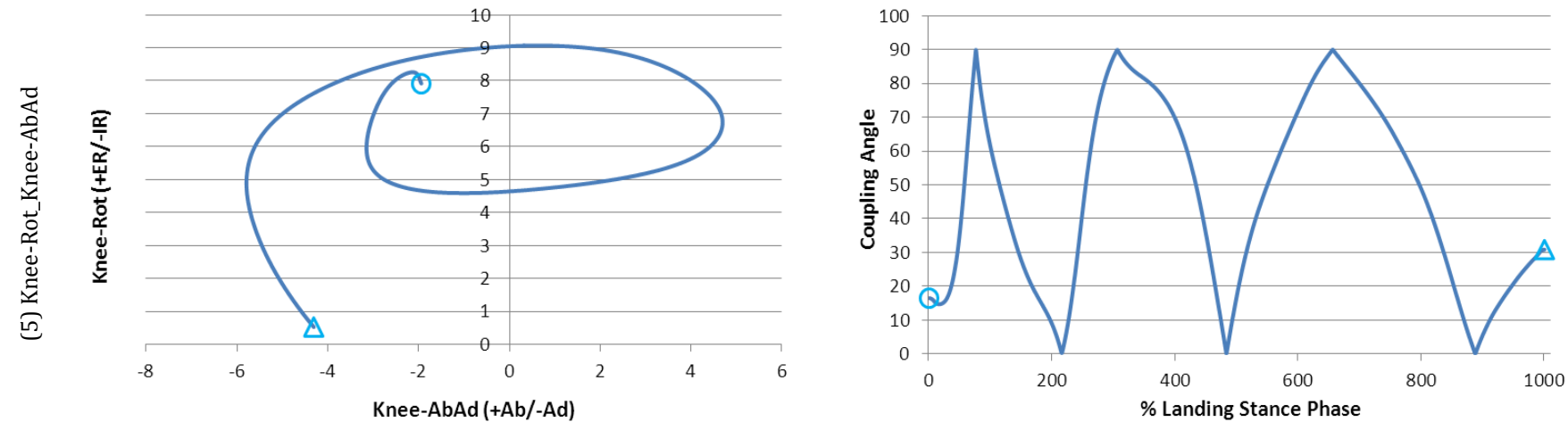


Figure 4-4 Example of one subject's dominant leg angle-angle plots (°) and coupling angle timeseries (°) for each coupling during the landing component of the task.

(○ = Touchdown, △ = End of Land)

Variability of the normalized timeseries data for the five coupling angles, and 3D joint angles of the hip and knee were calculated on a point by point basis from the mean ensemble curves of 20 and 10 trials for the land and cut data respectively. At each point of the stance phase for a specific variable (1-1000) a standard deviation value was calculated across the 20/10 trials, resulting in a 1000 point variability timeseries for that specific variable. Heiderscheit et al., (2002) suggested that averaging the variability within specific regions of the stance phase rather than across the entire stance phase would provide a more precise analysis for detecting between-group differences, Pollard et al., (2005) also utilised this method. The regions used for the calculation of average variability for the landing component of the task were, the initial 40 ms and initial 40% of the landing stance, between 15-30% of landing stance and the entire 100% of the landing stance period. The initial 40 ms of landing corresponds to the period where ACL injuries are suggested to occur (Krosshaug et al., 2007; Koga et al., 2010), and the initial 40% has been identified as the initial loading/ deceleration phase. Between 15-30% of the landing phase was utilised as this was where most subjects were found to have minimum knee flexion. ACL loading literature has found that ACL strain is greatest when the knee is loaded near full extension ( $<20^\circ$ ), and reduces with increased flexion (Arms et al., 1984; Renström et al., 1986; Markolf et al., 1995). Decreased knee flexion has also been widely proposed as a potential risk factor for ACL injury (Shultz et al., 2010). The entire landing phase will give an indication of the variability during the duration of foot contact for the land. The regions used for calculation of average variability for the cutting component of the task were the between 70-100% of cutting stance and the entire 100% of the cutting stance period. Between 70-100% was identified as the region where minimum knee flexion occurred during the cutting manoeuvre and was utilised for reasons already identified. Again the calculation of average variability during the entire cutting stance phase will give an indication of the variability during the duration of foot contact for the cut.

#### 4.3.3 *Statistical analysis*

Average variability in each of the specified regions was calculated for each variable during the landing and cutting components of the task. Gender differences in average coordination and movement variability were assessed using independent t-tests and Mann-Whitney tests for normally and not normally distributed data respectively. These tests were used to determine differences between genders. Significance for all tests was set at  $p < .05$ , except in cases where a moderate - large effect size was present with a p-value  $<0.1$ , in such cases significance was set at  $p < 0.1$ . All statistical analyses were performed using SPSS (PASW v18.0, IBM Inc. Armonk, NY). Cohen's d was utilised as a measure of effect size. This was calculated as the mean difference between genders divided by the pooled standard deviation. Interpretation of effect size was based on the scale proposed by Cohen (1988), 0.2, 0.5 and +0.8 represented small, moderate and large differences respectively.



## 4.4 Results

### 4.4.1 Jump Landing

Table 4-1 presents a summary of the significant differences present in variability of the 3D joint angles at the hip and knee joint when averaged across four regions of the landing stance phase.

Table 4-1. Significant gender differences in the variability of 3D Knee Joint Angles during landing.											
Average variability (°) for both legs of males (♂) and females (♀) during the initial 40% and 40 ms of landing and also during 15-30%, and 100% of landing stance phase are shown. Group differences (°), Cohen's d and p-values are also presented. The scale for classification of Cohen's d effect size was, 0.2 = <b>small</b> , 0.5 = <b>moderate</b> , > 0.8 = <b>large</b>											
Jump Land		Dominant Leg					Non-Dominant Leg				
		♂	♀	Diff	d	p-value	♂	♀	Diff	d	p-value
Abduction Adduction											
	15-30%	<b>1.62</b>	1.10	0.53	<b>1.06</b>	0.042					
Internal-External Rotation											
Knee	Initial 40%	<b>3.26</b>	1.53	1.73	<b>1.09</b>	0.004	<b>3.10</b>	1.52	1.58	<b>1.50</b>	0.002
	Initial 40 ms	<b>3.27</b>	1.73	1.54	<b>1.11</b>	0.005	<b>3.21</b>	1.77	1.44	<b>1.30</b>	0.002
	15-30%	<b>3.47</b>	1.60	1.87	<b>1.15</b>	0.004	<b>3.33</b>	1.56	1.73	<b>1.47</b>	0.002
	100%	<b>2.66</b>	1.24	1.42	<b>1.02</b>	0.007	<b>2.79</b>	1.25	1.59	<b>1.46</b>	0.004

Males had more variability in the frontal and transverse plane knee joint angles. The variability of the transverse plane knee angle in males was greater than the females on the dominant and non-dominant leg in all regions of the landing stance phase. The variability of the frontal plane knee angle was greater in males when compared to female subjects on the dominant leg. This difference was only reported between 15-30% of the landing stance phase. No differences in variability of the sagittal plane knee angles or any of the 3D hip angles was present between the males and females during the landing component of the task. Figure 4-5 presents the variability of 3D knee joint angles that showed significant gender differences during landing.

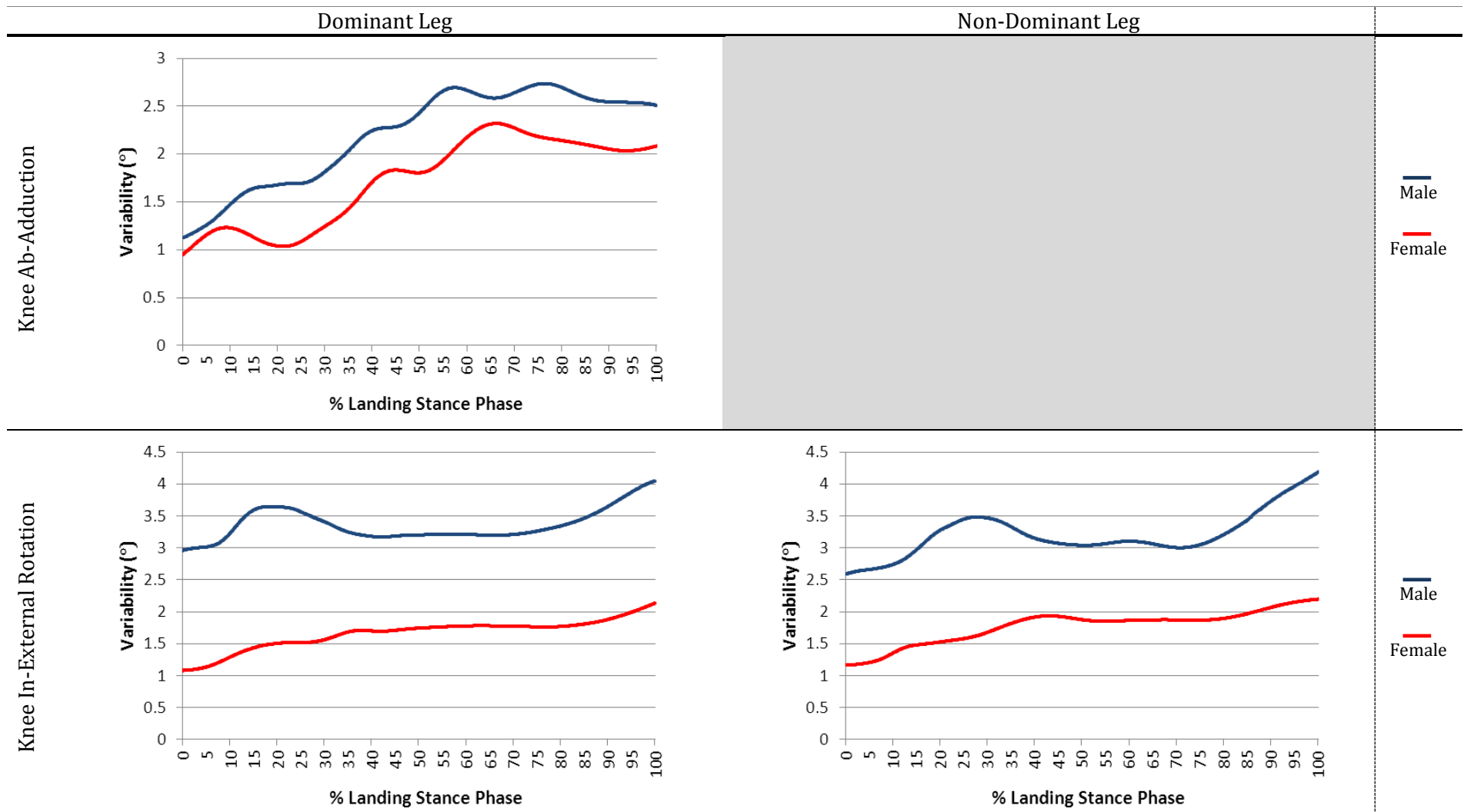


Figure 4-5. Variability of knee abduction-adduction and knee internal-external rotation angle ( $^{\circ}$ ), with significant gender differences during landing.

Table 4-2 presents a summary of the significant differences present in the variability of selected 3D joint and segment rotation vector coding couplings when averaged across four regions of the landing stance phase.

Table 4-2. Significant gender differences in the variability of selected 3D joint and segment angle coupling combinations during landing.											
Average variability (°) for both legs of males (♂) and females (♀) during the initial 40% and 40 ms of landing and also during 15-30%, and 100% of landing stance phase are shown. Group differences (°), Cohen's d and p-values are also presented. The scale for classification of Cohen's d effect size was, 0.2 = <small>small</small> , 0.5 = <small>moderate</small> , > 0.8 = <small>large</small> .											
Jump Land		Dominant Leg					Non-Dominant Leg				
		♂	♀	Diff	d	p-value	♂	♀	Diff	d	p-value
Thigh-Rot Leg-Rot	Initial 40 ms	25.37	22.56	2.81	0.83	0.062	26.20	22.55	3.64	1.08	0.011
	Initial 40%						26.60	23.15	3.47	1.33	0.001
	15-30%						27.11	24.46	2.64	0.90	0.040
	100%						25.70	22.71	2.99	1.20	0.004
Hip-Ab-Ad_ Knee-Rot	Initial 40 ms						21.10	24.08	2.98	1.09	0.010
	Initial 40%						21.08	23.50	2.42	0.924	0.035
Hip-Rot_ Knee-Ab-Ad	Initial 40 ms	20.92	18.08	2.84	0.76	0.013					
	Initial 40%	20.60	19.24	1.36	0.61	0.043					
	100%	20.20	18.20	2.00	0.96	0.023					
Knee-Rot_ Knee-Ab-Ad	Initial 40 ms	23.74	20.60	3.14	1.23	0.003					
	Initial 40%	24.00	22.12	1.88	0.92	0.036					

Males had more variability in all but two of the joint and segment couplings. Males and females had similar levels of variability in the thigh-abduction leg-abduction coupling. Males had more variability in all stages of the landing stance phase for the thigh-rotation leg-rotation coupling, when compared to females. Males had more variability in the initial stages of landing (40% and 40 ms) for the hip-abduction adduction knee-rotation coupling, hip-rotation knee-abduction adduction coupling, and knee-rotation knee-abduction adduction coupling, when compared to females. Figure 4-6 and Figure 4-7 present the variability of selected joint/segment couplings that showed significant gender differences.

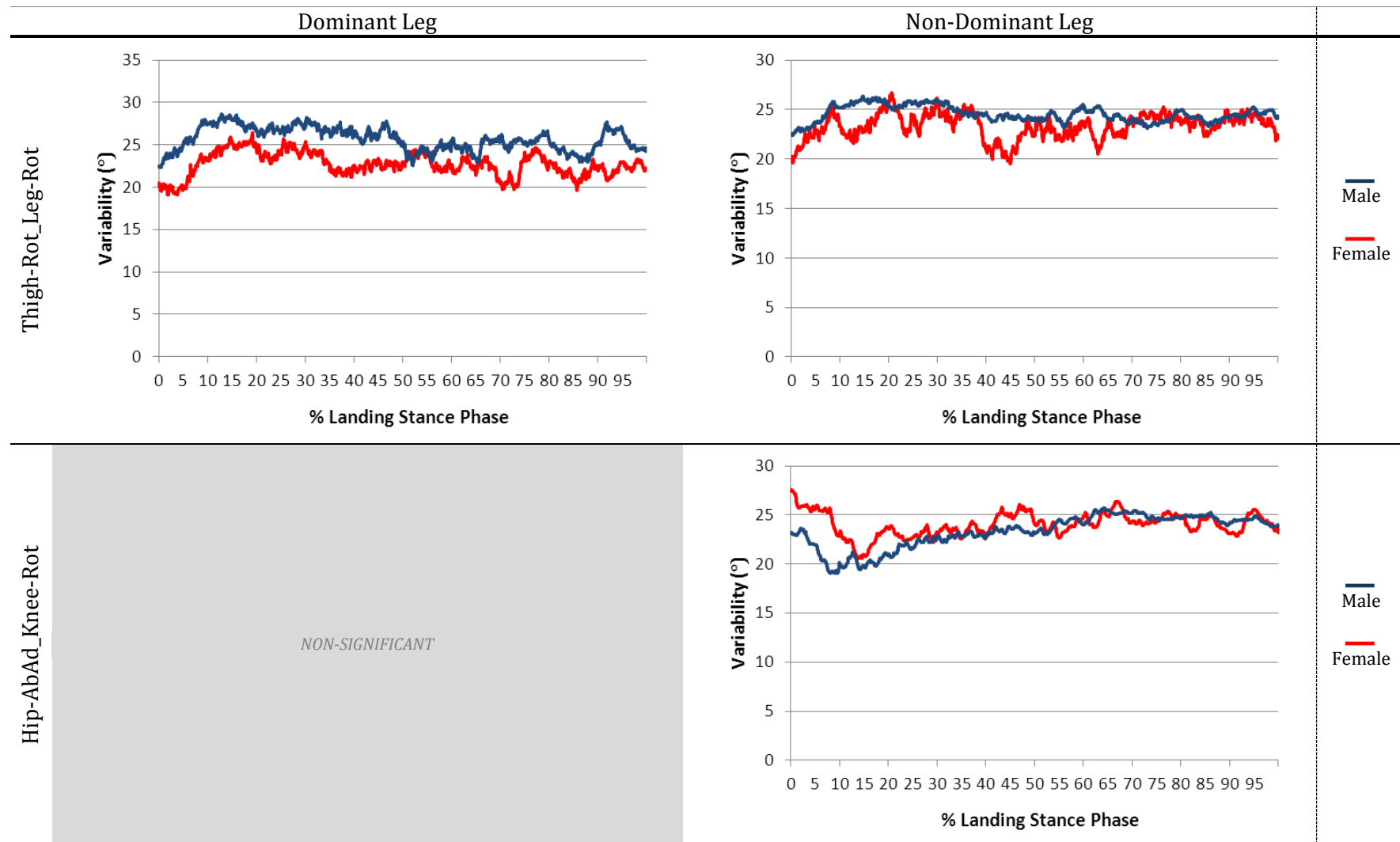


Figure 4-6. Variability of selected joint/segment couplings with significant gender differences during landing.

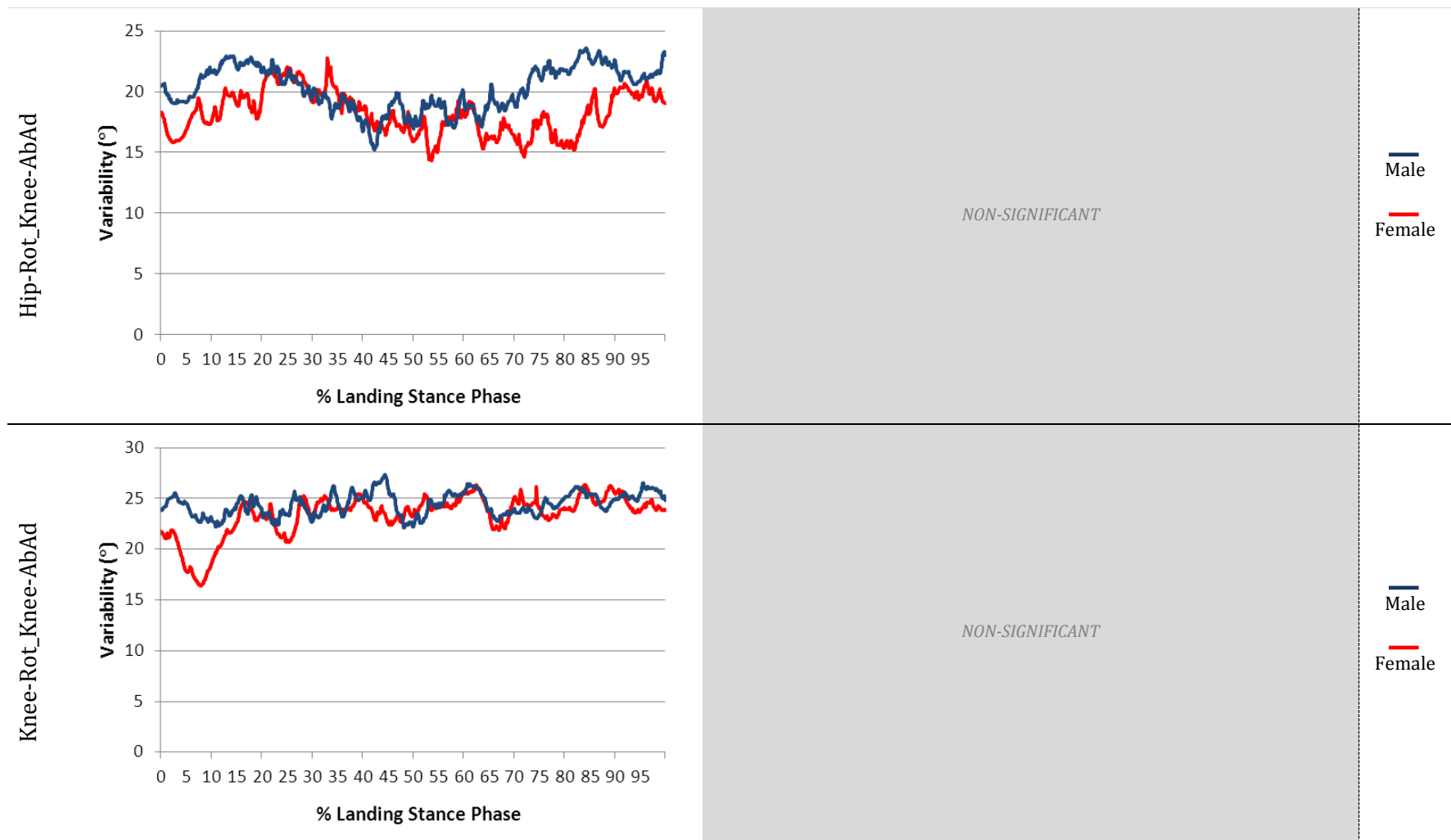


Figure 4-7. Variability of selected joint/segment couplings with significant gender differences during landing.

#### 4.4.2 Cutting Manoeuvre

Table 4-3 presents a summary of the significant differences present in variability of the 3D joint angles at the hip and knee joint when averaged across two regions of the cutting stance phase.

Table 4-3. Significant gender differences in the variability of 3D Joint Angles of Hip and Knee during cutting.											
Average variability (°) for both legs of males (♂) and females (♀) during 70-100%, and 100% of cutting stance phase are shown. Group differences (°), Cohen's d and p-values are also presented. The scale for classification of Cohen's d effect size was, 0.2 = <b>small</b> , 0.5 = <b>moderate</b> , > 0.8 = <b>large</b> . Data presented in <i>italics</i> has a large effect size and differences significant when p=0.1.											
Cut		Dominant Leg				Non-Dominant Leg					
		♂	♀	Diff	d	p-value	♂	♀	Diff	D	p-value
Hip	<b>Abduction Adduction</b>										
	70-100%						<b>4.37</b>	3.53	0.84	<b>1.00</b>	0.049
	<b>Flexion Extension</b>										
	100%						<b>7.55</b>	6.07	1.48	<b>0.88</b>	0.019
Knee	<b>Abduction Adduction</b>										
	70-100%						<b>3.05</b>	<i>1.37</i>	<i>1.68</i>	<b>1.14</b>	<i>0.070</i>
	<b>Internal-External Rotation</b>										
	70-100%	<b>3.72</b>	2.29	1.43	<b>1.46</b>	0.004	<b>4.22</b>	2.68	1.55	<b>1.18</b>	0.047
	100%	<b>3.74</b>	2.07	1.67	<b>1.91</b>	0.001	<b>4.00</b>	2.28	1.71	<b>1.39</b>	0.12

Males had more variability in sagittal, frontal and transverse plane knee joint angles, and frontal plane hip angles. Sagittal plane knee angle variability was greater in males when compared to females across the entire cutting stance phase on the non-dominant leg. Frontal plane knee angle variability was greater in males when compared to females during 70-100% of the cutting stance phase on the non-dominant leg. This difference was not significant but had a large effect size with a 76% percentage difference in variability between genders. Transverse plane knee angle variability and frontal plane hip angle variability were also greater in males when compared to females during 70-100% of the cutting stance phase on the dominant and non-dominant leg. Figure 4-8 presents the variability of 3D joint angles of the knee that showed significant gender differences during cutting.

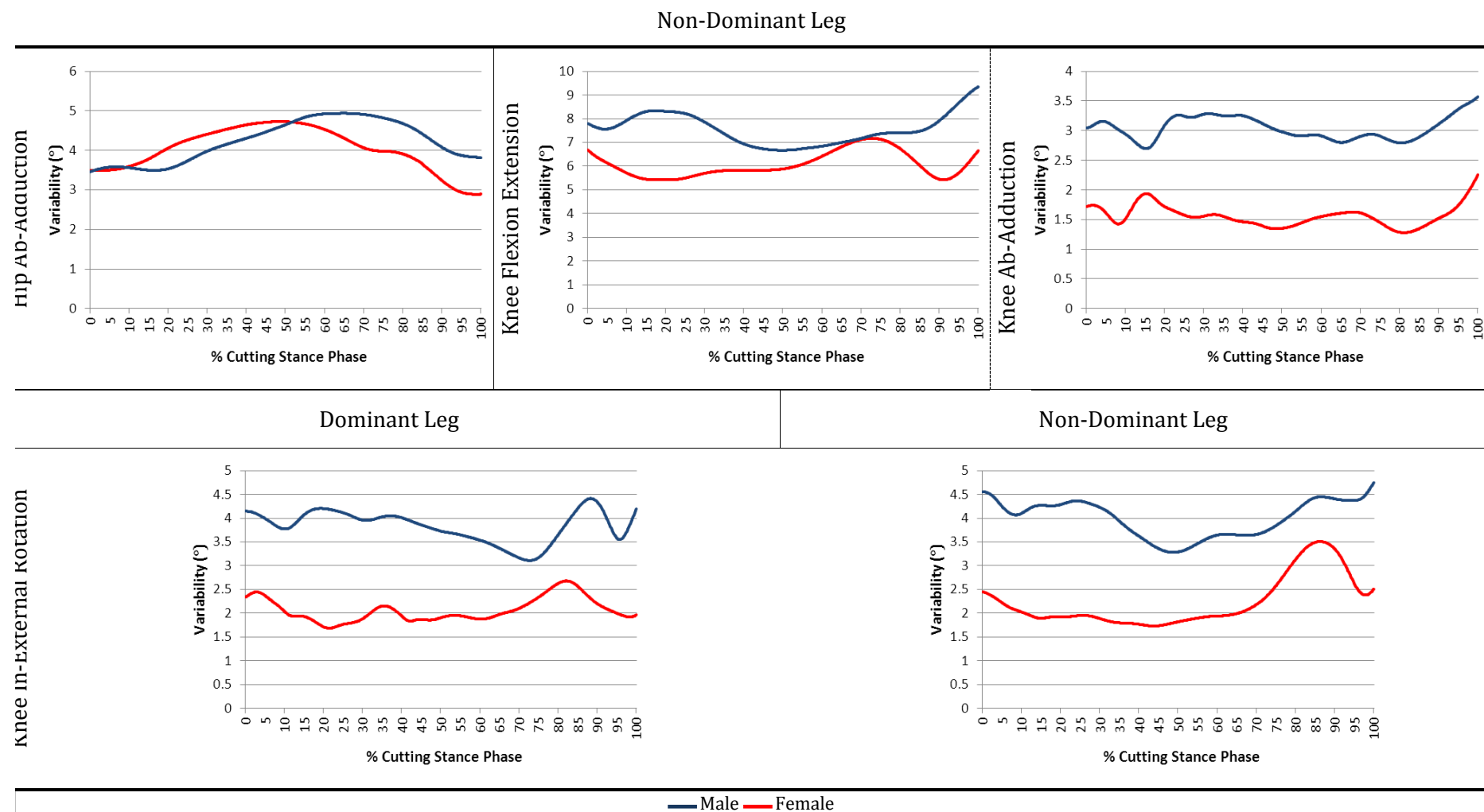


Figure 4-8. Variability of hip and knee joint angles (°), with significant gender differences during cutting.

There were no significant gender differences to report in the variability of selected 3D joint and segment rotation vector coding couplings during cutting.

#### **4.5 Discussion**

In the present study, 3D lower limb joint angles and joint/segment coupling variability was compared between male and female athletes during a maximal drop jump-land and unanticipated cutting task. This investigation adds to the current body of evidence indicating that females have less variability in joint/segment couplings (Pollard et al., 2005). These lower levels of variability have the potential to increase the risk of ACL injury in females when compared to a male population.

The variability in transverse plane knee angles was decreased in females during all regions of the landing and cutting stance on both legs. Internal rotation (Wascher et al., 1993; Markolf et al., 1995; Fleming et al., 2001) and external rotation (Fung and Zhang, 2003) of the knee joint have both been suggested as components of ACL injury mechanisms. Internal rotation at the knee has been shown to increase the load on the ACL when combined with knee extension. The decreased variability reported in the female athletes during regions of low knee flexion in landing (15-30%) and cutting (70-100%), therefore, has the potential to produce a repetitive strain on the ACL which is likely to induce wear and increase risk of rupture. In addition Fung and Zhang (2003) demonstrated an increased risk of ACL impingement when external rotation of the knee was combined with knee adduction. It is debatable if internal rotation of the knee caused increased load on the ACL or external rotation on the knee joint increasing risk of ACL impingement is more likely to induce an ACL injury. It is however apparent that transverse plane knee motion affects the ACL. Transverse plane knee motion is the transverse rotation of the shank relative to the thigh; therefore, the reduced variability of the thigh-rotation leg-rotation coupling in females when compared to males is of additional interest. This gender difference is reported in every region of landing on the non-dominant leg, and in the initial 40 ms of landing on the dominant leg.

Females also had less variability in frontal plane knee angles. Females were only shown to be less variable during regions which corresponded to minimal knee flexion (15-30% of landing, 70-100% cutting). Gender differences in variability were only present in the dominant leg during landing and the non-dominant leg during cutting. It is likely that high demands are placed on the dominant leg during landing with a tendency for individuals over-utilise the dominant leg to decelerate (Wong et al., 2007). Task demands may also be higher on the non-dominant leg during cutting (Cowley et al., 2006; Rosado, 2006), with an athlete's dominant side usually preferred for push off during side-cutting. Gender differences in the variability of frontal plane knee angles may only emerge in these limbs during these tasks due to the increased demands placed on the respective legs. Increased knee abduction angle is



commonly reported in females when compared to males during athletic movements (Malinzak et al., 2001; Hewett et al., 2004; McLean et al., 2004; Ford et al., 2005; McLean et al., 2008) and is also a predictor of future ACL injury risk (Hewett et al., 2005). Markolf et al., (1995) has identified the combination of knee abduction and internal rotation loads on a flexed knee load the ACL. The knee-rotation knee-abduction adduction coupling is therefore, especially important. Females had less variability in this coupling during the initial regions of landing (first 40% and 40 ms) on the dominant leg. ACL injuries are reported to occur most frequently during these initial regions of landing and this is also where minimum knee flexion occurs.

Although sagittal plane knee kinematics are addressed via the regions of landing and cutting utilised for the analysis of variability, the variability of the sagittal plane knee angle was also examined. The only region of landing and cutting that showed a gender difference in the variability of sagittal plane knee angle was the entire cutting stance (100%). Females had less variability of sagittal plane knee angle when it was averaged across the entire cutting stance period. This consistency in knee flexion angles in the female population could result in a more consistent shear loading pattern on the ACL over numerous repetitions of a movement task such as cutting.

The frontal plane angle at the hip was the only 3D hip angle that showed significant gender differences in its variability. Females had less variability in frontal plane hip angle during the region of lowest knee flexion during the cutting stance phase (70-100%). Ferber et al., (2003) proposed that increased hip adduction in females runners could predispose them to ACL injury risk, by the increase in knee abduction thought to occur due to a more adducted hip. The coupling of hip-abduction adduction and knee rotation was examined as an extension of the relationship between knee-abduction adduction and knee-rotation previously outlined for its role in ACL loading. Interestingly and in opposition to all the other variables, the variability of the hip-abduction adduction knee rotation coupling was greater in females compared to males. A hip-abduction adduction knee rotation loading combination has not been explicitly linked with ACL load; therefore, the increased variability in females for this coupling is not thought to be sufficient to counteract the decreased variability reported in all other couplings.

There was no gender difference in transverse plane hip angle variability; the variability of hip-rotation knee abduction adduction coupling did however differ by gender. Females demonstrated less variability in the hip-rotation knee abduction adduction coupling. McLean et al., (2005a) have reported internal rotation at the hip to be associated with peak knee valgus moment. Besier et al., (2003) proposed that the internal rotation position may compromise the activation of the medial muscle groups, previously reported to support the

valgus loads at the knee. This association between internal rotation at the hip and knee valgus moment has only been reported during cutting movements therefore, definite risk of ACL injury associated with this coupling is unknown. It is however another lower limb coupling that shows significantly less variability in the female athletes when compared to the males.

The gender differences in the variability of sagittal, frontal and transverse plane knee angles and frontal plane hip angles as previously outlined supports our hypothesis with females demonstrating less variability in comparison to males. These knee and hip angles have been previously related to ACL loading and injury risk and their consistency in the female athletes may indicate a more concentrated loading pattern which may act to induce repetitive microtrauma on the ACL and increase injury risk. The impact of, for example, a  $1.87^\circ$  difference in the variability of the transverse plane knee angle, as seen on the dominant leg during the 15-30% region of landing, may seem negligible, however when considered as a percentage of the average maximal transverse plane knee angle displacement it is equal to a 26% difference. The gender differences in the other knee and hip angles were of similar magnitude ( $14 \pm 8\%$ ), and when considered as a percentage of the average variability demonstrated by both genders, the average difference was ranged from 19% to a 78% ( $56 \pm 20\%$ ) difference. The variability of joint angles has not been analyzed and reported in this manner previously which makes comparison with previous research difficult. However when these percentage differences are considered they are in line with differences previously reported between males and females (Pollard et al., 2005) and when subjects with patellofemoral pain (PFP) were compared with controls (Hamill et al., 1999; Heiderscheit et al., 2002).

The gender differences in the variability of the thigh-rotation leg-rotation, hip-rotation knee abduction adduction and knee-rotation knee abduction adduction couplings also supports our hypothesis with females demonstrating less variability in comparison to males. As outlined these joint/segment couplings have been previously related to ACL loading and injury risk and their consistency in the female athletes may indicate a less flexible coordination pattern in these female athletes. These gender differences in joint/segment coupling only occur during the landing component of the task, where the majority of the deceleration from landing is required. If these decreased levels of variability are replicated in complex landing tasks in a match situation it may limit their ability to adapt to perturbations such as uneven playing surface (Boden et al., 2000), being off balance or being pushed (Krosshaug et al., 2007). The percentage differences between genders for joint/segment coupling (RANGE-8-15%, AVG- $15 \pm 8\%$ ) are smaller than that reported for kinematic variability, and those previously reported comparing gender and those with and without PFP (Hamill et al., 1999; Heiderscheit et al., 2002; Pollard et al., 2005). When comparing the

variability across the initial 40% from the current investigation with similar data for males and females reported by Pollard et al., (2005), the variability in joint/segment couplings from females in the current investigation is similar to the previously reported levels of variability in males (Pollard et al., 2005). Heiderscheit et al., (2002) and Hamill et al., (1999) both used continuous relative phase as a measure of coordination which makes direct variability comparison difficult. The analysis of coordination variability utilising vector coding is limited and it is plausible that the differences in task demands between the present investigation and that of Pollard et al., (2005) may explain the differences in average coupling variability values for males and females. The landing component of the task in the current investigation requires a rapid bilateral deceleration from a maximal drop-jump which differs from Pollard et al., (2005) where a single legged change of direction was executed at speed. The potentially greater demands of decelerating from a maximal jump may increase the coordination variability required to repeatedly execute the movement and explain the increased variability reported in both genders for the current investigation.

#### **4.6 Conclusion**

A reduction in variability of joint angles and joint/segment couplings was observed for female athletes in the current investigation, when compared to males during a drop-land and cut manoeuvre. Variability of knee and hip joint angles was less in females during the landing and cutting phases of the task. The variability of the joint/segment couplings was only reduced in females during the initial regions of the landing component of the task in the majority of cases.

These gender differences in movement and coordination variability may explain the gender bias in ACL injury incidence, the reduced movement variability reported in females may act to induce repeated micro trauma on the ACL increasing its susceptibility to injury. If the reduced coordination variability demonstrated in the female athletes, results in less flexible coordination patterns during match situations, they may also have a decreased ability to adapt to perturbations experienced in game play. Poor adaptation to these perturbations may result in unsafe movement response in females which are associated with the occurrence of an ACL injury.

Further research is necessary to address whether lower levels of coordination variability actually result in a less adaptable movement system. This could be implemented by comparing groups who demonstrate varying levels of coordination variability, and assess their response to an unanticipated perturbation such as landing on an uneven surface or being pulled off balance just prior to landing. Forward dynamics musculoskeletal modeling may also further inform the hypothesis on the role of movement variability and the potential occurrence of repeated micro trauma, thought to degrade and weaken the ACL.

## 4.7 References

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## **CHAPTER 5**

### **EFFECTS OF PREVIOUS ANTERIOR CRUCIATE LIGAMENT INJURY AND REHABILITATION ON JOINT KINEMATICS AND KINETICS DURING A MATCH SPECIFIC LAND-CUT TASK**



## 5.1 Abstract

This study compared the lower limb mechanics of the previously injured leg of ACL injured subjects (ACLR, n=18), against their non-injured leg and a control (nACL, n=18) leg. Three dimensional hip knee and ankle angles were calculated during a maximal drop jump land and unanticipated cutting task (n=20). No significant differences were reported between the previously injured and contralateral non-injured leg. ACLR subjects had increased hip flexion on both legs during landing when compared to nACL controls. The previously injured leg also had increased knee abduction moment, and transverse plane knee range of motion when compared to a nACL control leg, both of which have been previously linked to the development of osteoarthritis.

## 5.2 Introduction

Anterior cruciate ligament (ACL) injuries are recognised as one of the most common and serious sports injuries with upwards of 250,000 ACL injuries occurring in the United States each year (Boden et al., 2000). Reconstructive surgery is typically recommended after ACL injury, to restore the knee joint stability and function required for sports participation. Up to 80% of athletes who undergo surgery are unable to successfully return to their preinjury-level of sport participation and therefore, quit their sports (Söderman et al., 2002; Chong and Tan, 2004; Ardern et al., 2010). Athletes who are successful in rehabilitating from surgery and returning to their sport (ACLR subjects) have been shown to be at an increased risk of repeated ACL injury to both the previously reconstructed knee (Oates et al., 1999; Orchard et al., 2001; Kvist, 2004; Myklebust and Bahr, 2005; Salmon et al., 2005; Tanaka et al., 2010) and the contralateral knee (Shelbourne et al., 1998; Oates et al., 1999; Orchard et al., 2001; Salmon et al., 2005). Additionally up to 50% of ACLR individuals will display signs of osteoarthritis (OA) 10 years post injury (Roos et al., 1995; Von Porat et al., 2004; Pinczewski et al., 2007; Meuffels et al., 2009; Holm et al., 2010; Øiestad et al., 2010). Altered biomechanics and neuromuscular function as a result of the initial ACL injury, affecting both the injured and the contralateral leg, are likely to increase the risk of a repeated ACL injury (Swärd et al., 2010) and degenerative joint disease (Deneweth et al., 2010).

ACLR subjects have demonstrated altered lower limb kinematics and kinetics during everyday tasks such as walking gait (Devita et al., 1998; Bush-Joseph et al., 2001; Butler et al., 2009; Scanlan et al., 2010; Webster et al., 2011), and a range of more demanding tasks such as downhill running (Tashman et al., 2004), stair ambulation and pivot combinations (Kowalk et al., 1997; Ristanis et al., 2003; Tsarouhas et al., 2010), single leg hopping (Deneweth et al., 2010; Orishimo et al., 2010), drop vertical jumps (Decker et al., 2002), and drop land and pivot combinations (Ristanis et al., 2005). These altered biomechanics were shown to occur in the sagittal plane at the hip (Devita et al., 1998; Decker et al., 2002), knee (Kowalk et al., 1997; Deneweth et al., 2010; Orishimo et al., 2010) and ankle joint (Kowalk et al., 1997;

Decker et al., 2002), and in the frontal (Tashman et al., 2004; Webster et al., 2011) and transverse plane (Ristanis et al., 2003; Tashman et al., 2004; Ristanis et al., 2005; Deneweth et al., 2010; Scanlan et al., 2010) at the knee joint. The altered frontal and transverse plane knee joint mechanics demonstrated by ACLr subjects have been strongly proposed as influential in the development of OA in an ACLr and ACL deficient population (Stergiou et al., 2007). Transverse plane hip kinetics, frontal plane knee kinematics, and sagittal plane knee kinetics have also been identified as risk factors potentially predictive of a second ACL injury from biomechanical measures during landing (Paterno et al., 2010).

Few of these previous investigations have utilized tasks that closely replicate match situations. Ristanis and colleagues (2005) have utilized the most match specific task which involved a jumping from a 40 cm box, landing and pivoting at 90° to walk away. Bush-Joseph et al. (2001) utilized a jog and diagonal cut task as one of their higher demand activities. These tasks are definitely advancements on the previous drop vertical jump and stair descent and pivot tasks, however they still lack the unanticipated and high intensity nature of match situations. Due to the fact that ACL injury rehabilitation aims to return individuals to full competitive participation in their sport, measurement of the performance of these ACLr subjects during tasks that replicate match conditions is essential, in order to accurately identify altered joint mechanics that may predispose ACLr individuals to repeated ACL injury or the development of OA.

#### *5.2.1 Aims and significance of the study*

The purpose of this study was to evaluate and compare the lower limb kinetic and kinematic landing performances of ACLr individuals, against the contralateral non-injured leg and a healthy control (nACL) during the performance of a maximal drop-jump land and unanticipated cutting task. 3D kinematics and kinetics of the hip, knee and ankle were measured for both legs during the land, and for the push off leg during cutting in both directions. The assessment of lower limb mechanics during this novel drop-jump land and cut task will provide new information on any biomechanical adaptations present in ACLr subjects during the performance of high risk movement tasks. This information may highlight risk factors for ACL re-injury and or the development of OA, as well as inform therapists regarding the design of rehabilitation protocols.

Based on previous investigations utilizing ACLr subjects it was hypothesized that the differences detailed in Table 5-1 would be shown in the previously injured leg (PI) when compared to the contralateral non-injured (NI) and healthy control leg (C):

Table 5-1.Hypothesized ACLr joint mechanics				
Plane			PI -V- NI	PI-V-C
Sagittal Plane	Angles	Hip: Knee: Ankle:		> Extension  >Plantar Flexion >ROM >P.Flex Angular Velocity
	Internal Moments	Hip: Knee: Ankle:	>Extension > Asymmetry	>Extension > Asymmetry
Frontal Plane	Angles	Hip Knee:	>Abduction >Adduction	>Abduction
	Internal Moments	Knee:		>Abduction <Adduction
Transverse Plane	Angles	Hip: Knee: Ankle:	>ROM >Ext. Rotation	>ROM
	Internal Moments	Hip: Knee: Ankle:	NI - net impulse <0* >Int. Rotation	
	* net internal rotation hip transverse plane moment impulse = >risk			

## 5.3 Methods

### 5.3.1 Subject information

Eighteen subjects who had previously undergone ACL reconstruction, rehabilitation (ACLR subjects) and who were back in full competitive participation in their sport (Males n=9, age  $26 \pm 4$  years, height  $1.78 \pm 0.1$  m, mass  $81.74 \pm 19.42$  kg, time since injury  $5 \pm 3$  years, Females n=9, age  $22 \pm 2$  years, height  $1.69 \pm 0.06$  m, mass  $66.21 \pm 7.51$  kg, time since injury  $4 \pm 2$  years) were recruited for the present investigation through local sports clubs physiotherapists and orthopedic surgeons. All ACLr subjects were screened prior to participation, this process is outlined in the subsequent section (5.3.2 ACLr Subject

Screening). A further 18 gender, height, weight and sport matched subjects who had no history of knee injury (nACL) were also recruited for the present study (Males  $n=9$ , age  $22 \pm 3$  years, height  $1.81 \pm 0.09$  m, mass  $80.39 \pm 5.36$  kg, Females  $n=9$ , age  $22 \pm 2$  years, height  $1.67 \pm 0.07$  m, mass  $63.81 \pm 6.12$  kg). nACL control subjects had no history of serious lower limb injury and were injury free for a period of six months prior to testing. All subjects played at a standard of intermediate/senior club, intercollegiate, inter-county or above and were categorized as highly experienced in their sport. Approval for the participation of human subjects in this investigation was granted by the University of Limerick Research Ethics Committee.

### 5.3.2 ACLr Subject screening

All ACLr subjects were required to conform to a number of criteria. Firstly each ACLr subject must have received their ACL injury through a non or indirect-contact mechanism as outlined in Section 2.4.2. ACLr subjects were also required to be fully rehabilitated (cleared by their physiotherapist and surgeon) following ACL reconstructive surgery and back in full participation (training and competitive matches) in their chosen sport and injury free in both lower limbs for a 6 month period prior to testing.

All previously ACL injured subjects were also required to pass two separate screening assessments, an IKDC and Knee Injury Questionnaire, and a Functional Screening Protocol. See Appendix A3 for a copy of questionnaires. The IKDC knee evaluation form (Irrgang et al., 2001) is a knee-specific, measure of symptoms, function, and sports activity (Hefti et al., 1993; Risberg et al., 1999; Vitkus et al., 2002; Anderson et al., 2006; Crawford et al., 2007; Hambly and Griva, 2008). It is utilised in this investigation to assess to full rehabilitation of the ACLr subjects. The subject's answers to these questions were reviewed prior to participation to ensure they were capable of full sports participation without pain or giving way in the knee. The Knee Injury History Evaluation questionnaire is utilised with the ACLr subjects to gather data on the ACL injury and surgical reconstruction performed. Only subjects with a bi-lateral ACL injury, without additional lower limb injury within 6 months prior to testing were included to participate. The Functional Screening Protocol consisted of a battery of four functional ability hopping tests (Itoh et al., 1998).

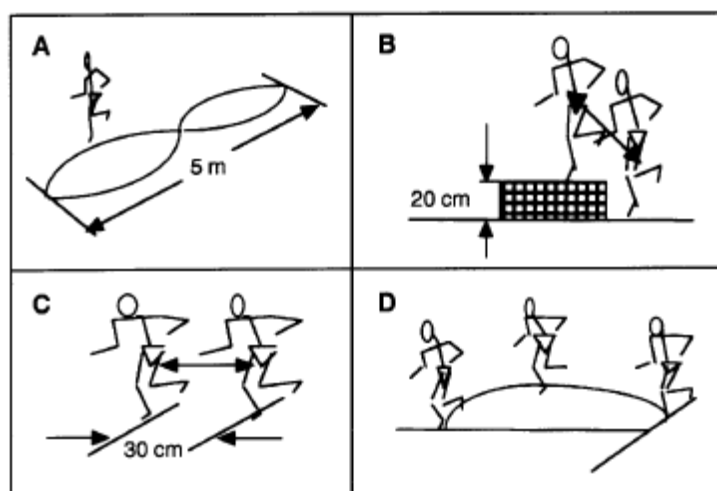


Figure 5-1. Functional Ability Hopping Tests (Itoh et al., 1998)

A) In the figure-of-eight hop test, the subjects are required to hop in a figure of eight over a 5-m distance as fast as possible, and the time taken to perform two consecutive laps will be measured.

B) In the up-down hop test, the subjects are required to hop vertically up and down a 20-cm-high step for 10 repetitions as quickly as possible without turning backward, and the time taken to perform the test will be measured.

C) In the side hop test, the subjects are required to hop transversely more than 30 cm for 10 repetitions as quickly as possible, and the time taken to perform the test was measured.

D) In the single hop test, the subjects are required to hop forward as far as they can and the distance covered will be measured.

The battery of functional tests was implemented on both legs to assess adequate levels of symmetry and functional ability for the previously injured limb. Subjects were required to demonstrate values on the previously injured leg within 10% of the values on the non-injured leg, on at least two of the tests. This 10% guideline is well within normal symmetry values reported for ACLr subjects (Gustavsson et al., 2006). Utilising this screening protocol several potential ACLr subjects were excluded from participation.

### 5.3.3 Experimental set-up and testing protocol

See Section 3.3.2 as an identical experimental set-up was utilized in the present investigation was used for the investigation reported in Chapter 3.

Following subject screening and prior to initiation of the testing protocol each subject completed an informed consent form and pre-test questionnaire to screen further for any potential health risks. Non-injured control subjects with no history of ACL injury, completed an injury history and sporting experience questionnaire to ensure all subjects were injury free and adequately experienced in their sport.

The testing protocol was then completed as outlined in Chapter 3. Following a static trial, subjects completed a minimum of 20 trials of the dynamic task with a random cutting direction. Ten successful trials in each direction were required from each subject. Successful

trials required the subject to run in the correct cutting direction in the direction of the visual cue, through the mapped out pathway orientated at 45°. Both feet were also required to land on their respective force plates during the jump land as was the appropriate cutting leg (e.g. the right leg when cutting in left direction) during the cut. Subjects received 1 minute rest between trials (Pollard et al., 2004; Pollard et al., 2005; Sigward and Powers, 2006) to prevent the potential effects of fatigue.

#### 5.3.4 Data reduction

Data reduction procedures were performed on data for all subjects as outlined in Section 3.3.8, Chapter 3 and shown in Figure 5-2. Various discrete measures were calculated during the different phases of the task (i.e. the first 40ms of the land, the entire land, the entire cut). During the landing phase touchdown (TD), peak angles, peak angular excursion range of motion (ROMmax), and touchdown and peak moments of the hip, knee and ankle were reported in all three planes. During the first 40ms of the land and the entire cutting phase peak angles, angular range of motion (ROM), and peak moments of the hip, knee and ankle were reported in all three planes. Ankle joint angles and moments were limited to flex-extension and inv-eversion.

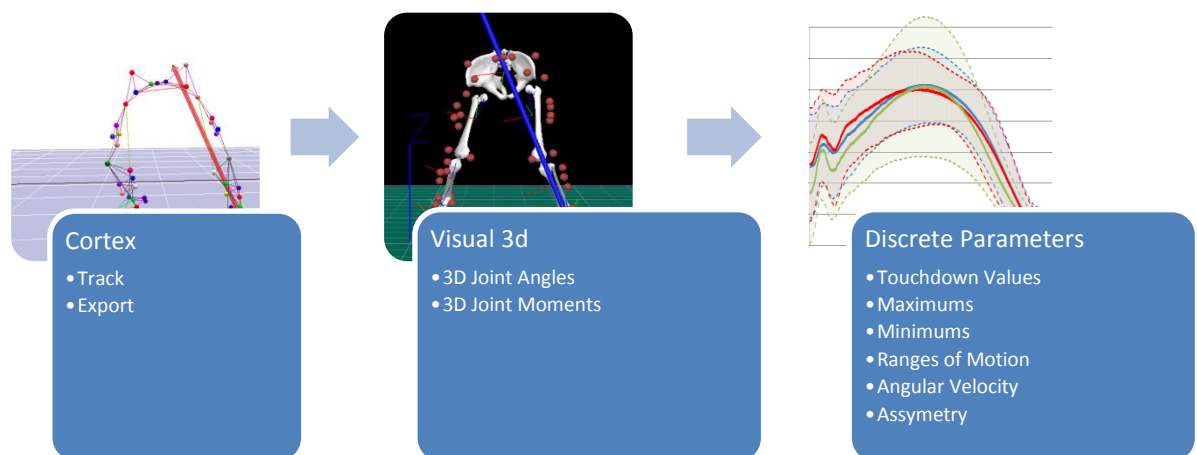


Figure 5-2. Main steps in data reduction process

#### 5.3.5 Statistical Analysis

Differences between ACLr and nACL subjects (ACLR previously injured leg –V- dominance matched nACL leg), and within ACLr subjects (previously injured ACLr leg –V- contralateral injury free ACLr leg) for the each of the above discrete measures for hip, knee and ankle joint angles and moments, were compared using a repeated measures ANOVA. Leg and trial were within-subject factors; and ACL injury status (ACLR or nACL) was the between-subject factor. These methods were utilised for the data from first 40 ms of the land, the entire land stance phase and the entire cut stance phase. Significance for all tests was set at  $p < .05$ , except in

cases where a moderate - large effect size was present with a p-value <0.1, in such cases significance was set at  $p < 0.1$ . All statistical analyses were performed using SPSS (PASW v18.0, IBM Inc. Armonk, NY). Partial eta<sup>2</sup> ( $\eta^2$ ) was also reported as a measure of effect size. It was calculated using the formula:  $\eta^2 = SS_{\text{effect}} / (SS_{\text{effect}} + SS_{\text{error}})$ , where  $SS_{\text{effect}}$  = effect variance and  $SS_{\text{error}}$  = error variance. Interpretation of effect size was based on the scale for effect size classification of Hopkins (2000). This scale is based on  $f$ -values for effect size and these were converted to  $\eta^2$  using the formula:  $f = (\eta^2 / (1 - \eta^2))^{0.5}$ . Consequently, the scale for classification of  $\eta^2$  was < 0.04 = trivial, 0.041 to 0.249 = small, 0.25 to 0.549 = medium, 0.55 to 0.799 = large, and >0.8 = very large.

## 5.4 Results

### 5.4.1 Jump Landing

Table 5-2 presents a summary of the significant differences present at the hip and knee joint in the frontal and transverse plane joint angles during landing.

Table 5-2. Significant differences in 3D Joint Angles of Hip, Ankle and Knee between ACLr and nACL subjects during landing.							
Average angles (°) for the previously injured leg of ACLr subjects (PI), and the dominance matched control leg of nACL subjects (C) are shown. Group differences (°), Partial eta <sup>2</sup> ( $\eta^2$ ) and p-values are also presented. The scale for classification of $\eta^2$ was < 0.04 = trivial, 0.041 to 0.249 = small, 0.25 to 0.549 = medium, 0.55 to 0.799 = large, and >0.8 = very large. PI=ACLR previously injured leg, NI=ACLR non-injured leg, C=nACL dominance matched control leg							
Jump Land		PI V NI	PI V C				
	1st 40ms		PI	C	Diff	$\eta^2$	p-value
Hip	Max Flexion	NO SIGNIFICANT DIFFERENCES	25.91	15.71	10.20	0.250	0.003
	Min Flexion	NO SIGNIFICANT DIFFERENCES	21.03	11.67	9.35	0.237	0.003
Knee	Ext-Internal Rotation ROM	NO SIGNIFICANT DIFFERENCES	5.04	3.46	1.58	0.131	0.035
Stance Phase			PI	C	Diff (°)	$\eta^2$	p-value
Hip	Touchdown Flexion	NO SIGNIFICANT DIFFERENCES	21.06	11.67	9.38	0.239	0.003
	Max Flexion	NO SIGNIFICANT DIFFERENCES	44.70	31.23	13.47	0.27	0.002
	Min Flexion	NO SIGNIFICANT DIFFERENCES	20.94	11.20	9.74	0.265	0.002
	Abduction-Adduction ROMmax	NO SIGNIFICANT DIFFERENCES	5.42	4.04	1.39	0.172	0.015
Knee	Ext-Internal Rotation ROMmax	NO SIGNIFICANT DIFFERENCES	9.31	6.94	2.34	0.144	0.027
Ankle	Max Dorsi-Flexion	NO SIGNIFICANT DIFFERENCES	89.41	94.74	5.32	0.145	0.026

No differences were found within ACLr subjects i.e. between previously injured (PI) and non-injured (NI) ACLr legs. Differences between the previously injured leg of ACLr subjects and the dominance matched leg on the nACL subjects (control leg) were found at the hip, and

knee joints during the initial 40 ms and the hip, knee and ankle joints during the entire stance phase of the land. Control subjects landed with a more extended hip at touchdown and throughout the landing phase, and also had more frontal plane ROM than ACLr subjects when compared the control leg was compared to the PI leg. At the knee the PI leg had increased ext-internal rotation ROM, during the first 40 ms and the entire landing phase when compared to the control leg. At the ankle the PI leg had less dorsi-flexion during the landing phase when compared to the control leg. Figure 5-3 presents individual knee frontal plane ROM for each subject to allow comparison with the work of Paterno et al., (2010). The frontal plane ROM values for each subject consist of ROMs in both the abduction (+ive) and adduction (-ive) direction, The PI, NI and control leg groups had 14, 11 and 7 legs in abduction during an average of 12, 12, and 8 trials respectively. The average abduction ROM for these PI, NI and control leg group trials was 6.46°, 6.52° and 4.55° respectively

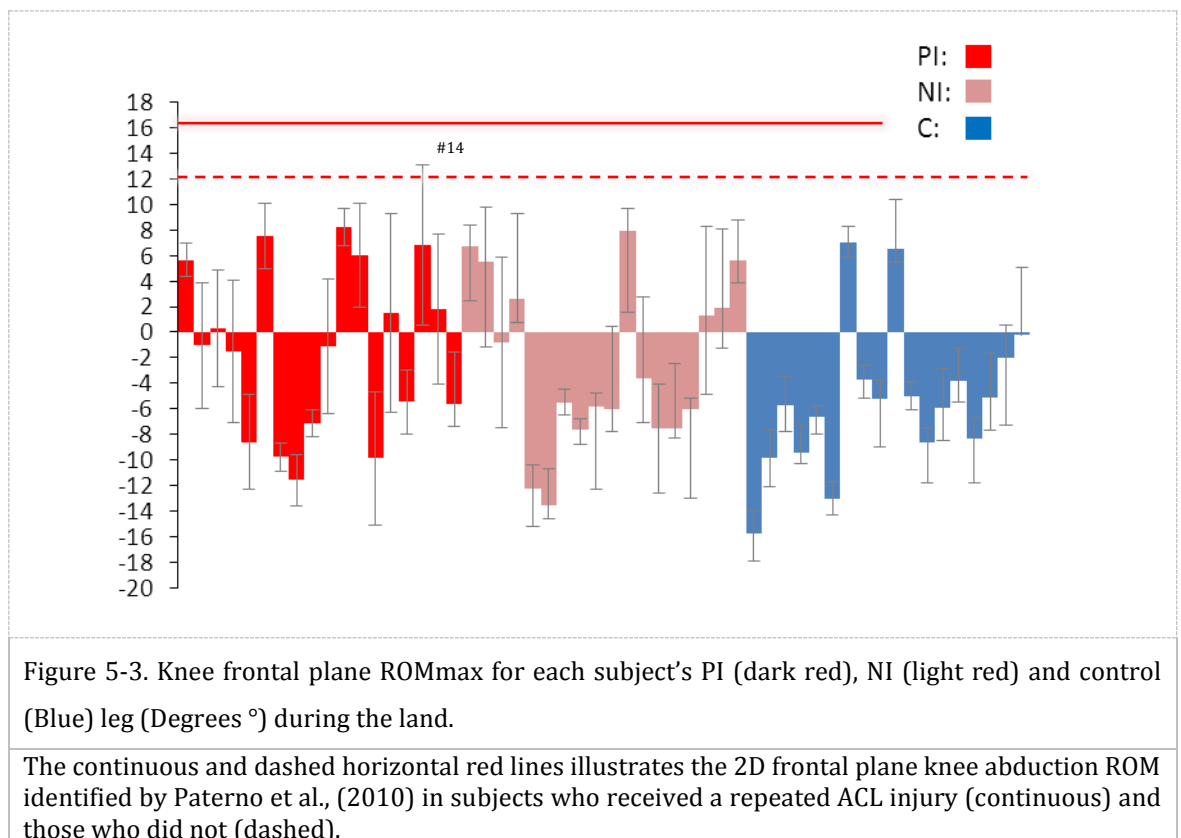


Figure 5-4 presents the 3D joint angles over the entire stance phase of landing for the hip, knee and ankle joint rotations that had significant group differences.



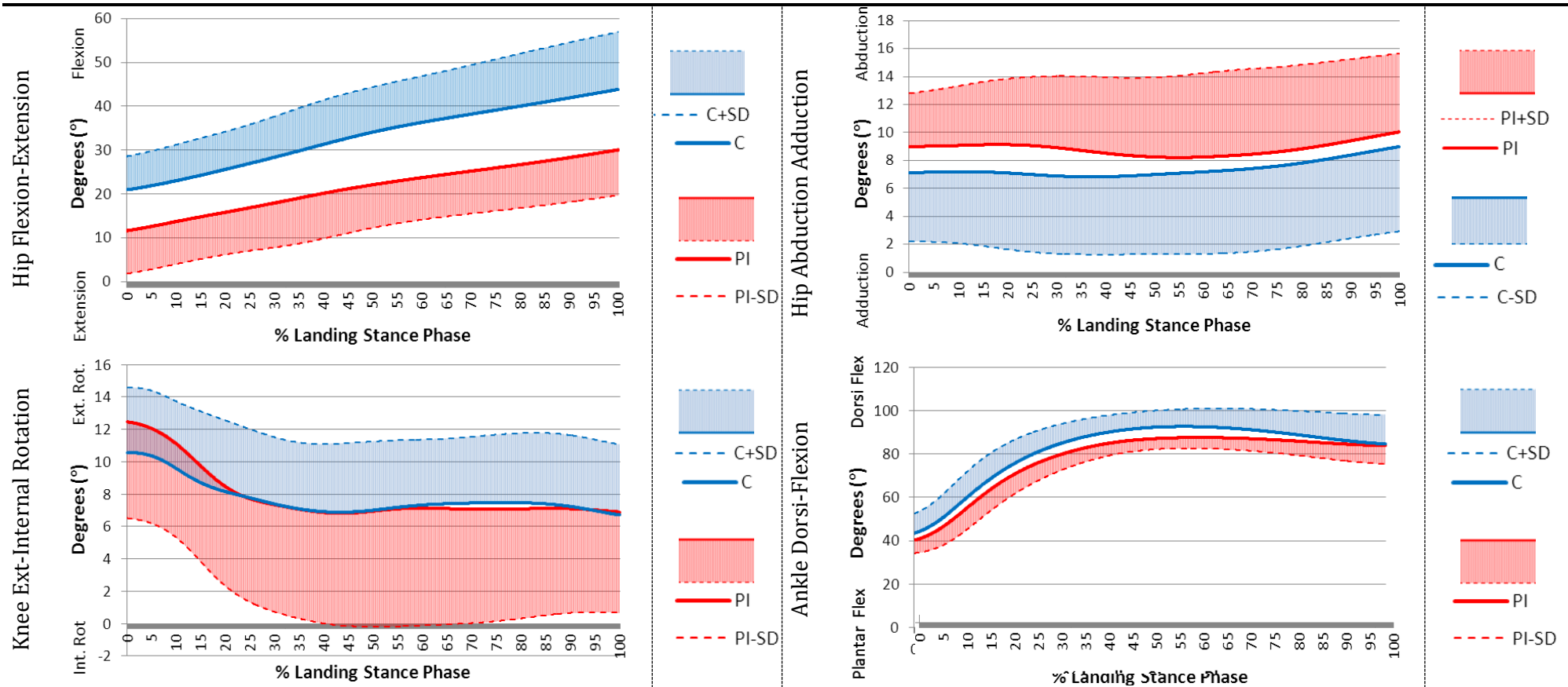


Figure 5-4. Joint angles of hip, knee and ankle for PI and control with significant differences between the previously injured leg of the ACLr subjects the healthy control leg of nACL subjects during landing.

Table 5-3 presents a summary of the significant differences present at the hip, knee and ankle joint in the 3D joint moments during landing.

Table 5-3. Significant differences in 3D Joint Moments of Ankle and Knee within ACLr subjects during landing.											
Average moments (Nm/kg.m) for the previously injured leg of ACLr subjects (PI), and the non-injured leg of ACLr subjects (NI) are shown. Group differences ( $\eta^2$ ), Partial eta <sup>2</sup> ( $\eta p^2$ ) and p-values are also presented. The scale for classification of $\eta p^2$ was < 0.04 = <b>trivial</b> , 0.041 to 0.249 = <b>small</b> , 0.25 to 0.549 = <b>medium</b> , 0.55 to 0.799 = <b>large</b> , and >0.8 = <b>very large</b> . PI=AClr previously injured leg, NI=AClr non-injured leg, C=nACL dominance matched control leg											
Jump Land		PI V NI					PI V C				
1st 40ms		PI	NI	Diff	$\eta p^2$	p-value	PI	C	Diff	$\eta p^2$	p-value
Knee	Max Extension	NO SIGNIFICANT DIFFERENCES					0.850	<b>1.02</b>	0.172	<b>0.180</b>	0.012
	Stance Phase	PI	NI	Diff	$\eta p^2$	p-value	PI	C	Diff	$\eta p^2$	p-value
Ankle	Max Plantar Flexion	NO SIGNIFICANT DIFFERENCES					-0.942	<b>-1.10</b>	0.154	<b>0.118</b>	0.047

No differences in 3D joint moments were shown between the PI leg and NI leg of the ACLr subjects during landing. In the initial 40ms of the land the PI leg had decreased extension moment providing decreased resistance to flexion at the knee. When the entire landing phase was investigated the PI leg also had decreased ankle plantar flexion moments. This provided decreased resistance to ankle dorsi-flexion when compared to the control leg. Figure 5-5 presents the 3D joint moments over the entire stance phase of the landing component for the knee and ankle joint moments that had significant group differences.

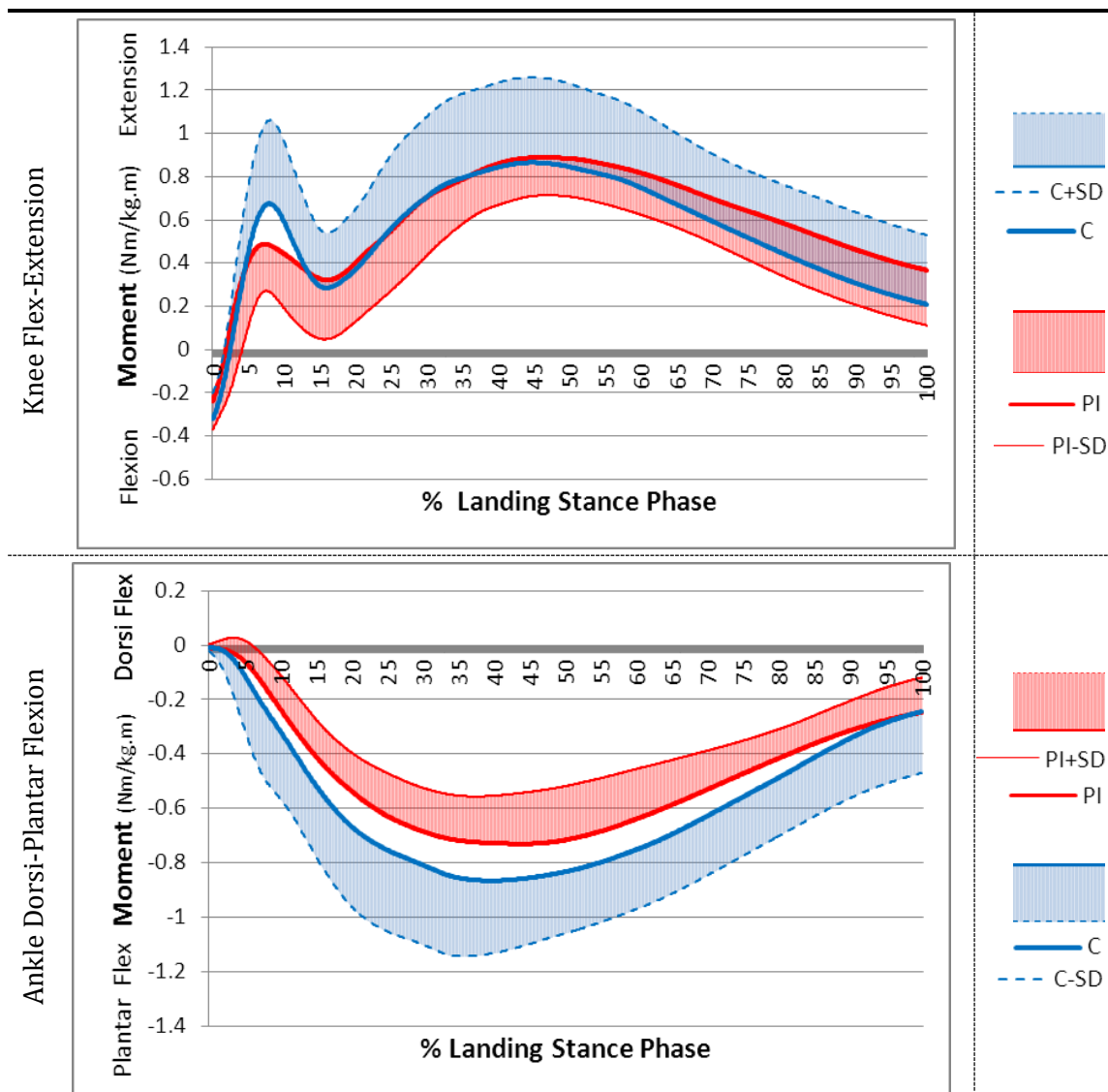


Figure 5-5. Joint moments of knee and ankle for the PI and control legs with significant group differences over the landing component of the task.

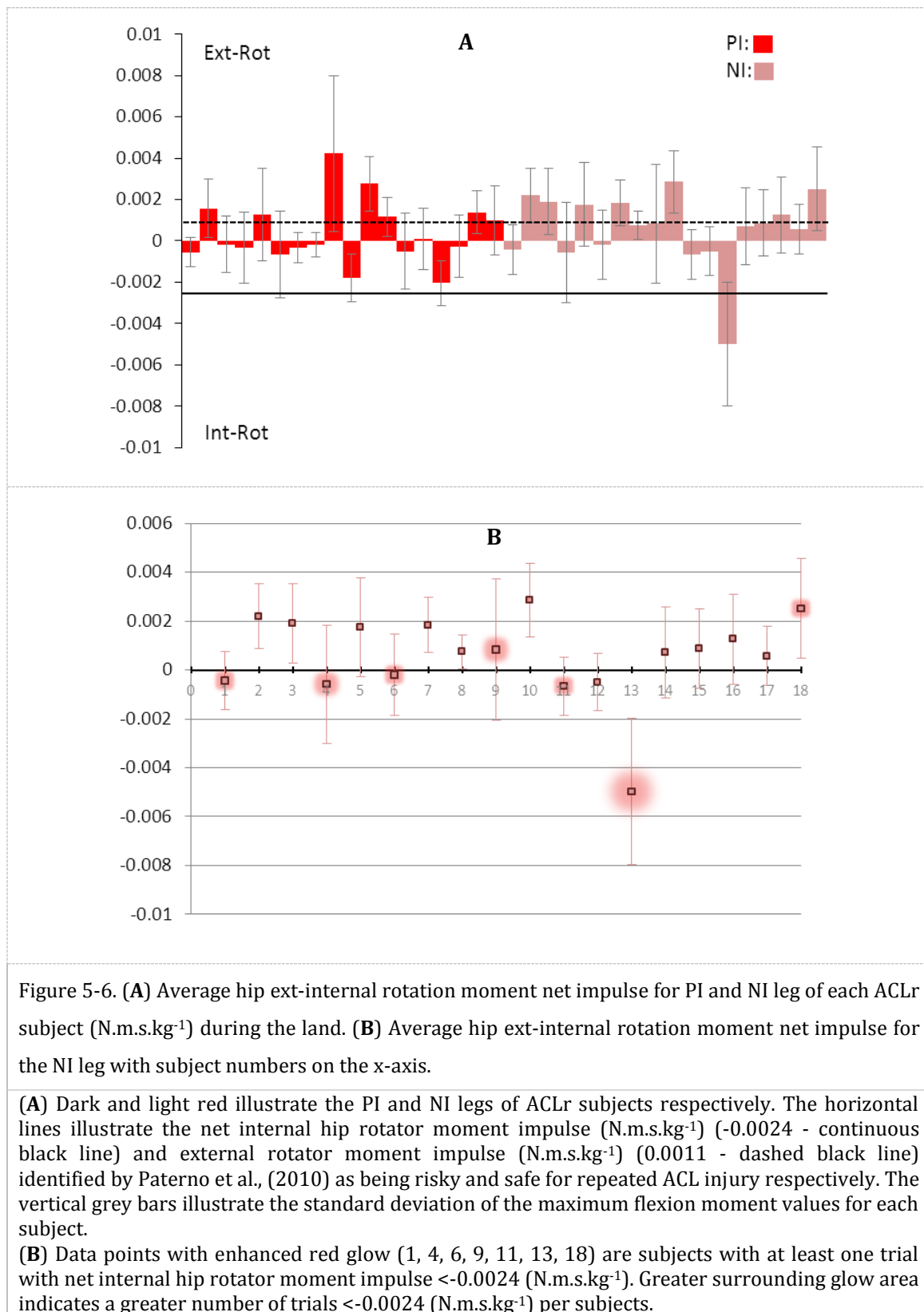
As a post analysis, asymmetry of initial contact knee flex-extension moment (inter-leg differences) and net hip transverse plane moment impulse (area under moment time series curve) were reported in Table 5-4 and Figure 5-6 to allow evaluation alongside the work of Paterno et al., (2010) . There was no significant difference and a trivial effect size for differences in asymmetry of initial contact knee flex-extension moment between the ACLr and nACL subjects. There was also no difference in net hip transverse plane moment impulse between PI and NI legs. On closer inspection of Figure 5-6 there are seven NI legs with a least one trial in the risk zone for repeated ACL injury according to Paterno et al., (2010) (average no. trials /person =  $5 \pm 6$ ). Subject #13 was identified as especially at risk with an average net internal hip rotator moment impulse  $< -0.0024(\text{N.m.s.kg}^{-1})$  with 17/20 trials  $< -0.0024(\text{N.m.s.kg}^{-1})$ .

Table 5-4. Average asymmetry of knee flex-extension moment, ( $\text{N.m.s.kg}^{-1}$ ) and net impulse of

the transverse plane hip moment (N.m.s.kg<sup>-1</sup>) during the first 40ms of landing.

Average knee flex-extension moment differences (Nm/kg.m), for ACLr subjects (PI-NI), and the nACL subjects (control-contralateral control leg) are shown. Average net impulse for ACLr subjects PI and NI leg are shown. Partial eta2 ( $\eta^2$ ) and p-values are also presented. The scale for classification of  $\eta^2$  was < 0.04 = **trivial**, 0.041 to 0.249 = **small**, 0.25 to 0.549 = **medium**, 0.55 to 0.799 = **large**, and >0.8 = **very large**. PI=ACLR previously injured leg, NI=ACLR non-injured leg, C=nACL dominance matched control leg

Jump Land		Dominant Leg			
1st 40ms	ACLR	nACL	Diff	$\eta^2$	p-value
Average asymmetry of knee flex-extension moment	0.125	0.125	0.0002	<b>0.000</b>	0.990
	PI	NI	Diff	$\eta^2$	p-value
Average hip ext-internal rotation moment net impulse	3.7 x10 <sup>-4</sup>	5.92 x10 <sup>-4</sup>	2.22 x10 <sup>-4</sup>	<b>0.009</b>	0.702



## 5.4.2 Cutting Manoeuvre

Table 5-5 presents a summary of the significant differences present at the hip, knee and ankle joint in the 3D joint angles during cutting.

Table 5-5. Significant differences in 3D Joint Angles of Hip, Ankle and Knee between ACLr and nACL subjects during cutting.											
Average angles (°) for the previously injured leg of ACLr subjects (PI), the non-injured leg of ACLr subjects (NI) and the dominance matched control leg of nACL subjects (C) are shown. Group differences (°), Partial eta2 ( $\eta$ p2) and p-values are also presented. The scale for classification of $\eta$ p2 was < 0.04 = trivial, 0.041 to 0.249 = small, 0.25 to 0.549 = medium, 0.55 to 0.799 = large, and >0.8 = very large. PI=AClr previously injured leg, NI=AClr non-injured leg, C=nACL dominance matched control leg											
Cut		PI V NI					PI V C				
		PI	NI	Diff (°)	$\eta$ p2	p-value	PI	C	Diff (°)	$\eta$ p2	p-value
Hip	Max Flexion	NO SIGNIFICANT DIFFERENCES					55.23	36.77	18.47	0.363	<0.001
	Min Flexion						22.03	4.23	17.80	0.404	<0.001
	Int-External Rotation ROM						17.85	22.85	5.00	0.191	0.01
Knee	Max Flexion	NO SIGNIFICANT DIFFERENCES					-37.09	-29.84	7.25	0.121	0.044
Ankle	Max Dorsi Flexion	NO SIGNIFICANT DIFFERENCES					91.42	98.21	6.79	0.218	0.005
	Dorsi-Flexion ROM	31.31	35.48	4.17	0.279	0.024	31.31	43.06	11.75	0.401	<0.001

The majority of differences were reported between the PI leg and the nACL control leg during the cutting component of the manoeuvre. Dorsi-flexion at the ankle on the previously injured also showed differences when compared with the NI leg and the control leg. The PI leg had less maximum dorsi-flexion the control leg. The PI leg had less dorsi-flexion ROM than both the NI and control legs. This dorsi-flexion ROM demonstrated in the PI leg was however greater than that demonstrated on the NI leg. At the knee the previously injured leg had more knee flexion than the control leg. At the hip the PI leg was more flexed throughout the cutting component similar to the landing component of the task. The hip of the PI leg also had less transverse plane ROM compared to the control leg. Figure 5-7 presents the 3D joint angles over the entire stance phase of the cutting component of the task for the hip, knee and ankle joint rotations that had significant group differences.

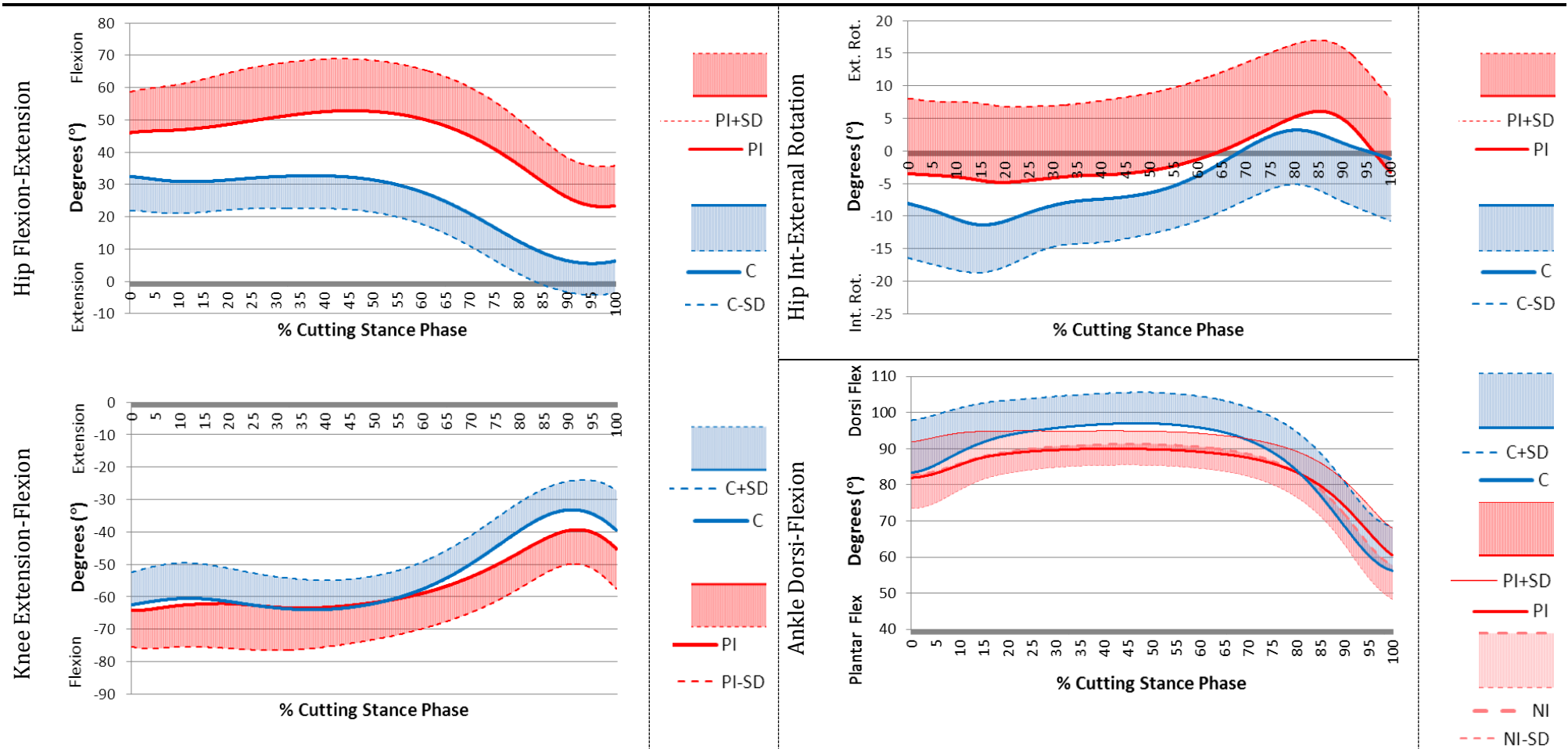


Figure 5-7. Joint angles of hip, knee and ankle for PI, NI and control leg with significant differences between the previously injured leg of the ACLr subjects the contralateral ACLr leg and healthy control leg during landing.

Table 5-6 presents a summary of the significant differences present at the hip, knee and ankle joint in the 3D joint moments during cutting.

Table 5-6. Significant differences in 3D Joint Moments of Hip, Ankle and Knee within ACLr subjects during cutting.							
Average moments (Nm/kg.m) for the previously injured leg of ACLr subjects (PI), and the non-injured leg of ACLr subjects (NI) are shown. Group differences ( $^{\circ}$ ), Partial eta <sup>2</sup> ( $\eta p^2$ ) and p-values are also presented. The scale for classification of $\eta p^2$ was < 0.04 = <b>trivial</b> , 0.041 to 0.249 = <b>small</b> , 0.25 to 0.549 = <b>medium</b> , 0.55 to 0.799 = <b>large</b> , and >0.8 = <b>very large</b> PI=AClr previously injured leg, NI=AClr non-injured leg, C=nACL dominance matched control leg							
Cut		PI V NI	PI V C				
			PI	C	Diff	$\eta p^2$	p-value
Hip	Max Extension (Nm/kg.m)	NO SIGNIFICANT DIFFERENCES	0.591	<b>0.762</b>	0.172	<b>0.158</b>	0.02
Knee	Abduction (Nm/kg.m)	NO SIGNIFICANT DIFFERENCES	<b>0.303</b>	0.232	0.071	<b>0.135</b>	0.32
Ankle	Max Dorsi-Plantar Flexion (Nm/kg.m)	NO SIGNIFICANT DIFFERENCES	-0.004	<b>0.008</b>	0.022	<b>0.232</b>	0.004
	Min Dorsi-Plantar Flexion (Nm/kg.m)	NO SIGNIFICANT DIFFERENCES	-0.935	<b>-1.24</b>	0.306	<b>0.246</b>	0.003

No differences in 3D joint moments were shown between the PI leg and NI leg of the ACLr subjects. At the hip the PI leg of the ACLr individuals had decreased extension moment, providing decreased resistance to hip flexion when compared to that of the nACL control leg. At the knee the PI leg of the ACLr individuals had increased abduction moment, providing increased resistance to knee adduction when compared to that of the nACL control leg. At the ankle the PI leg of the ACLr individuals had decreased maximum and minimum dorsi-plantar flexion moment providing decreased resistance to ankle plantar and dorsi-flexion when compared to that of the control leg. Figure 5-8 presents the 3D joint moments over the entire stance phase of the cutting component of the task for the hip knee and ankle joint moments that had significant group differences.



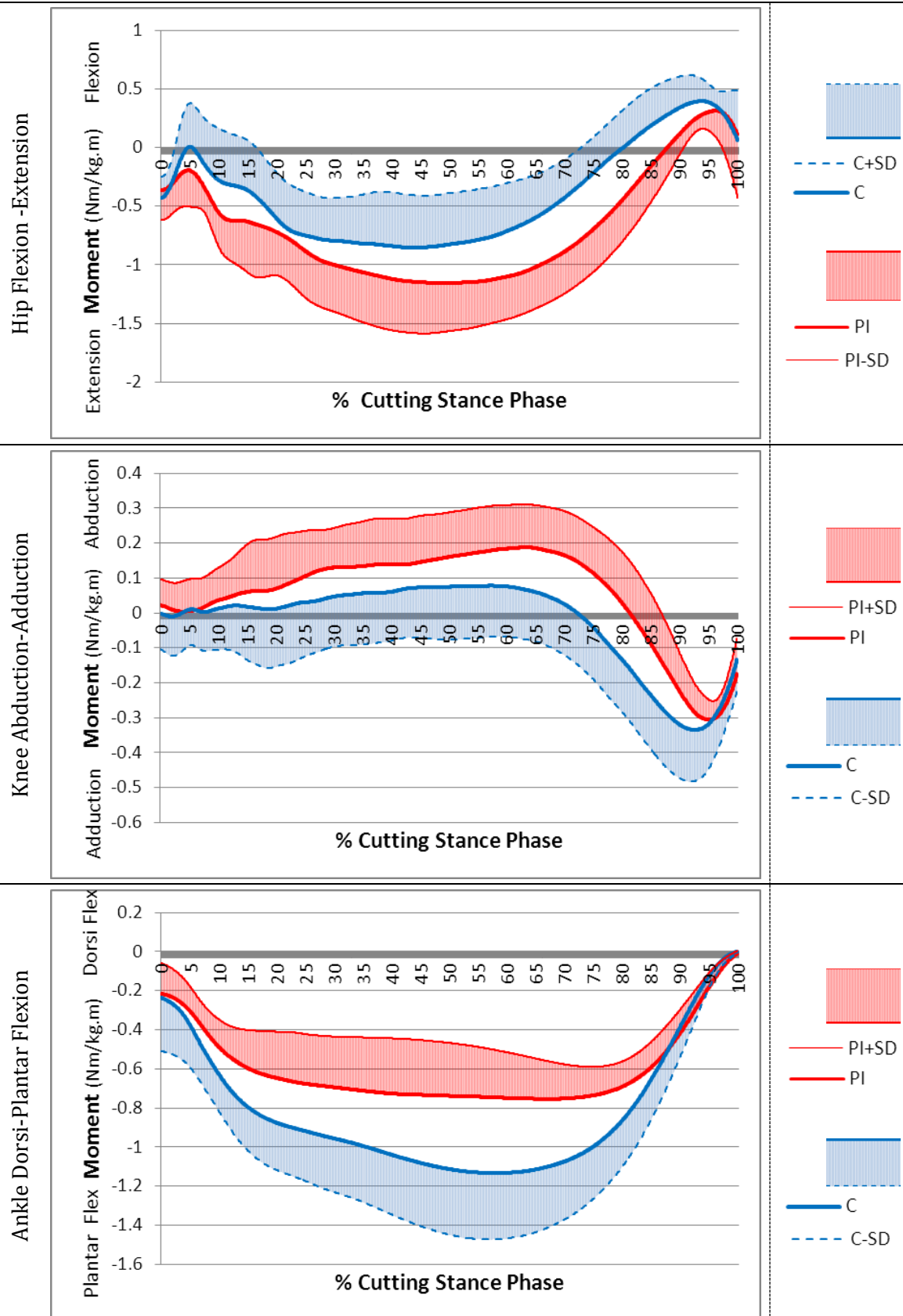


Figure 5-8. Joint moments for the PI and control legs with significant group differences during the cutting component of the task.

## 5.5 Discussion

In the present study, lower limb mechanics were assessed in the previously injured (PI) leg of ACLr subjects and compared against an internal control of the contralateral non-injured (NI) leg and an external control of a healthy control leg from a matched control subject. The surgical reconstruction and rehabilitation following an ACL injury aims to return individuals to full competitive participation in their sport, therefore, assessment of the performance of these individuals in tasks that replicate competitive sporting demands is essential. Both landing from a jump and changes of direction such as side-cutting have been previously identified as high risk movements that occur regularly in match situations. The assessment of ACLr individual's performance of the utilized drop-jump land and cut manoeuvre is therefore beneficial, providing new information on lower limb mechanics that may predispose ACLr individuals to repeated ACL injury and the development of OA.

### 5.5.1 *Altered joint mechanics in ACLr subjects*

When the lower limb biomechanics of the ACLr subject's previously injured (PI) leg were compared to the nACL subjects control leg, several differences were reported at all joints during all phases of the task. At the hip, previous research has only reported sagittal plane kinematic differences between previously injured leg of ACLr subjects and a healthy control. The present investigation however has reported differences in all three planes between the PI and control leg hips. In terms of the sagittal plane the more extended hip reported in previous work (Decker et al., 2002) was shown in the control group of the current investigation, with the previously injured leg performing and landing and cutting manoeuvre with a more flexed hip posture. Decker et al.'s, (2002) investigation utilized a drop vertical jump. It is plausible that the added demands of the task utilized in the current investigation elicited an alternative response in the ACLr subjects. Poor neuromuscular control of the trunk has been linked with the increased risk of lower limb injuries (Bullock-Saxton et al., 1994; Beckman and Buchanan, 1995; Devlin, 2000; Beynnon et al., 2006). Zazulak et al., (2007a; 2007b) have identified deficits in neuromuscular control of the trunk and core proprioception as predictors of knee and ACL injury in female athletes. Paterno et al., (2010) also reported that decreased postural stability predicted repeated ACL injuries. It is plausible that the increased flexion at the hips of the ACLr subjects may originate from deficits in trunk control which may increase risk of repeated ACL injury.

Frontal plane hip joint kinematics and kinetics have received limited if any research attention in an ACLr population. The PI leg had increased frontal plane ROMmax in the frontal plane when compared to the control leg during landing. The average difference between groups was  $\sim 1^\circ$  and with a small effect size this difference was not considered large enough

to merit consideration as a factor to increase either the risk of re-injury or the development of OA. The PI leg of the ACLr subjects also showed different transverse plane hip kinematics in comparison the control leg of the nACL subjects, during the cutting component of the task. The PI leg of the ACLr subjects had less transverse plane ROM than the control leg during the cutting movement. Altered transverse plane hip kinematics have not been previously demonstrated in ACLr subjects, when compared to nACL control subjects. The difference in ROM between the groups however is small at  $\sim 5^\circ$  with a small effect size. It is still plausible that the ACLr subjects may have performed the task differently in order to attempt to protect the ACL by limiting proximal transverse plane hip movement.

The PI leg of the ACLr subject was hypothesized to show several differences at the knee joint when compared with the control leg. None of the previous investigations utilizing semi-demanding tasks (stair descent and pivot, landing) identified differences in sagittal plane knee kinematics i.e. flexion extension (Kowalk et al., 1997; Decker et al., 2002) between the PI leg of ACLr subjects and a control. This was partially replicated in the current investigation with no differences in sagittal plane knee kinematics when compared to a control leg during landing. This is in contrast to previous investigations utilising landing from a horizontal hop, that have reported decreased touchdown knee flexion (Deneweth et al., 2010) and sagittal plane ROM in the PI leg when compared to the control leg during landing (Orishimo et al., 2010). Both legs of the ACLr subjects and the control leg had average knee flexion angles greater than  $20^\circ$  at initial contact (PI=  $-24.40 \pm 5.91$ , NI=  $-23.10 \pm 4.64$ , Control=  $-22.10 \pm 5.12$ ) and throughout the landing stance phase. Paterno et al., (2010) did not find any link between decreased knee flexion and repeated ACL injury therefore, it is likely that any increased risk of repeated ACL injury is not due to decreased knee flexion during the movements similar to Paterno et al.'s, (2010) task or the drop-land component of the current task. The cutting component of the task in the present investigation did however produce knee sagittal plane kinematics that differentiated between the PI and control legs. The peak in knee extension that occurs towards the end of the cutting movement (Figure 5-7) is smaller on the PI leg when compared to the nACL control leg. This translates to smaller minimum flexion or a more extended knee in the control leg. The knee flexion angles of both the PI and control legs are greater than  $20^\circ$  which indicates limited risk of ACL injury due to this increased extension in the control leg during cutting.

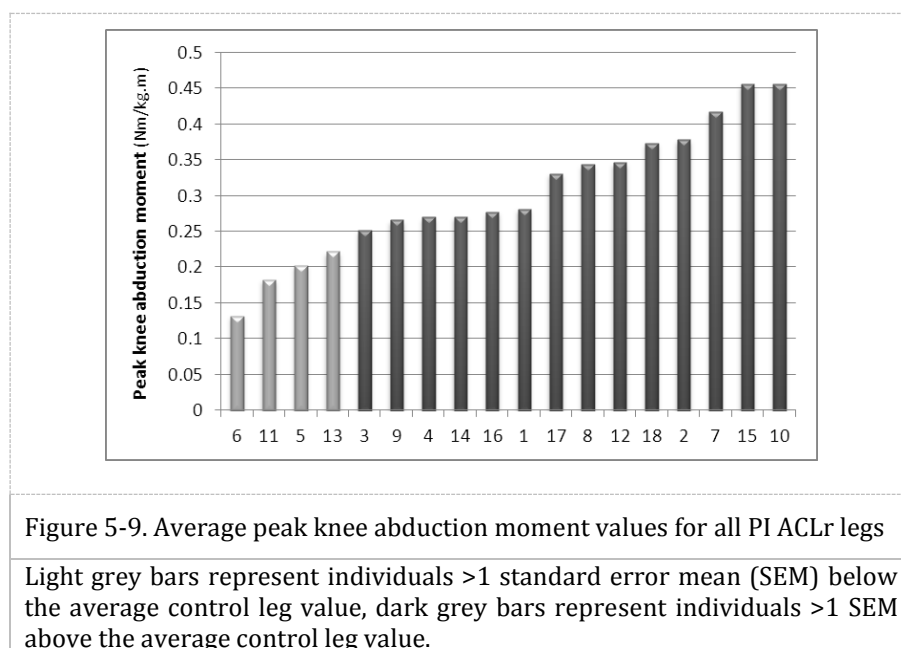
Regarding knee joint moments in the sagittal plane, previous investigations utilising walking, jogging and jog and cut tasks have reported reductions in external knee flexion moment in the PI leg when compared to a healthy control (Timoney et al., 1993; Bush-Joseph et al., 2001). Based on these previous findings we hypothesized that the PI leg would demonstrate increased internal extension moment providing increased resistance to external

flexion moment when compared to the control leg. This was not the case with the PI leg demonstrating less knee extension moment when compared to the control leg during the initial 40 ms of the land. This decreased knee extension moment would provide decreased resistance to knee flexion when compared to the nACL control leg during this period. No differences in sagittal plane knee moments were reported during the entire landing or cutting stance phase. This decreased resistance to knee flexion in the ACLr subjects is not likely to increase risk of repeated ACL injury as knee flexion moment is associated with decreased ACL strain (Draganich and Vahey, 1990). A substantial difference between these previous investigations specifically the Bush et al., (2001) investigation and the present one is the movement task utilised. The task in the present investigation is unanticipated and requires maximal exertion. The landing component of the current task is also not present in that of Bush et al., (2001), who reported these sagittal plane knee joint moments during a jog and cut task. It is likely that the demands of landing from a maximal drop jump may illicit different knee joint kinetics than that reported by Bush et al., (2001) during a jog and cut task. There was no significant difference in bilateral asymmetry in knee sagittal plane moment between the ACLr and nACL subjects. This measure among others was used to predict repeated ACL injury (Paterno et al., 2010) which may indicate that the ACLr individuals in this population at less risk of re-injury. The drop-vertical jump task utilized by Paterno et al., (2010) does not have the maximal and unanticipated nature of the task utilized in the present investigation, and makes injury prediction difficult for the present population based on these previous findings.

In the frontal plane, increased knee abduction and adduction have been associated with the PI leg when compared to a control leg. Abduction ROM of  $\sim 16.2^\circ$  at the knee have been reported in ACLr individuals who suffer a repeated ACL injury (Paterno et al., 2010). The knee abduction angle measured in Paterno et al.'s, (2010) investigation was a 2D measure and therefore, application of its findings to the three-dimensional knee abduction angles measured in the present investigation is difficult. Similar measures of 2D knee abduction have however been reported to be consistent with those reported in previous investigations utilising similar tasks (McLean et al., 2005). The work of Paterno et al., (2010) however reports values that are larger than those previously reported in a similar task, which were proposed to increase risk of initial ACL injury in females ( $9^\circ$ ) (Hewett et al., 2005). There were no differences between the 3D knee abduction ROM values reported in the PI or control legs of the present investigation during the landing or cutting component of the task. The average frontal plane ROMmax values included both adduction and abduction ROM. The average of the abduction ROM values for each group during the landing component of the task (PI= $7.23^\circ$ , NI= $7.47^\circ$ , control= $7.25^\circ$ ) were less than what was reported in Paterno et al., (2010), and below what would increase risk of re-injury according to both Paterno et al.,

(2010) and Hewett et al., (2005). There were no individuals that demonstrated knee abduction ROMmax values that would indicate risk of repeated ACL injury as shown in Figure 5-3 according to the values reported by Paterno et al., (2010). However, assuming that these values are an overestimate of a 3D knee abduction value, one subject in the PI group (#14) may be at increased risk of repeated ACL injury due to increased knee abduction ROM during landing.

Increased internal abduction moment at the knee has been suggested as a predictor of OA at the knee, and has been demonstrated in ACLr subjects during walking gait (Butler et al., 2009) when compared to a matched control group. The present investigation reported similar findings with the PI leg of the ACLr subjects demonstrating increased internal knee abduction moment when compared with the nACL control leg during the cutting component of the task. The abduction moment values reported for the ACLr and nACL subjects in the present investigation (Figure 5-9) are within the range of values reported by Butler et al., (2009) ( $0.36 \pm 0.08$  Nm/kg.m). The PI leg showed an 11% larger peak knee adduction moment than the control leg, with a small effect size. This suggests that larger internal knee abduction moments are present in ACLr individuals post reconstruction, rehabilitation and return to sport, during low and high impact tasks. These larger internal knee abduction moments may therefore, be a contributor to the development of OA in an ACLr population.



External knee abduction moment on the other hand has been reported as a risk factor for initial ACL injury. Therefore, it was hypothesized that it may also be a risk factor for repeated ACL injury in the current ACLr population, with ACLr subjects demonstrating decreased internal knee adduction moments. This was not the case, with no differences in knee adduction moment between PI and control legs. Paterno et al., (2010) did not report any

association between external knee abduction moment and repeated ACL injury therefore, it is likely that this measure may not differentiate ACLr individuals who are at risk of re-injury.

Increased transverse plane range of motion has been demonstrated at the knee in ACLr subjects, during low impact stair descent and pivot (Ristanis et al., 2003) and high impact land and pivot tasks (Ristanis et al., 2005) when compared to a healthy control. The present investigation had similar findings with increased transverse plane ROM and ROMmax on the PI leg when compared to the control leg during the initial 40 ms and the entire landing phase respectively. The values in the present investigation for transverse plane ROM are much smaller than that reported by Ristanis et al., (2003; 2005), (PI:9.31 Control:6.94 (Present Investigation) PI: 21.68 Control: 19.01 (Ristanis et al., 2003)). This may be due to differences in the utilized marker set between investigations, where Ristanis et al., (2003; 2005) utilized wand markers for the calculation of transverse plane rotations, also the pivot required by Ristanis et al., (2003; 2005) was 90° as opposed the 45° cutting angle of the present investigation. The wand marker set utilised by Ristanis et al., (2003; 2005) is considered a suboptimal method for estimating tibial rotation in comparison to the marker clusters utilised in the present investigation (Manal et al., 2000). The increased pivot/ cutting angle may also explain the increased transverse plane ROM values reported by Ristanis et al., (2003; 2005). The relative difference between the groups however is similar in both investigations, but with a small effect size in the present investigation.

Sagittal plane, ankle joint kinematics have been reported to differentiate ACLr individuals from controls (Kowalk et al., 1997; Decker et al., 2002). These differences have indicated increased energy absorption at the ankle joint during landing involving increased ROM and angular velocity. This was not seen in the present investigation with similar sagittal plane ankle kinematics in the PI, NI and control legs during impact phase of landing. Over the entire landing phase however the PI leg did have less plantar flexion moment at the ankle, providing less resistance to dorsi-flexion load, with a small effect size this had no effect on the rate of dorsi-flexion during the land. The PI leg did however have less dorsi-flexion but this did not result in any differences in dorsi-flexion ROM. The similar dorsi-flexion ROM and angular velocity between the PI, NI and control legs indicates similar energy absorption strategies. A number of differences were observed during the cutting component of the movement. The PI leg has less sagittal plane ROM than the NI and control leg, with a small and medium effect size respectively. This PI leg also had less maximal dorsi-flexion during the cut. These findings may indicate that the PI leg utilised the ankle joint less during the pushing off component of the cutting task. This is supported by the sagittal ankle joint moments which were lower in the PI leg when compared to the control leg throughout the cutting phase.

When the lower limb biomechanics of the ACLr subject's previously injured (PI) leg were compared to the biomechanics of the contralateral non-injured (NI) leg there were limited between leg differences. Hip transverse plane moment net impulse was compared between the PI and NI leg of ACLr subjects to allow comparison with the previous work of Paterno et al., (2010). This previous work utilized this measure to independently predict a second ACL injury. ACLr subjects with a net internal rotator moment impulse in the NI leg during the initial landing period were over 8 times more likely to sustain a repeated ACL injury. NI leg net internal rotator moment impulses of less than  $-0.0024$  (Nm\*s/kg) were reported as indicative of increased risk of re-injury. Seven subjects had at least one trial with the NI leg demonstrated a net internal rotator moment impulses of less than  $-0.0024$  (Nm\*s/kg). Six of these subjects did not have an average net internal rotator moment impulse less than  $-0.0024$  (Nm\*s/kg). Subject #13 as shown in Figure 5-6 had an average net internal rotator moment impulse of  $-0.00499$  (Nm\*s/kg) with 17 from 20 trials below  $-0.0024$  (Nm\*s/kg). Overall, the ACLr population, are not considered to be at risk of re-injury as identified by net internal rotator moment impulse on the NI leg. Subject #13 is considered to be at increased risk of re-injury however based on the findings of Paterno et al., (2010). The application of Paterno et al.'s, (2010) findings to the present results must be considered with caution given the different tasks performed. Further analysis of the drop-jump data from the current investigation are required before full application to Paterno et al.'s, (2010) findings.

Tashman et al., (2004) have reported increased knee adduction in the PI leg of ACLr individuals when compared to the NI leg. This was linked with the higher incidence and faster progression of knee OA (Cerejo et al., 2002). There were no such differences in knee adduction angles between the PI and NI leg in the present investigation with the PI, NI and control leg both reporting maximum adduction angles similar to that reported in the Tashman et al., (2004) investigation. Tashman et al., (2004) was the first to report differences in knee adduction between PI and NI legs via radiographic stereophotogrammetric analysis, allowing quantitative 3D motion assessment and visualization of bone motion. It is possible that the methods used in the current investigation (motion analysis via skin markers) was not sensitive enough to pick up the adduction differences identified by Tashman et al., (2004). Previous investigations utilising skin based markers, comparing the PI to the NI during single leg hopping (Deneweth et al., 2010) and to a control leg during gait (Butler et al., 2009; Scanlan et al., 2010) also failed to show differences in knee adduction. The larger internal knee abduction moments present in the PI leg when compared to the control leg of nACL subjects are present in both PI and NI legs with no within ACLr differences reported. This increased internal knee abduction moment may be predictive of future OA but if it was the sole contributor to OA development, the OA would affect both PI and NI legs, which is usually not the case.

Ristanis et al., (2003; 2005) found increases in PI knee transverse plane ranges of motion, when compared to both the NI leg and also the control leg, they therefore, concluded that the initial ACL injury caused the increased transverse plane ROM and that the surgical intervention and rehabilitation performed did not restore this to normal levels. A similar assumption cannot be drawn in the present investigation as there was no difference reported within the ACLr subjects in the present investigation i.e. both the PI and NI leg have similar levels of transverse plane knee ranges of motion. ACLr subjects are at an increased risk of OA development for the PI leg, therefore, given the present data it is unlikely to be due to increased transverse plane knee rotation alone. It may be the case that any knee joint damage from the initial ACL injury (e.g. meniscal damage), or reconstructive surgery on the PI leg in combination with the increased transverse plane knee ROM and other risk factors such as the knee abduction moment detailed above, may lead to this increased risk of OA development. The PI leg has also been previously reported to be more externally rotated when compared to the NI leg (Tashman et al., 2004; Scanlan et al., 2010), again this was not the case with both legs having similar degrees of knee external rotation throughout the landing and cutting phases. Transverse plane knee moments have also been previously reported to differ between legs of ACLr subjects (Tsarouhas et al., 2010); the NI leg was reported to have greater rotational knee moments than the PI leg. This was not the case in the present investigation with similar transverse plane internal knee moments between the PI, NI and control legs during the landing and cutting components of the task. It is likely that this asymmetry reported by Tsarouhas et al., (2010) may be due to a lack of confidence in the PI leg. Subjects in Tsarouhas et al.'s, (2010) investigation were on average ~13 months post follow-up, whereas the subjects in the present investigation were on average ~3 years post surgery. It is likely the ACLr subjects in the present investigation were more confident on their PI leg allowing them to complete a more demanding task with no such asymmetries in transverse plane knee moments.

At the ankle joint where the PI leg had less sagittal plane ROM than the NI leg. With a small effect size this 4.17° (12%) difference in range of motion is unlikely to induce sufficient loading to increase risk of degenerative joint disease or increase strain on the ACL and increase risk of re-injury.

It is clear from these results and their discussion that the main differences are present between the nACL and ACLr populations rather than between the PI and NI legs of the ACLr subjects. It can be assumed therefore, that the surgical and rehabilitation interventions were successful in allowing the ACLr subjects to regain similar lower limb biomechanics in both the PI and NI leg, but significant differences remain between the ACLr and nACL populations.



These altered lower-limb biomechanics characterized by the ACLr group may be risk factors for the development of OA and the occurrence of repeated ACL injury as previously outlined.

### 5.5.3 *Directions for future research*

Based on previous research identifying poor neuromuscular control of the trunk as a risk factors for knee and ACL injury and the increased hip flexion and limited trunk control suggested in the ACLr population of the current investigation, future research into this association in an ACLr population is merited.

Further investigations utilising tasks that replicate competitive sporting demands with an ACLr population are required. Limited differences were reported between PI and NI legs of the ACLr subjects, which may have been due to the overall bi-lateral nature of the task. A similarly demanding and match specific task of a more single leg nature will further explore any compensations present within the ACLr subjects PI and NI legs.

The differences present between the PI and control legs in knee abduction moment and transverse plane knee range of motion support previous investigations identifying these variables as risk factors for the development of OA. The similarity of these variables in the PI and NI legs of the ACLr subjects may in this population may indicate that the development of OA may not be solely attributed to these factors and that the structure and degradation of the PI knee joint from the initial injury and surgical reconstruction (e.g. meniscal damage and bone bruising) in combination with these joint mechanics may lead to OA development. Prospective studies investigating the contribution of these knee joint mechanics in combination with varying levels of initial joint damage is merited.

## 5.6 **Conclusions**

On the basis of the research outcomes obtained for the population tested, the following conclusions can be drawn:

- The PI leg of ACLr individuals performed a drop-jump land and cut task with similar hip, knee and ankle joint kinematics to that of the contralateral NI leg.
- ACLr subjects perform a drop-jump land and cut task with increased hip flexion when compared to a nACL control.
- The PI leg of ACLr subjects performs the cutting component of a drop-jump land and cut task with greater internal knee abduction moment than a nACL control leg.
- The PI leg of ACLr subjects performs the landing component of a drop-jump land and cut task with greater transverse plane range of motion at the knee than a nACL control leg.
- The PI leg of ACLr subjects performs the landing and cutting component of a drop-jump land and cut task with less maximum dorsiflexion than a nACL control leg.

- The PI leg of ACLr subjects performs the cutting component of a drop-jump land and cut task with less dorsi-flexion range of motion than the contralateral NI leg and a nACL control leg.

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## **CHAPTER 6**

### **THE ROLE OF PREVIOUS ANTERIOR CRUCIATE LIGAMENT INJURY ON LOWER LIMB JOINT AND SEGMENT COUPLING DURING A MATCH SPECIFIC LAND-CUT TASK**

## 6.1 Abstract

This study compared the the lower limb coordination of the previously injured leg of ACL injured subjects (ACLR, n=18), against their non-injured leg and a control (nACL, n=18) leg. The lower limb joint and segment couplings were calculated during maximal drop-jump land and unanticipated cutting task (n=20). No significant differences were reported in the interlimb coordination between the ACLR and nACL subjects. Differences between the previously injured and nACL control leg were present in the thigh abduction adduction – leg abduction adduction, thigh rotation – leg rotation, hip abduction adduction – knee rotation, hip rotation – knee abduction adduction, and knee rotation – knee abduction adduction couplings. The hip and thigh were the main areas where differences were reported in the previously injured leg compared to the control leg. Altered proximal neuromuscular function may be the origin of these altered coordination patterns.

## 6.2 Introduction

Some of the world's top athletes have been victims of an Anterior Cruciate Ligament (ACL) injury; soccer player Michael Owen, hurler Henry Shefflin, Australian rules football player Jason Winderlich and basketball player Shaun Livingston to name a few. Surgical reconstruction of the injured ACL is usually recommended to allow return to sports participation. Only 20% of the athletes who undergo reconstructive surgery return to their pre-injury-level of sport participation (Söderman et al., 2002; Chong and Tan, 2004). These athletes (ACLR subjects) are at an increased risk of repeated ACL injury (Shelbourne et al., 1998; Oates et al., 1999; Orchard et al., 2001; Kvist, 2004; Myklebust and Bahr, 2005; Salmon et al., 2005; Tanaka et al., 2010) and the development of osteoarthritis (OA) (Roos et al., 1995; Von Porat et al., 2004; Pinczewski et al., 2007; Meuffels et al., 2009; Holm et al., 2010; Øiestad et al., 2010).

Altered biomechanical and neuromuscular function of the lower limbs, as a result of the initial ACL injury, are thought to increase the risk of a repeated ACL injury (Swärd et al., 2010) and degenerative joint disease (Deneweth et al., 2010). A number of previous investigations have examined the angle-time histories of previously injured ACLR subjects in a variety of tasks ranging from walking to drop landing and pivot tasks (Ristanis et al., 2005; Webster et al., 2011). These investigations provide information on the movement patterns of distinct joints. ACL injury however is thought to be the result of a multi-planar interaction of various lower limb biomechanical risk factors. Therefore, composite variables such as intra-limb joint and segment couplings may provide further insight into any compensations present in the function of ACLR subjects' previously injured leg, which may increase the risk of re- injury and/or the development of OA.

Intra-limb joint and segment coordination has been examined previously in rehabilitated populations, currently injured populations and high injury risk populations (Stergiou and Bates, 1997; Hamill et al., 1999; Heiderscheit et al., 2002; Hamill et al., 2005). Investigations reporting intra-limb joint and segment coordination have reported the variability of the lower limb coordination in the majority of studies (Heiderscheit et al., 2002; Pollard, 2003; Hamill et al., 2005; Pollard et al., 2005). It has been shown that injured and high injury risk groups demonstrated lower levels of coordination pattern variability (Heiderscheit et al., 2002; Hamill et al., 2005; Pollard et al., 2005). Hamill et al., (1999) reported both the coordination and its variability in an investigation comparing groups with high and low Q-angles. Coordination and its variability in Hamill et al.'s, (1999) investigation were based on measures of continuous relative phase (CRP), and no differences were reported between groups. Stergiou and Bates, (1997) also reported lower limb coordination as a potential mechanism for lower limb running injuries, where lack of synchronisation between subtalar and knee joint actions was proposed as a potential injury mechanism. In Pollard, (2003) coordination and coordination variability were compared between male and female athletes during a cutting task. Coordination and its variability in Pollard's, (2003) investigation were based on measures of vector coding, with females reported to have decreased coordination variability in a number of couplings. Average coupling angle or coordination pattern also differed by gender in thigh rotation leg rotation and hip abduction-adduction knee rotation couplings. The coordination patterns of previously injured and rehabilitated ACL injured athletes have received limited research attention. To the author's knowledge lower limb coordination has not been previously measured in rehabilitated ACLr subjects during landing or cutting tasks. This composite variable of lower limb coordination, when measured during a match specific task such as landing or cutting may highlight any compensations present in the function of ACLr subjects' previously injured leg. These compensations can then be considered as factors that may increase the risk of re- injury and/or the development of OA in ACLr subjects.

#### 6.2.1 *Aim of the study*

The purpose of this study was to compare the lower limb coordination of ACLr individuals, against the contralateral non-injured leg and a healthy control (nACL) leg during the performance of a maximal drop-jump land and unanticipated cutting task. Specific joint / segment couplings were chosen based on loading patterns that are thought to increase ACL load as identified in Section 2.4.4. In accordance with the previous investigations (Heiderscheit et al., 2002; Pollard et al., 2005) vector coding was used to quantify the coordination of the following joint and segment couplings; thigh abduction-adduction leg abduction-adduction (thigh-abad\_leg-abad), thigh rotation leg rotation (thigh-rot\_leg-rot), hip abduction-adduction knee rotation (hip-abad\_knee-rot), hip rotation knee abduction-



adduction (hip-rot\_knee-abad), and knee rotation knee abduction-adduction (knee-rot\_knee-abad). This novel investigation may provide further insight into any compensations present in the function of ACLr subjects previously injured leg, which may increase the risk of re-injury and / or the development of OA.

## 6.3 Methods

### 6.3.1 *Subject information*

Eighteen subjects who had previously undergone ACL reconstruction, rehabilitation and who were now back in full competitive participation in their sport (ACLR subjects) (Males n=9, age  $26 \pm 4$  years, height  $1.78 \pm 0.1$  m, mass  $81.74 \pm 19.42$  kg, time since injury  $5 \pm 3$  years, Females n=9, age  $22 \pm 2$  years, height  $1.69 \pm 0.06$  m, mass  $66.21 \pm 7.51$  kg, time since injury  $4 \pm 2$  years) were recruited for the present investigation through local sports clubs physiotherapists and orthopaedic surgeons. All ACLr subjects were screened prior to participation, a process outlined Chapter 5. A further 18 gender, height, weight and sport matched subjects who had no history of knee injury (nACL) were also recruited for the present study (Males n=9, age  $22 \pm 3$  years, height  $1.81 \pm 0.09$  m, mass  $80.39 \pm 5.36$  kg, Females n=9, age  $22 \pm 2$  years, height  $1.67 \pm 0.07$  m, mass  $63.81 \pm 6.12$  kg). Control subjects had no history of serious lower limb injury and were injury free for a period of six months prior to testing. All subjects played at a standard of intermediate/ senior club, intercollegiate, inter-county or above and were categorized as highly experienced in their sport. Approval for the participation of human subjects in this investigation was granted by the University of Limerick Research Ethics Committee.

### 6.3.2 *Experimental set-up and testing*

See Section 3.3.2 as an identical experimental set-up was utilized in the present investigation as was used for the investigation reported in Chapter 3.

Following subject screening as detailed in Section 5.3.2 and prior to initiation of the testing protocol each subject completed an informed consent form and, a pre-test questionnaire to screen further for any potential health risks. Subjects with no history of ACL injury completed an injury history and sporting experience questionnaire to ensure all subjects were injury free and adequately experienced in their sport.

Testing protocol was then completed as outlined in Section 3.3.5.

### 6.3.3 *Data treatment and analysis*

The following data reduction procedures were performed on data for all subjects. Cortex (Motion Analysis Corporation, Santa Rosa, CA, USA) was used to track and export raw 3D marker coordinate data. The raw coordinate and ground reaction force data were low-pass

filtered with a fourth-order Butterworth filter with a 12 Hz and 50 Hz cut off frequency respectively. These cut off frequencies were selected based on similar values utilized in previous investigations using similar movement tasks (Besier et al., 2001; Malinzak et al., 2001; Ferber et al., 2003). The thigh, shank and foot segments were modeled as an assembly of cones, and the pelvis was modeled as a cylinder in Visual 3D™ (C-Motion, Rockville, MD, USA). Further detail on the specifications of these segment models are located in Chapter 3. The local coordinate system and joint centers of these segments were defined from a static trial.

Right-handed Cartesian local coordinate systems for the pelvis, thigh, shank and foot segments of the left leg were defined to describe position and orientation of each segment; this was mirrored in the frontal and transverse plane for the right leg to ensure consistent identification of anatomical movements (e.g. abduction and internal rotation) for both legs. Three-dimensional ankle, knee, and hip angles, and thigh and leg segment rotations were calculated using a joint coordinate system approach (Grood and Suntay, 1983). Joint centres were denoted by the midpoint between the medial and lateral calibration markers for the knee and ankle joints (Grood et al., 1984; Wu et al., 2002) and one quarter the distance between the greater trochanter markers in the medial direction (Weinhandl and O'Connor, 2010) for the hip. Initial ground contact and take off were determined by the instant when the vertical ground reaction force exceeded or fell below 10 N respectively. The end of the jump landing phase and the beginning of the cutting movement were defined specific to each trial by the pattern of the vertical ground reaction force on the cutting leg. The landing and cutting components of the task were separated. The land was defined from touchdown to the end of landing, and the cut from the start of cutting to take off on the push off leg. These two components were cropped and time normalized to 1001 data points.

Intralimb couplings were created for the segments and joints listed in Section 6.2.1, from time normalised three-dimensional lower limb segment rotations and joint angles. Using angle-angle plots, quantification of joint coordination was obtained using a modification of the vector coding technique advocated by Sparrow (1987) and consistent with previous work (Heiderscheit et al., 2002; Pollard et al., 2005). Coupling angles were calculated using the orientation to the right horizontal of the resultant vector between two adjacent data points in a timeseries as described and illustrated in section 4.3.3.

Following conversion from radians to degrees, the resulting coupling angles were between 0 - 180°. These values were then converted into a range between 0-90°. Therefore, a coupling angle of 45°, 90° or 0° would indicate equal contribution from both segments/joints or symmetrical motion. Alternatively a coupling angle of 90° or 0° would indicate no

contribution from the proximal or distal segment/ joint respectively or an asymmetrical motion.

Average coupling angle was calculated during several regions of the landing component of the task; the initial 40 ms and 40% of the landing stance, between 15-30% of landing stance and the entire 100% of the landing stance period. The regions utilised calculation of average coupling angle for the cutting component of the task were the between 70-100% of cutting stance and the entire 100% of the cutting stance period. These regions were chosen based on reasons outlined in Section 4.3.2.

#### 6.3.4 Statistical analysis

Average coupling angle in each of the specified regions was calculated for each coupling during the landing and cutting components of the task. Differences between ACLr and nACL subjects (ACLR previously injured leg -V- dominance matched nACL control leg), for average coupling angle were analysed using a repeated measures ANOVA test, with injury as a between and trial as a within subject factor. Differences within ACLr subjects (previously injured ACLr leg -V- injury free ACLr leg) for average coupling angle were assessed using a repeated measures ANOVA test, with trial and leg as within subject factors. Significance for all tests was set at  $p < .05$ , except in cases where a moderate - large effect size was present with a  $p$ -value  $< 0.1$ . In such cases significance was set at  $p < 0.1$ . All statistical analyses were performed using SPSS (PASW v18.0, IBM Inc. Armonk, NY). Partial  $\eta^2$  ( $\eta p^2$ ) was also reported as a measure of effect size. It was calculated using the formula:  $\eta p^2 = SS_{\text{effect}} / (SS_{\text{effect}} + SS_{\text{error}})$ , where  $SS_{\text{effect}}$  = effect variance and  $SS_{\text{error}}$  = error variance. Interpretation of effect size was based on the scale for effect size classification of Hopkins (2000). This scale is based on  $f$ -values for effect size and these were converted to  $\eta p^2$  using the formula:  $f = (\eta p^2 / (1 - \eta p^2))^{0.5}$ . Consequently, the scale for classification of  $\eta p^2$  was  $< 0.04$  = trivial, 0.041 to 0.249 = small, 0.25 to 0.549 = medium, 0.55 to 0.799 = large, and  $> 0.8$  = very large.

## 6.4 Results

Table 6-1 presents a summary of the differences present in the coupling angle of all 3D joint and segment couplings when averaged across several regions of the landing and cutting stance phases.

Table 6-1. Differences between ACLr and nACL subjects in the average coupling angle of selected 3D joint and segment angle coupling combinations during landing and cutting.

Average coupling angle (°) for PI, NI and control legs during the initial 40% and 40ms of landing and also during 15-30%, and 100% of landing stance phase are shown. Group differences (°), Partial eta<sup>2</sup> ( $\eta^2$ ) and p-values are also presented. The scale for classification of  $\eta^2$  was < 0.04 = **trivial**, 0.041 to 0.249 = **small**, 0.25 to 0.549 = **medium**, 0.55 to 0.799 = **large**, and >0.8 = **very large**. **PI=AClr previously injured leg, NI=AClr non-injured leg, C=nACL dominance matched control leg**. P-values in **bold** indicate significant differences with p<0.05. The underlined average coupling angle indicate the value of largest value.

Coupling		PI	NI	Diff	$\eta^2$	p-value	C	Diff	$\eta^2$	p-value
Thigh Abduction Adduction – Leg Abduction Adduction										
Land	100%	42.82	43.74	0.92	0.18	0.067	45.56	2.75	0.08	0.091
	0-40%	49.69	50.00	0.31	0.01	0.688	52.83	3.14	0.05	0.213
	0-40ms	56.02	58.35	2.33	0.18	0.067	58.24	2.22	0.02	0.469
	15-30%	47.56	47.31	0.25	<0.01	0.851	51.52	3.96	0.03	0.286
Cut	100%	44.70	45.10	0.40	<0.01	0.910	47.42	2.72	<0.01	0.762
	70-100%	63.15	63.22	0.07	<0.01	0.933	64.78	1.63	0.04	0.221
Thigh Rotation – Leg Rotation										
Land	100%	36.98	37.31	0.32	0.02	0.578	36.93	0.05	<0.01	0.971
	0-40%	43.03	43.23	0.20	<0.01	0.828	39.09	3.95	0.08	0.089
	0-40ms	<u>46.07</u>	46.44	0.38	<0.01	0.779	38.74	7.33	<b>0.12</b>	<b>0.039</b>
	15-30%	44.82	43.70	1.12	0.05	0.339	40.89	3.93	0.06	0.159
Cut	100%	40.87	39.83	1.04	0.04	0.439	38.46	2.41	0.04	0.238
	70-100%	<u>34.98</u>	33.61	1.38	<0.01	0.833	34.37	0.62	<b>0.22</b>	<b>0.004</b>
Hip Abduction Adduction – Knee Rotation										
Land	100%	55.78	55.73	0.05	<0.01	0.934	54.97	0.81	<0.01	0.737
	0-40%	61.87	62.26	0.39	0.03	0.491	60.33	1.54	0.01	0.581
	0-40ms	65.29	66.11	0.82	0.07	0.292	62.58	2.71	0.03	0.314
	15-30%	62.96	63.30	0.34	0.01	0.687	61.01	1.96	0.01	0.578
Cut	100%	49.21	49.49	0.27	0.18	<b>0.075</b>	48.67	0.54	<b>0.26</b>	<b>0.002</b>
	70-100%	<u>56.21</u>	52.23	3.97	<b>0.22</b>	<b>0.041</b>	53.89	2.32	<0.01	0.699
Hip Rotation – Knee Abduction Adduction										
Land	100%	<u>40.19</u>	39.51	0.68	<b>0.21</b>	<b>0.048</b>	35.16	5.03	<b>0.24</b>	<b>0.002</b>
	0-40%	<u>40.73</u>	39.23	1.51	<b>0.28</b>	<b>0.020</b>	36.15	4.59	<b>0.13</b>	<b>0.030</b>
	0-40ms	42.03	40.93	1.10	0.04	0.414	37.55	4.48	0.08	0.105
	15-30%	40.11	37.94	2.17	0.19	0.062	36.85	3.26	0.06	0.138
Cut	100%	35.26	35.34	0.08	0.08	0.230	31.10	4.16	<0.01	0.915
	70-100%	30.55	28.44	2.11	<0.01	0.938	27.73	2.82	<0.01	0.597
Knee Rotation – Knee Abduction Adduction										
Land	100%	45.77	45.58	0.18	0.03	0.455	47.51	1.75	0.01	0.511
	0-40%	40.03	39.93	0.10	0.01	0.646	44.85	4.82	0.06	0.162
	0-40ms	38.49	39.17	0.68	0.02	0.528	<u>48.95</u>	10.5	<b>0.22</b>	<b>0.004</b>
	15-30%	<u>37.59</u>	36.15	1.45	<b>0.27</b>	<b>0.023</b>	41.51	3.92	<0.01	0.578
Cut	100%	40.18	40.75	0.57	<0.01	0.896	<u>41.53</u>	1.35	<b>0.15</b>	<b>0.020</b>
	70-100%	32.01	32.17	0.16	0.03	0.480	31.82	0.19	0.01	0.512

Several differences were reported between the previously injured (PI) leg and both the non-injured (NI) and control leg controls. Differences between the PI and nACL control leg were present in the thigh-abad\_leg-abad, thigh-rot\_leg-rot, hip-abad\_knee-rot hip-rot\_knee-abad, and knee-rot\_knee-abad couplings. Differences between the PI leg and contralateral NI leg were present in the hip-rot\_knee-abad, and knee-rot\_knee-abad couplings (Table 6-2). The

differences between the PI and NI leg were further investigated and compared to the differences present between the PI and NI matched legs on the nACL subjects. Neither coupling had significantly different intra-subject leg differences between the ACLr and nACL populations. Even though differences were reported between the PI and NI leg in these couplings, these intra-subject leg differences were not significantly different from the intra-subject leg differences in the nACL population.

Table 6-2. Intra-subject leg differences (Average ( $\pm$  Standard Deviation)) for ACLr and nACL subjects in the average coupling angle of the 3D joint and segment angle coupling combinations that had significant PI-NI leg differences.

Average between leg difference ( $^{\circ}$ ) for ACLr and nACL subjects during selected regions of landing and cutting stance phase are shown. Group differences ( $^{\circ}$ ), Partial eta<sup>2</sup> ( $\eta^2$ ) and p-values are also presented. The scale for classification of  $\eta^2$  was  $< 0.04$  = trivial, 0.041 to 0.249 = small, 0.25 to 0.549 = medium, 0.55 to 0.799 = large, and  $> 0.8$  = very large.

Coupling		ACLR	nACL	Diff	$\eta^2$	p-value
Hip Abduction Adduction – Knee Rotation						
Cut	70-100%	25.34 (5.17)	22.23 (5.07)	3.11	0.05	0.179
Hip Rotation – Knee Abduction Adduction						
Land	100%	0.68 (1.36)	0.33 (1.80)	0.35	0.01	0.511
	0-40%	1.51 (2.49)	1.10 (2.72)	0.41	$< 0.01$	0.641
Knee Rotation – Knee Abduction Adduction						
Land	15-30%	1.45 (2.97)	2.16 (3.87)	-0.72	0.01	0.537

Figure 6-1, Figure 6-2 and Figure 6-3 present the average coupling angle of all the joint/segment couplings that showed significant groups differences.

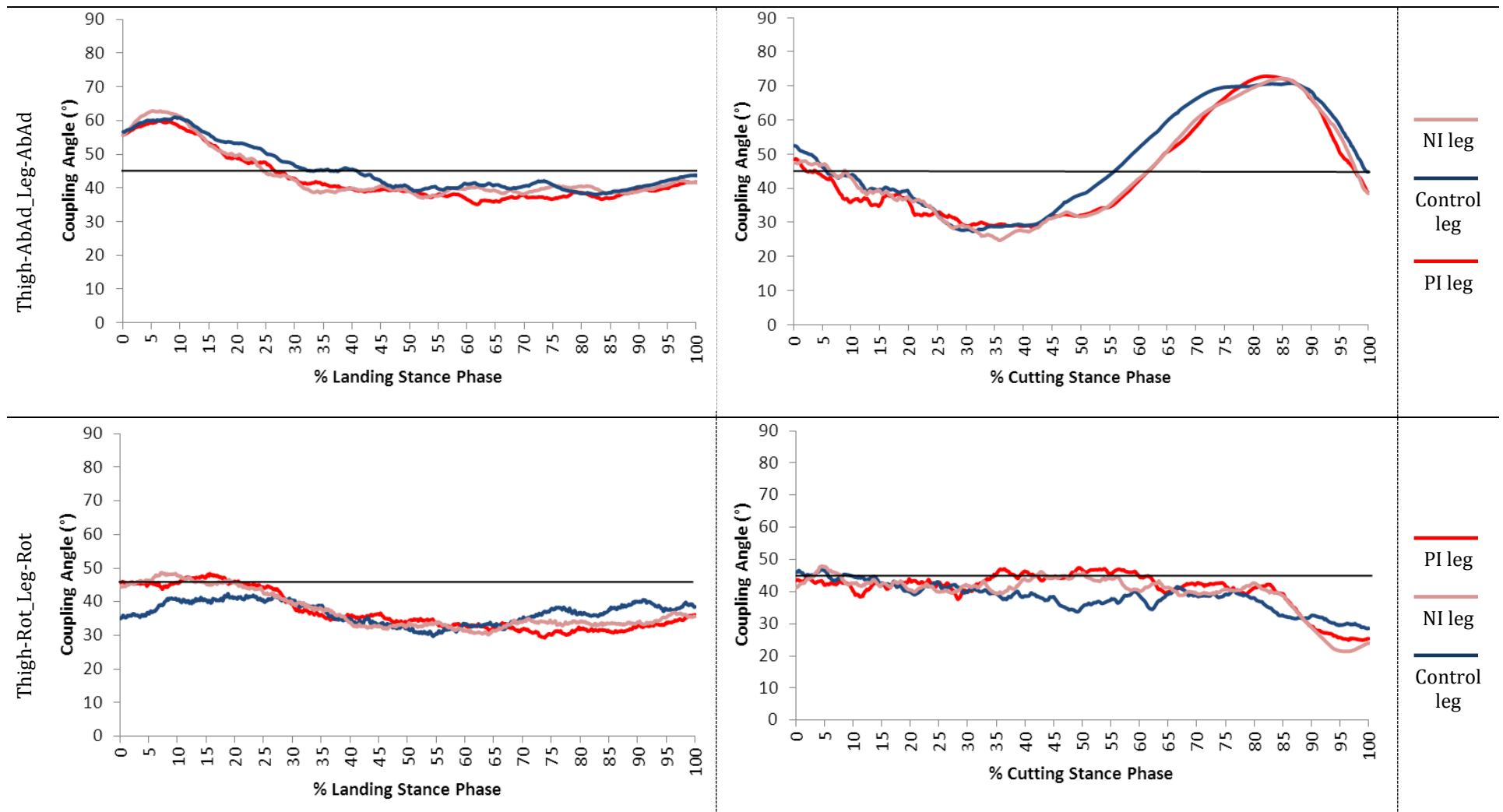


Figure 6-1. Average coupling angle of selected joint/segment couplings during landing and cutting.

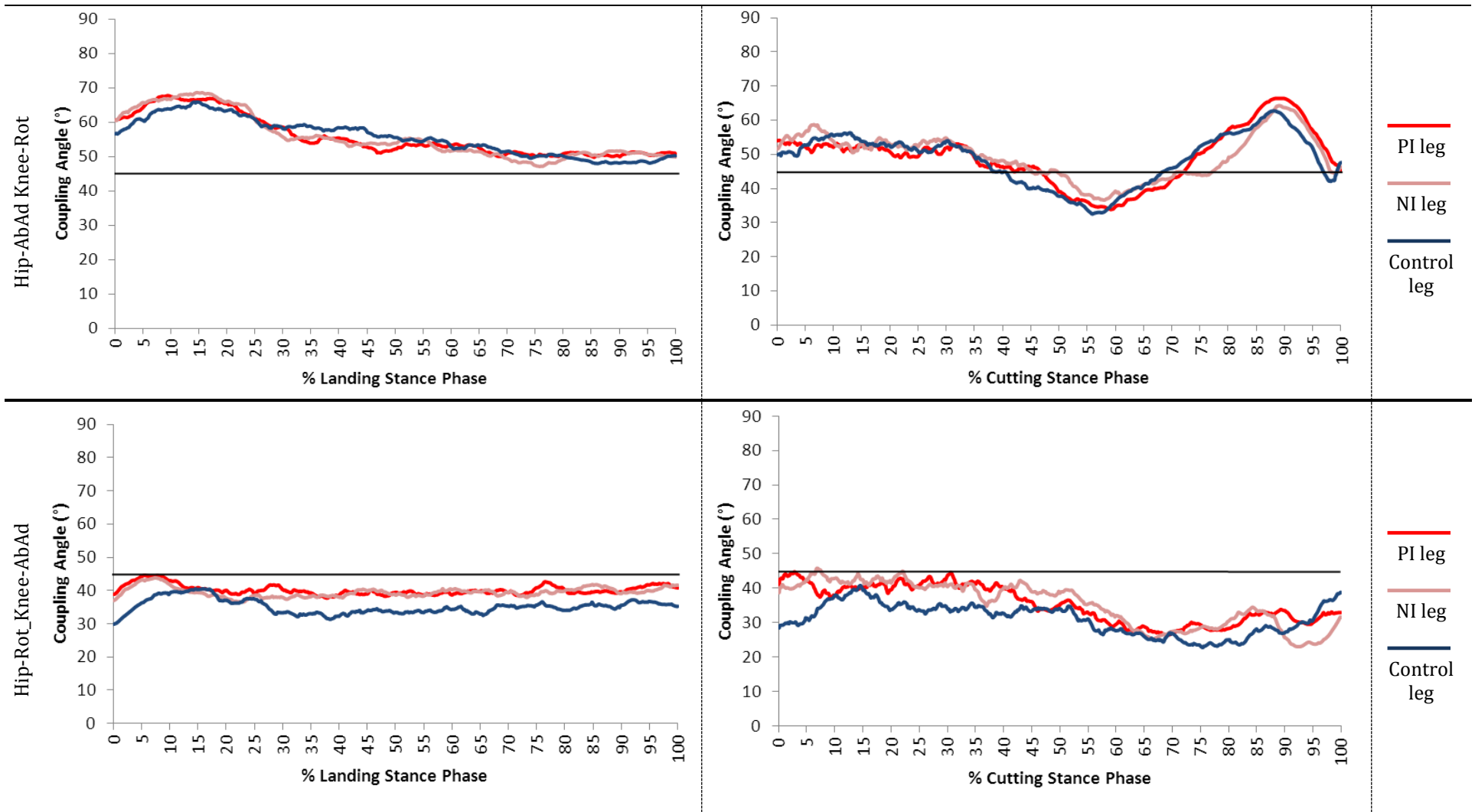


Figure 6-2. Average coupling angle of selected joint/segment couplings during landing and cutting.

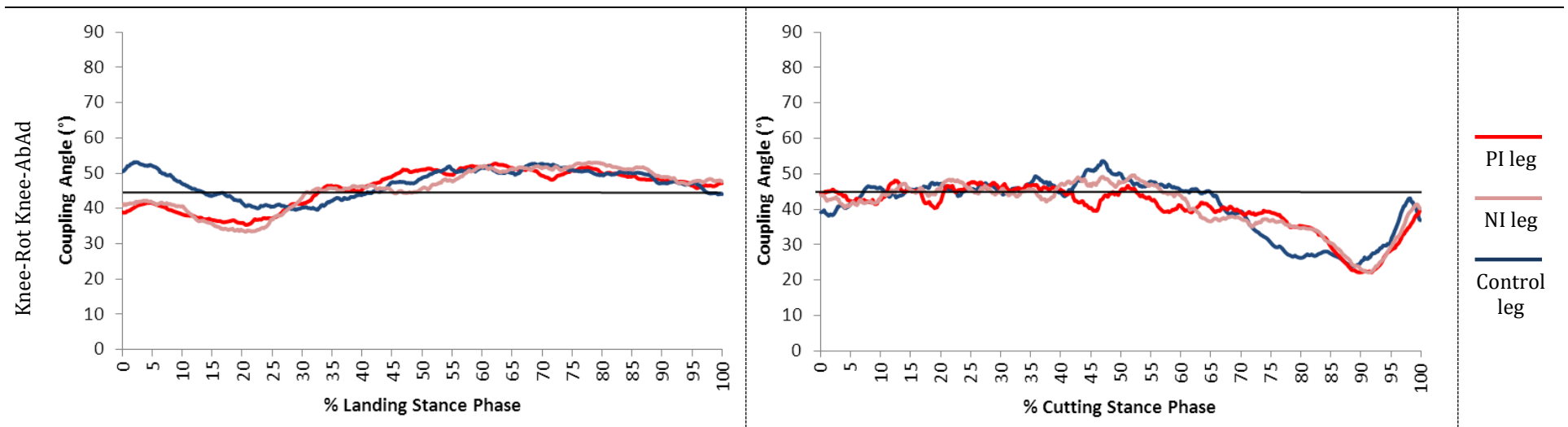


Figure 6-3. Average coupling angle of selected joint/segment couplings during landing and cutting.



## 6.5 Discussion

In the present study, the average coupling angle of lower extremity joint and segment couplings of ACLr individuals was compared against the contralateral non-injured leg and a healthy control (nACL) leg during the performance of a maximal drop-jump land and unanticipated cutting task.

The coordination of the PI leg of ACLr subjects was reported to be significantly different to the coordination reported in the contralateral NI leg for a number of couplings during various regions of the landing and cutting task. These ACLr subject between leg differences were compared to nACL subject between leg differences for the selected couplings and regions of stance where significant PI NI leg differences were identified. No coupling was reported to have a significant difference between the ACLr and nACL populations in the intra-subject leg differences. It can therefore be assumed that the surgical and rehabilitation interventions were successful in allowing the ACLr subjects to regain similar lower limb biomechanics in both the PI and NI leg, and that any compensation present has affected both PI and NI legs.

The thigh-abad\_leg-abad coupling coordination pattern, illustrated in Figure 6-1, fluctuates between a coordination pattern dominated by either frontal plane thigh or leg motion throughout the stance phases of the landing and cutting task. The PI, NI and control legs all follow this pattern with no differences between the PI, NI or control legs.

The thigh-rot\_leg-rot coupling coordination pattern, illustrated in Figure 6-1, fluctuate just below 45° or a symmetrical movement throughout the stance phases of the landing and cutting task. The range of the coordination pattern illustrated in Figure 6-1 is a lot narrower than that reported in Pollard, (2003); the general pattern for the cutting component of the task is similar with a gradual increase in the contribution of transverse plane leg motion. There are regions that demonstrate differences between the PI and control legs. The initial 40 ms of landing shows a PI leg with movement dominated by thigh rotation and a control leg with movement dominated by leg rotation. If this is considered alongside the findings of the previous chapter, the increased transverse plane range of motion (ROM) at the knee in the PI leg may be originate from increased thigh rotation. Increased transverse plane ROM at the knee has been suggested as a factor increasing the risk of OA development (Stergiou 2007). If the contribution of thigh rotation on the PI leg could be decreased it is likely that transverse plane ROM at the knee may also decrease, along with the associated risk of OA development. This pattern is reversed in the latter stages (70-100%) of the cutting component of the task, where the PI leg has significantly less contribution from the transverse rotation of the shank. During this region of the cutting task limited shank rotation could be dangerous when the upper body and thigh are turning in the direction of the cutting movement. This altered

coordination during cutting in the PI leg of ACLr subjects could place strain on the ACL and increase risk of re-injury. The magnitude of this difference ( $0.62^\circ$ ) is however not thought to be enough to strain the ACL to a magnitude sufficient to cause injury. Pollard, (2003) reported increased contribution from the transverse plane thigh motion in female athletes during 35-45% of an isolated decelerating cutting task similar to what was reported in the PI leg during the landing component of the task in the present investigation. The magnitude of the difference reported in Pollard, (2003) was  $14.2^\circ$  highlighting the minimal difference in the thigh-rot\_leg-rot coupling for the present investigation.

The average hip-abad\_knee-rot coupling coordination pattern, illustrated in Figure 6-1b is dominated by frontal plane hip motion during landing. This is reversed midway through the cut followed by a return to a pattern dominated again by frontal plane hip motion at ~90% of the cutting stance phase. This hip-abad\_knee-rot coupling coordination pattern reverts to a symmetrical pattern at the end of the cut or the push off. The range of coordination pattern in the current investigation is narrower than that reported in Pollard, (2003) but the general pattern for the cutting component of the task is very similar with a swap between contribution from transverse plane motion and frontal plane hip motion after 40% of the cutting. The PI, NI and control legs appear to follow a similar coordination pattern; significant differences were reported during the cut however between the PI and control leg. On average over the entire cutting stance phase the PI leg was reported to have greater contribution from the frontal plane motion at the hip. The magnitude of this difference ( $0.54^\circ$ ) is not thought to be enough to alter coordination patterns sufficiently in order to increase the risk of re-injury or development of degenerative knee joint disease. Pollard, (2003) reported a  $21.3^\circ$  larger contribution from the frontal plane hip motion in the female population during 35-45% of the cutting stance phase, again highlighting the minimal difference in the hip-abad\_knee-rot coupling for the present investigation.

The average hip-rot\_knee-abad coupling coordination pattern, illustrated in Figure 6-2 remains dominated by frontal plane knee motion for the landing and cutting task. The range of coordination pattern in the current investigation is narrower than that reported in Pollard, (2003) but the general pattern for the cutting component of the task is very similar with swap between contribution from transverse plane motion and frontal plane hip motion after 40% of the cutting task. The control leg has an even greater contribution from frontal plane motion at the knee during the land when compared to the PI leg. The increased contribution from frontal plane motion at the knee in the control leg may act to control the transverse plane rotation hip and possibly the thigh. Increased transverse plane ROM at the knee has been suggested as a factor increasing the risk of OA development (Stergiou 2007). Therefore, control or reduction of transverse plane rotation at the hip or thigh could result in decreased risk of the developing of degenerative knee joint disease.

The knee-rot\_knee-abad coupling coordination pattern illustrated in Figure 6-3, fluctuates around 45° or a symmetrical movement after the initial landing phase until the latter stages of the cut. The general coordination pattern for the cutting component of the task in the present investigation replicates a minimized version of the pattern in Pollard, (2003), moving through a narrower range. ACLr and nACL subjects diverge on either side of symmetrical in the initial landing period, and both populations become dominated by frontal plane knee motion in the latter stages of the cut. The initial deceleration phase of the landing for the control leg coordination pattern is dominated by transverse plane knee and the PI leg coordination pattern is dominated by the frontal plane knee motion. This is replicated to a lesser degree in the latter stages of the cut where both PI and control legs are dominated by frontal plane knee motion which is more prominent in the PI leg. The increased contribution of the transverse plane knee motion in this coupling, which may involve external rotation, could act to control the frontal plane motion at the knee. This increase in frontal plane knee motion control may decrease the knee abduction moment and knee abduction angles thought to increase the risk of the OA development (Cerejo et al., 2002) and repeated ACL injury (Paterno et al., 2010).

## **6.6 Conclusion**

There were limited between leg differences in coordination patterns between the PI and NI leg of the ACLr subjects. This replicates the similar joint kinematics and kinetics between the PI and NI leg in Chapter 5. The ACL reconstructive surgery and rehabilitation is therefore, thought to restore a level of symmetry in lower limb coordination. The exact origin of this symmetry is unclear from the results of the current investigation; it may due to the PI leg returning to pre-injury coordination patterns or a compensation that has affected both the PI and NI leg.

Most of the differences reported were between the PI and control leg. The hip and thigh were the main areas where alterations were reported in the PI leg compared to the control leg. Altered proximal control may be the driver in producing these altered coordination patterns. Future work investigating the influence of neuromuscular and strength training at the hip joint on these altered coordination patterns may be insightful.

Future research investigating the reproducibility of these altered coordination patterns may also be of interest. If these altered coordination patterns are reproduced over several trials in an overly consistent/ invariable manner this could lead to cartilage degeneration and premature osteoarthritis in ACLr knees (Gao and Zheng, 2010).

## 6.7 References

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## **CHAPTER 7**

# **THE ROLE OF PREVIOUS ANTERIOR CRUCIATE LIGAMENT INJURY ON THE VARIABILITY OF JOINT KINEMATICS, KINETICS AND COORDINATION DURING A MATCH SPECIFIC LAND-CUT TASK**

## 7.1 Abstract

This study compared the movement and coordination variability of the previously injured leg of ACL injured subjects (ACLR, n=18), against their non-injured leg and a control (nACL, n=18) leg. The variability of lower limb joint angles and couplings were calculated during a maximal drop-jump land and unanticipated cutting task (n=20). The previously injured leg had less variability than the non-injured leg in the knee rotation–knee abduction adduction coupling, and more variability than the nACL leg in frontal and transverse knee joint angles and hip rotation–knee abduction adduction coupling. Reduced coordination variability could produce a more repetitive loading pattern linked to cartilage degeneration. Increased movement and coordination variability may stem from proprioceptive deficits on the previously injured leg and decrease the ability to adapt to perturbations.

## 7.2 Introduction

Injuries to the anterior cruciate ligament (ACL) of the knee continue to be a common and highly publicised occurrence in athletes. Up to 80% of the athletes who undergo reconstructive surgery do not successfully return to pre-injury-level sport participation and quit their sports (Söderman et al., 2002; Chong and Tan, 2004; Ardern et al., 2010). The 20% who are successful in returning to their sport (ACLR subjects) are at an increased risk of repeated ACL injury (Shelbourne et al., 1998; Oates et al., 1999; Orchard et al., 2001; Kvist, 2004; Myklebust and Bahr, 2005; Salmon et al., 2005; Tanaka et al., 2010) and the development of osteoarthritis (OA) (Roos et al., 1995; Von Porat et al., 2004; Pinczewski et al., 2007; Meuffels et al., 2009; Holm et al., 2010; Øiestad et al., 2010). Altered biomechanical and neuromuscular function of the lower limbs, as a result of the initial ACL injury, are likely to increase the risk of a repeated ACL injury (Svärd et al., 2010) and degenerative joint disease (Deneweth et al., 2010).

Regular repetition of these altered kinematics and associated abnormal loading patterns during daily activity could lead to cartilage degeneration and premature osteoarthritis in ACLR knees (Gao and Zheng, 2010). Altered levels of movement variability in ACLR individuals, has therefore, also been suggested as a factor increasing risk of injury and development of degenerative joint disease. After ACL reconstruction the previously injured (PI) knee has been shown to demonstrate a more variable and unpredictable behaviour (Moraiti et al., 2010) in comparison to matched nACL subjects. Similarly Kiefer et al., (2008) reported more variable hip-ankle coordination in ACLR individuals when compared to matched nACL subjects during a slow balance task. A small cohort of the ACLR subjects tested in Kiefer et al.'s, (2008) study went on to second injury (Foster, 2009). During a slow balance task, these ACL re-injury subjects demonstrated very high variability—higher than the other ACLR subjects. During the fast balance task, these ACLR re-injury subjects locked down the

lower extremity and due to increased task difficulty limited the degrees of freedom, and their variability decreased. All these responses were in opposition to what was demonstrated in the non-injured control (nACL) subjects. When ACLr subjects were exposed to a demanding task they demonstrated lower variability than a control similar to decreases reported in ACL deficient (ACLd) subjects (Stergiou et al., 2004; Georgoulis et al., 2006; Moraiti et al., 2010; Zampeli et al., 2010).

The variability demonstrated may be functional to the demands of a task. ACLr individuals feel “secure” enough to increase and add extra movement during gait however since proprioceptive deficits may be present, the temporal structure of the stride-to-stride variations are not restored to normative levels. On the contrary, the rigidity found in the ACLd knees signifies that they are more “careful” in the way they walk, trying to eliminate extra movement (Georgoulis et al., 2006; Moraiti et al., 2010; Zampeli et al., 2010; Decker et al., 2011). These “careful” rigid movements are replicated by the ACLr individuals in the more difficult balancing task, similar as would be expected when an individual is introduced to a new skill (Bernstein, 1967). The relationship between previous ACL injury and movement variability is a highly under investigated one. Its potential as a risk factor for re-injury and development of degenerative joint disease strongly merits further investigation.

#### *7.2.1 Aim of the study*

The purpose of this study was to compare variability of hip and knee joint kinematics (movement variability), and lower extremity joint and segment coupling (coordination variability), of ACLr individuals, against the contralateral non-injured leg and a healthy control (nACL) during the performance of a maximal drop-jump land and unanticipated cutting task. The hip and knee joint kinematics analysed were flexion-extension, abduction-adduction and internal-external rotation at the hip and knee. Specific joint / segment couplings were chosen based on loading patterns that are thought to increase ACL load as identified in Section 2.4.4. The joint and segment couplings analysed were; thigh abduction-adduction leg abduction-adduction (thigh-abad\_leg-abad), thigh rotation leg rotation (thigh-rot\_leg-rot), hip abduction-adduction knee rotation (hip-abad\_knee-rot), hip rotation knee abduction-adduction (hip-rot\_knee-abad), and knee rotation knee abduction-adduction (knee-rot\_knee-abad). Based on the variability hypothesis the previously injured leg of ACLr individuals is expected to demonstrate lower levels of both movement and coordination variability when compared to the contralateral non-injured ACLr leg. This decreased variability may result in a more repetitive loading pattern linked to cartilage degeneration and premature osteoarthritis in ACLr knees. The previously injured leg of ACLr individuals is expected to demonstrate higher levels of both movement and coordination variability when compared to the match nACL control leg. These higher levels of variability may originate from proprioceptive deficits present in the ACLr individuals.



## 7.3 Methods

### 7.3.1 Subject information

Eighteen subjects who had previously undergone ACL reconstruction, rehabilitation and who were now back in full competitive participation in their sport (ACLR subjects) (Males  $n=9$ , age  $26 \pm 4$  years, height  $1.78 \pm 0.1$  m, mass  $81.74 \pm 19.42$  kg, time since injury  $5 \pm 3$  years, Females  $n=9$ , age  $22 \pm 2$  years, height  $1.69 \pm 0.06$  m, mass  $66.21 \pm 7.51$  kg, time since injury  $4 \pm 2$  years) were recruited for the present investigation through local sports clubs physiotherapists and orthopaedic surgeons. All ACLR subjects were screened prior to participation, this process is outlined Chapter 5. A further 18 gender, height, weight and sport matched subjects who had no history of knee injury (nACL) were also recruited for the present study (Males  $n=9$ , age  $22 \pm 3$  years, height  $1.81 \pm 0.09$  m, mass  $80.39 \pm 5.36$  kg, Females  $n=9$ , age  $22 \pm 2$  years, height  $1.67 \pm 0.07$  m, mass  $63.81 \pm 6.12$  kg). Control subjects had no history of serious lower limb injury and were injury free for a period of six months prior to testing. All subjects played at a standard of intermediate/ senior club, intercollegiate, inter-county or above and were categorized as highly experienced in their sport. Approval for the participation of human subjects in this investigation was granted by the University of Limerick Research Ethics Committee.

### 7.3.2 Experimental set-up and testing

See Section 3.3.2 as an identical experimental set-up was utilized in the present investigation was used for the investigation reported in Chapter 3.

Following subject screening as detailed in Section 5.3.2 and prior to initiation of the testing protocol each subject completed an informed consent form, a pre-test questionnaire to screen further for any potential health risks. Subjects with no history of ACL injury completed an injury history and sporting experience questionnaire to ensure all subjects were injury free and adequately experienced in their sport.

Testing protocol was then completed as outlined in Section 3.3.5.

### 7.3.3 Data treatment and analysis

The following data reduction procedures were performed on data for all subjects. Cortex (Motion Analysis Corporation, Santa Rosa, CA, USA) was used to track and export raw 3D marker coordinate data. The raw coordinate and ground reaction force data were low-pass filtered with a fourth-order Butterworth filter with a 12 Hz and 50 Hz cut off frequency respectively. These cut off frequencies were selected based on similar values utilised in previous investigations utilising similar movement tasks (Besier et al., 2001; Malinzak et al., 2001; Ferber et al., 2003). The thigh, shank and foot segments were modeled as an assembly of cones, and the pelvis was modeled as a cylinder in Visual 3D™ (C-Motion, Rockville, MD,

USA). Further detail on the specifications of these segment models are located in Chapter 3. The local coordinate system and joint centers of these segments were defined from the static trial.

Right-handed Cartesian local coordinate systems for the pelvis, thigh, shank and foot segments of the left leg were defined to describe position and orientation of each segment; this was mirrored in the frontal and transverse plane for the right leg to ensure consistent identification of anatomical movements (e.g. abduction and internal rotation) for both legs. Three-dimensional ankle, knee, and hip angles, and thigh and leg segment rotations were calculated using a joint coordinate system approach (Grood and Suntay, 1983). Joint centres were denoted by the midpoint between the medial and lateral calibration markers for the knee and ankle joints (Grood et al., 1984; Wu et al., 2002) and one quarter the distance between the greater trochanter markers in the medial direction (Weinhandl and O'Connor, 2010) for the hip. Initial ground contact and take off were determined by the instant when the vertical ground reaction force exceeded or fell below 10 N respectively. The end of the jump landing phase and the beginning of the cutting movement were defined specific to each trial by the pattern of the vertical ground reaction force on the cutting leg. The landing and cutting components of the task were separated. The land was defined from touchdown to the end of landing, and the cut from the start of cutting to take off on the push off leg. These two components were cropped and time normalized to 1001 data points.

Intralimb couplings were created for the segments and joints listed in Section 7.2.1, from time normalised three-dimensional lower limb segment rotations and joint angles. Using angle-angle plots, quantification of joint coordination was obtained using a modification of the vector coding technique advocated by Sparrow, (1987) and consistent with previous work (Heiderscheit et al., 2002; Pollard et al., 2005). Coupling angles were calculated using the orientation to the right horizontal of the resultant vector between two adjacent data points in a timeseries as described and illustrated in Section 4.3.3.

Following conversion from radians to degrees, the resulting coupling angles were between 0 - 180°. These values were then converted into a range between 0-90°. Therefore, a coupling angle of 45°, 90° or 0° would indicate equal contribution from both segments/joints or symmetrical motion. Alternatively a coupling angle of 90° or 0° would indicate no contribution from the proximal or distal segment/ joint respectively or an asymmetrical motion.

Variability of the normalized timeseries data for the five coupling angles, and 3D joint angles of the hip and knee were calculated on a point by point basis from the mean ensemble curves of 20 and 10 trials for the land and cut data respectively, as described in Section 4.3.2. Heiderscheit et al., (2002) suggested that averaging the variability within specific regions of

the stance phase rather than across the entire stance phase would provide a more precise analysis for detecting between-group differences, Pollard et al., (2005) also utilised this method. The regions utilised for the calculation of average variability for the landing component of the task were the initial 40 ms and 40% of the landing stance, between 15-30% of landing stance and the entire 100% of the landing stance period. The regions utilised calculation of average variability for the cutting component of the task were the between 70-100% of cutting stance and the entire 100% of the cutting stance period. These regions were chosen based on reasons outlined in Section 4.3.2.

#### *7.3.4 Statistical analysis*

Average variability in each of the specified regions was calculated for each variable during the landing and cutting components of the task. Differences between ACLr and nACL subjects (ACLR previously injured leg -V- dominance matched nACL control leg), for average variability were assessed using independent t-tests and Mann-Whitney U tests, for normally and not normally distributed data respectively. Differences within ACLr subjects (previously injured ACLr leg -V- injury free ACLr leg) for average variability were assessed using paired t-tests and Wilcoxin signed-rank tests, for normally and not normally distributed data respectively. Significance for all tests was set at  $p < .05$ , except in cases where a moderate - large effect size was present with a p-value  $< 0.1$ , in such cases significance was set at  $p < 0.1$ . All statistical analyses were performed using SPSS (PASW v18.0, IBM Inc. Armonk, NY). Cohen's d was utilised as a measure of effect size. It was calculated as the mean difference between genders divided by the pooled standard deviation. Interpretation of effect size was based on the scale proposed by Cohen (1988), 0.2, 0.5 and +0.8 represented small, moderate and large differences respectively.

## 7.4 Results

### 7.4.1 Jump Landing

Table 7-1 presents a summary of the significant differences present in variability of the 3D joint angles at the knee joint when averaged across four regions of the landing stance phase.

Table 7-1 Significant differences in the variability of 3D Knee Joint Angles between ACLr and nACLR subjects during landing.											
Average variability (°) for PI and control legs during the initial40% and 40ms of landing and also during 15-30%, and 100% of landing stance phase are shown. Group differences (°), Cohen’s d and p-values are also presented. The scale for classification of Cohen’s d effect size was, 0.2 = <span>small</span> , 0.5 = <span>moderate</span> , > 0.8 = <span>large</span> .											
Jump Land		ACLR_Pi-NI					ACLR-Control				
		PI	NI	Diff	d	p-value	PI	C	Diff	d	p-value
Knee	<b>Abduction</b>										
	<b>Adduction</b>										
	Initial 40%						<b>1.99</b>	1.26	0.73	<span>1.56</span>	<0.001
	Initial 40ms						<b>1.70</b>	1.07	0.63	<span>1.38</span>	<0.001
	15-30%						<b>2.01</b>	1.22	0.79	<span>1.57</span>	<0.001
	100%						<b>2.48</b>	1.72	0.73	<span>1.22</span>	0.001
	<b>Internal-External</b>										
	<b>Rotation</b>										
	Initial 40%						<b>3.36</b>	1.90	1.45	<span>1.42</span>	<0.001
	Initial 40ms						<b>3.02</b>	1.73	1.30	<span>1.21</span>	0.001
	15-30%						<b>3.37</b>	2.00	1.38	<span>1.27</span>	0.001
	100%						<b>3.22</b>	2.16	1.06	<span>1.10</span>	0.001

The PI leg of the ACLr subjects had more variability than the control leg in the transverse and frontal plane knee angle . This was the case in every region of landing. Similar variability in hip and knee angles was reported in the PI and NI leg of ACLr subjects. Figure 7-1 presents the variability of 3D knee joint angles with significant group differences during landing.

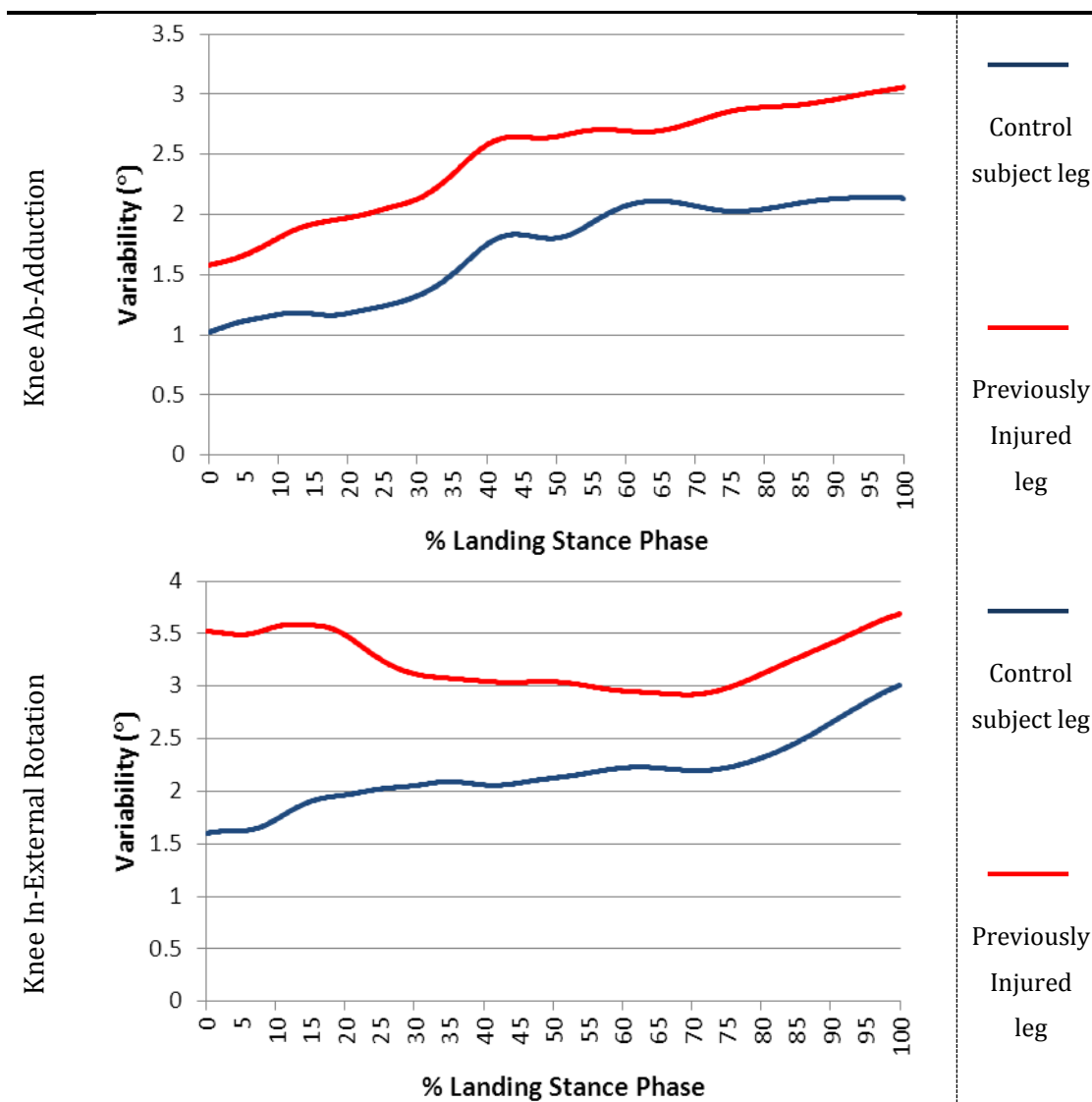


Figure 7-1 Variability of knee abduction-adduction and knee internal-external rotation angle (°), with significant ACLr-nACL differences during landing.

Table 7-2 presents a summary of the significant differences present in the variability of selected 3D joint and segment couplings when averaged across four regions of the landing stance phase.

Table 7-2. Significant differences between ACLr and nACL subjects in the variability of selected 3D joint and segment angle coupling combinations during landing.

Average variability (°) for PI, NI and control legs during the initial 40% and 40ms of landing and also during 15-30%, and 100% of landing stance phase are shown. Group differences (°), Cohen's d and p-values are also presented. The scale for classification of Cohen's d effect size was, 0.2 = **small**, 0.5 = **moderate**, > 0.8 = **large**

Jump Land		ACLR_Pi-NI					ACLR-Control				
		PI	NI	Diff	d	p-value	PI	C	Diff	d	p-value
<b>Hip-Rot_Knee-Ab-Ad</b>	100%						<b>20.6</b>	18.86	1.75	<b>0.91</b>	0.011
<b>Knee-Rot_Knee-Ab-Ad</b>	Initial 40ms	22.30	<b>23.81</b>	1.50	<b>0.56</b>	0.038					
	Initial 40%	23.43	<b>24.50</b>	1.12	<b>0.73</b>	0.012					
	100%	23.98	<b>25.00</b>	0.99	<b>1.02</b>	0.001					

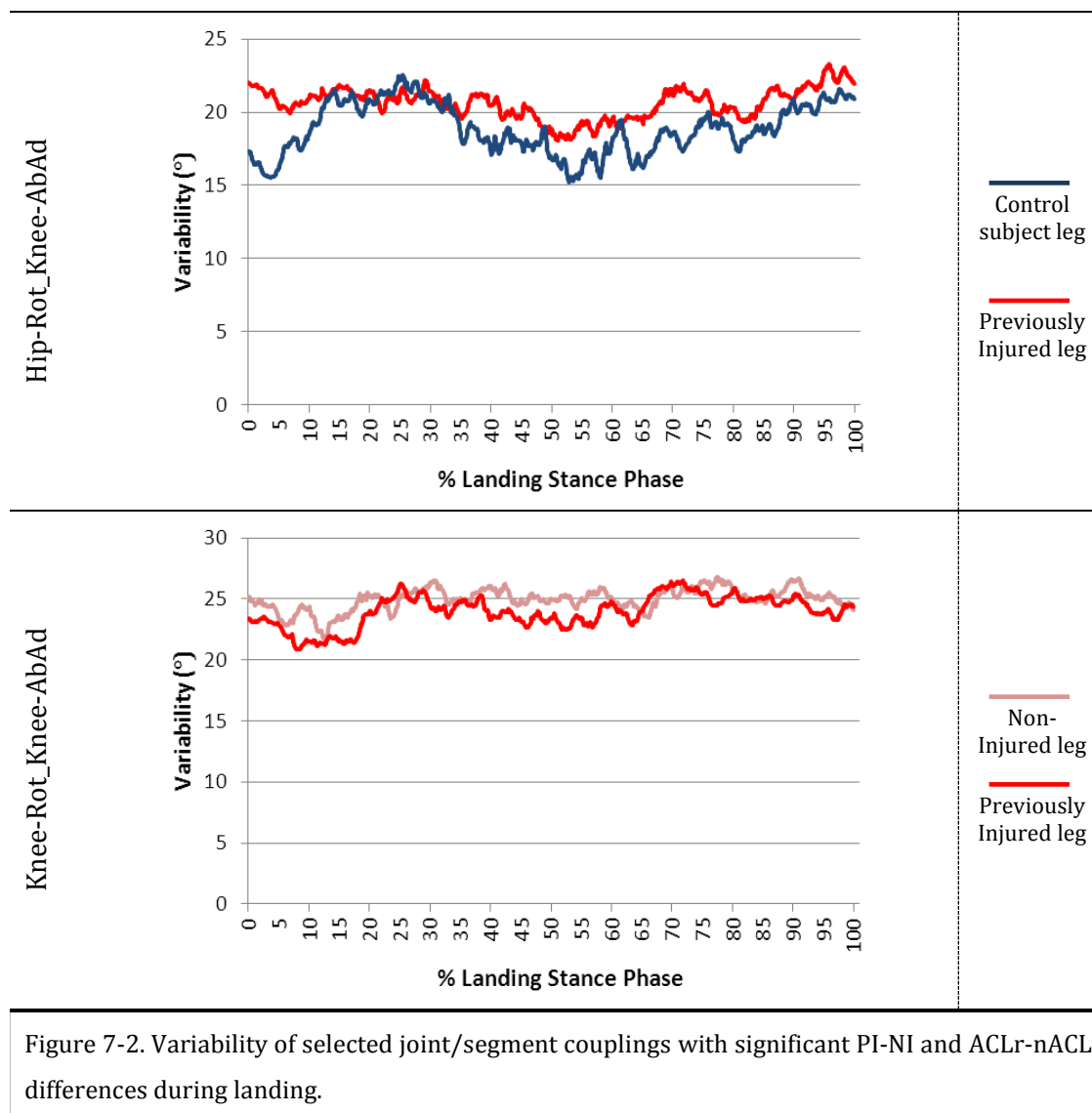
Two couplings showed differences when the PI leg of ACLr subjects was compared to the NI or control leg. The PI leg had increased variability in the hip-rot\_knee-abad coupling when compared to a control leg. Alternatively the PI leg had less variability in the knee-rot\_knee-abad coupling than the NI leg.

Table 7-3. Intra-subject leg differences (Average (± Standard Deviation)) for ACLr and nACL subjects in the variability of the Knee-Rot\_Knee-AbAd coupling.

Average between leg difference (°) for ACLr and nACL subjects during selected regions of the landing stance phase are shown. Group differences (°), Cohen's d and p-values are also presented. The scale for classification of Cohen's d effect size was, 0.2 = **small**, 0.5 = **moderate**, > 0.8 = **large**.

Coupling	ACLR	nACL	Diff	d	p-value
Knee Rotation – Knee Abduction Adduction					
Initial 40ms	-1.5 (2.80)	0.39 (2.77)	-1.89	<b>0.67</b>	<b>0.051</b>
Initial 40%	-1.12 (1.69)	-0.15 (1.92)	-1.28	<b>0.71</b>	<b>0.050</b>
100%	-0.99 (1.10)	-0.13 (1.56)	-0.86	<b>0.64</b>	<b>0.065</b>

These differences between the PI and NI leg were further investigated and compared to the differences present between the PI and NI matched legs on the nACL subjects, see Table 7-3. All intra-subject leg differences reported in the ACLr subjects were significantly different from the intra-subject leg differences reported in the nACL subjects. Figure 7-2 presents the variability of selected joint/segment couplings with significant group differences.



#### 7.4.2 Cutting

Table 7-4 presents a summary of the significant differences present in variability of the 3D joint angles at the knee joint when averaged across two regions of the cutting stance phase. The PI leg had more transverse plane hip and frontal plane knee angle variability than the nACL control leg. The PI leg had less variability in variability of the sagittal plane knee angle when compared to both the NI and nACL control legs.

Table 7-4. Significant differences in the variability 3D knee joint angles between ACLr and nACL subjects during cutting.

Average variability (°) for PI, NI and control legs during 70-100% and 100% of cutting stance phase are shown. Group differences (°), Cohen's d and p-values are also presented. The scale for classification of Cohen's d effect size was, 0.2 = **small**, 0.5 = **moderate**, > 0.8 = **large**.

Cut		ACLR_Pi-NI					ACLR-Control				
		PI	NI	Diff	d	p-value	PI	C	Diff	d	p-value
<b>Internal-External Rotation</b>											
Hip	70-100%						<b>4.92</b>	6.44	1.52	<b>0.74</b>	0.010
	100%						<b>5.06</b>	6.86	1.80	<b>0.95</b>	0.035
<b>Flexion Extension</b>											
Knee	70-100%						5.41	<b>7.17</b>	1.77	<b>0.76</b>	0.015
	100%	5.66	<b>6.90</b>	1.24	<b>0.70</b>	0.044	5.66	<b>6.94</b>	1.30	<b>0.84</b>	0.017
<b>Abduction Adduction</b>											
	100%						<b>2.35</b>	1.67	0.68	<b>0.98</b>	0.008

The difference in the variability of frontal plane knee motion between the PI and NI leg was further investigated and compared to the difference present between the PI and NI matched legs on the nACL subjects, see Table 7-5. The between leg differences reported by the ACLr subjects were not significantly different from the nACL subjects with a p-value >0.1.

Table 7-5. Absolute intra-subject leg differences (Average (±Standard Deviation)) for ACLr and nACL subjects in the variability of sagittal plane knee motion.

Average between leg difference (°) for ACLr and nACL subjects during selected regions of the cutting stance phase are shown. Group differences (°), Cohen's d and p-values are also presented. The scale for classification of Cohen's d effect size was, 0.2 = **small**, 0.5 = **moderate**, > 0.8 = **large**.

	ACLR	nACL	Diff	d	p-value
<b>Knee Flexion Extension</b>					
100%	-1.24 (2.43)	-1.01 (2.16)	-0.24	<b>0.10</b>	0.759



Figure 7-3 presents the variability of 3D knee joint angles that showed significant group differences during cutting.

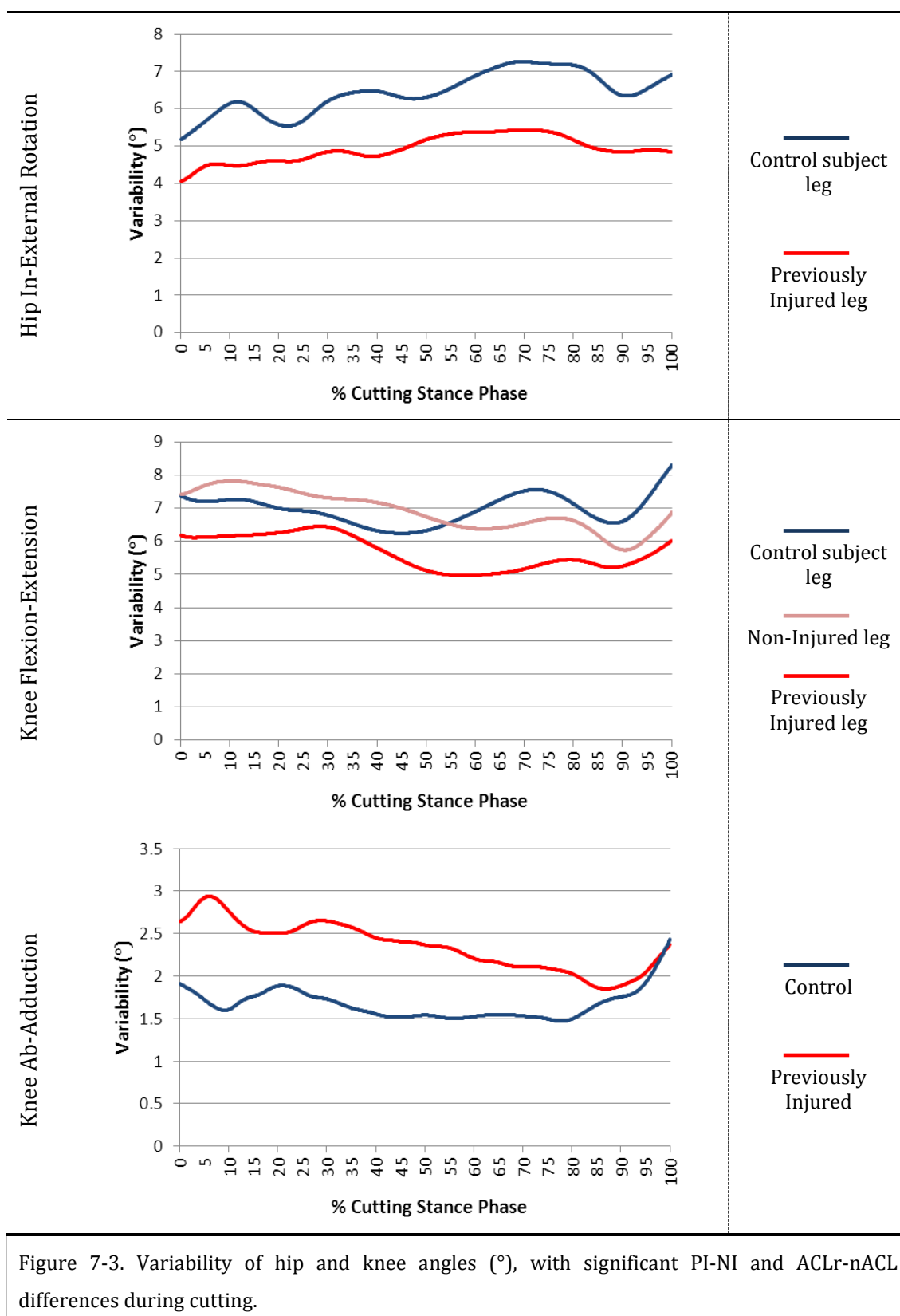


Figure 7-3. Variability of hip and knee angles (°), with significant PI-NI and ACLr-nACL differences during cutting.

Table 7-6 presents a summary of the significant differences present in the variability of selected 3D joint and segment rotation couplings when averaged across two regions of cutting stance.

Table 7-6. Significant differences present in the variability of selected 3D joint and segment angle coupling combinations between ACLr and nACL subjects during cutting.

Average variability (°) for PI, NI and control legs during 70-100% and 100% of cutting stance phase are shown. Group differences (°), Cohen's d and p-values are also presented. The scale for classification of Cohen's d effect size was, 0.2 = **small**, 0.5 = **moderate**, > 0.8 = **large**

Cut		ACLR_Pi-NI					ACLR-Control				
		PI	NI	Diff	d	p-value	PI	C	Diff	d	p-value
Knee-Rot_ Knee-Ab-Ad	100%	<b>23.2</b>	22.4	0.87	<b>0.56</b>	0.048					

The only joint angle coupling that showed significant differences between PI and NI legs and PI and nACL control legs was in the knee-rot\_knee-abad coupling. The PI leg of the ACLr subjects had more variability in this coupling when compared to the NI leg of the ACLr subjects. The difference between the PI and NI leg was further investigated and compared to the difference present between the PI and NI matched legs on the nACL subjects, see Table 7-7. The between leg differences reported by the ACLr subjects were not significantly different from the nACL subjects with a p-value >0.1 and a small effect size.

Table 7-7. Absolute intra-subject leg differences (Average (Standard Deviation)) for ACLr and nACL subjects in the variability of the Knee-Rot\_Knee-AbAd coupling.

Average between leg difference (°) for ACLr and nACL subjects during selected regions of the cutting stance phase are shown. Group differences (°), Cohen's d and p-values are also presented. The scale for classification of Cohen's d effect size was, 0.2 = **small**, 0.5 = **moderate**, > 0.8 = **large**

	ACLR	nACL	Diff	d	p-value
Knee Rotation – Knee Abduction Adduction					
100%	1.59 (1.71)	1.19 (0.75)	0.40	<b>0.32</b>	0.95

Figure 7-4 presents the variability of the knee rotation - knee abduction adduction coupling.

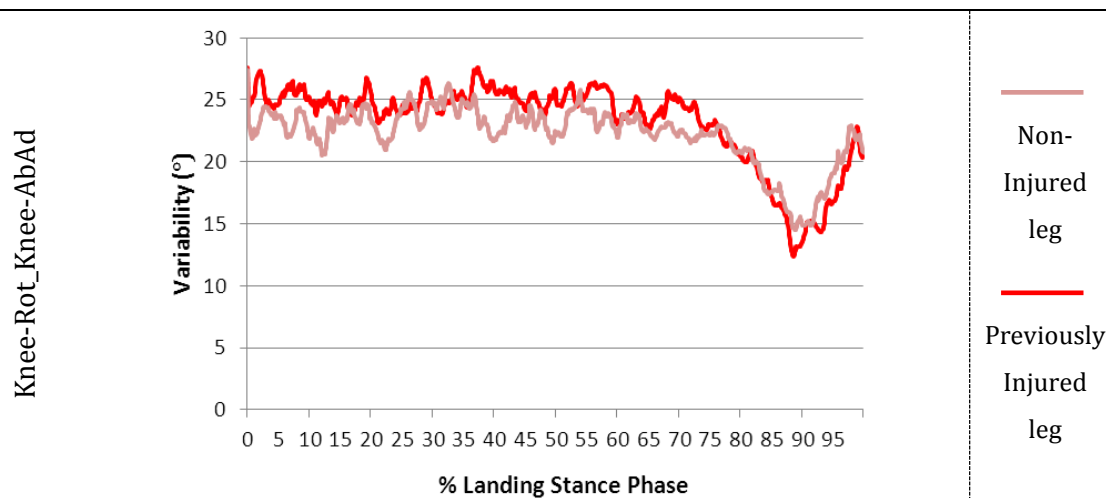


Figure 7-4. Variability of the Knee-Rot\_Knee-AbAd coupling which had significant ACLr-nACL differences during cutting.

## 7.5 Discussion

In the present study, the variability of hip and knee joint kinematics (movement variability), and lower extremity joint and segment coupling (coordination variability), of ACLr individuals was compared against the contralateral non-injured leg and a healthy control (nACL) leg during the performance of a maximal drop-jump land and unanticipated cutting task. Based on the variability hypothesis the previously injured leg of ACLr individuals was expected to demonstrate lower levels of movement and coordination variability when compared to the contralateral non-injured ACLr leg and higher levels of movement and coordination variability when compared to match control nACL leg.

As expected the PI leg of ACLr individuals had lower levels of variability than the NI leg on a number of variables. The PI leg had lower levels of knee-rot\_knee-abad coupling variability than the NI leg during landing. The knee-rot\_knee-abad coupling is especially important when comparing the PI and contralateral NI leg of ACLr individuals. Frontal and transverse plane rotation of the knee joint has been associated with the previously injured leg of ACLr individuals. Increased abduction at the knee joint has been linked with repeated ACL injury (Paterno et al., 2010) in ACLr individuals. Increased knee adduction has also been reported in ACLr individuals (Tashman et al., 2004) and has been linked with higher incidence and faster progression of knee OA (Cerejo et al., 2002). In contrast to the current findings, when these frontal and transverse plane knee joint rotations were investigated in isolation no differences were reported between the PI and NI leg of ACLr subjects (see Chapter 5). The PI leg also demonstrated decreased variability in the flexion and extension patterns of the knee joint when compared to the NI leg during the cutting component of the task. Sagittal plane knee motion has an important role in regulating the shear strain on the ACL with increased knee extension producing more ACL strain (Arms et al., 1984; Markolf et al., 1995). Discrete

measures of knee flexion-extension also failed to distinguish between the PI and NI leg of ACLr individuals (see Chapter 5). The PI leg had higher levels of variability in the knee-rot\_knee-abad coupling than the NI leg during cutting. Although this is in contrast to expectations the increase in variability reported is of minimal magnitude and ES. This between leg PI and NI leg difference was compared to between leg differences reported for nACL subjects, interestingly both groups demonstrated similar differences indicating the increase in variability reported in the PI leg for ACLr individuals is normal and similar to what is reported by nACL subjects. A similar analysis was performed on the between leg differences reported for the knee-rot\_knee-abad coupling during landing and flexion and extension patterns of the knee joint during cutting. The knee abduction adduction coupling was the only variable to show a notable difference between ACLr and nACL subjects when comparing the between leg differences in variability. With moderate effect sizes and p-values < 0.1 the between leg difference in variability in the ACLr subjects was considered markedly different from that demonstrated by the nACL subjects.

This decrease in the variability in the knee-rot\_knee-abad coupling in the PI leg of ACLr individuals in comparison to the NI leg supports our expectations. This is consistent with previous research reporting decreased coordination variability in injured and high risk populations (Hamill et al., 1999; Heiderscheit et al., 2002; Pollard et al., 2005). As outlined the joint motions in this coupling have been related to repeated ACL injury and the development of degenerative joint disease. Lower coordination variability could result in a more repetitive loading pattern which may be linked to cartilage degeneration and premature osteoarthritis in ACLr knees. Levels of coordination variability lower than what is considered optimal may also render the system more predictable and decrease the potential to adapt to perturbations in game situations, increasing the risk of repeated ACL injury in this population. The magnitude of the knee-rot\_knee-abad coupling variability in the PI and NI leg of ACLr subjects is within range of values previously reported for nACL males and females during an unanticipated cutting manoeuvre (Pollard et al., 2005). The differences between the PI and NI leg (4-7% difference) are relatively small when considered alongside reductions in variability noted in injured limbs and populations at increased risk of injury. Heiderscheit et al., (2002) reported thigh rotation-leg rotation coordination variability in the injured limb of patellofemoral pain patients that was ~36% lower than the coordination variability reported in non-injured limb. Pollard et al., (2005) reported various joint/segment coupling combinations in females that were on average 26% (19-29%) lower than the coordination variability reported in the male population. The 2-3% reductions in knee-rot\_knee-abad coupling variability on the PI leg may not be sufficient to result in a repetitive loading pattern and inflexible movement system with the potential for the development of osteoarthritis and the incidence of repeated ACL injury respectively.

As expected the PI leg of ACLr individuals had higher levels of movement and coordination variability than a matched control leg from a nACL individual on a number of variables. During landing the PI leg had more variability in the frontal and transverse plane knee rotations and also the hip-rot\_knee-abad coupling, than a control leg. During cutting the PI leg had more variability in transverse plane hip motion and frontal plane knee motion, than a control leg. Additionally, the PI leg was also reported to have decreased sagittal plane knee motion variability than a control leg during cutting. As previously outlined, frontal and transverse plane knee motion have been previously identified as risk factors for repeated ACL injury and the development of degenerative joint disease. The combination of hip rotation and frontal plane knee motion has also been previously linked with ACL injury. Ireland et al., (1999) referred to it as the “position of no return” when an athlete enters a posture with an internally rotated hip and abducted knee.

The increased variability in transverse hip and knee joint rotations, frontal plane knee joint rotations and hip-rot\_knee-abad coupling are unlikely to increase risk of degenerative joint disease; it would act to distribute loading and avoid repetitive loading on isolated areas of knee joint cartilage. The risk of re-injury on the other hand may be increased by these higher levels of variability. If the levels of variability reported by the nACL population are considered optimal, increases in variability above this level may render a movement system noisy and unpredictable, potentially acting to decrease the ability to adapt to perturbations in game situations (Stergiou and Decker, 2011). Increased movement variability has been previously reported in ACLr populations during walking gait and a balance task (Kiefer et al., 2008; Moraiti et al., 2010). ACLr individuals were thought to feel “secure” enough to add extra movement during gait however proprioceptive deficits may result in an unstable movement system generating higher levels of movement variability. The contrasting findings for the sagittal plane motion at the knee, with less movement variability on the PI leg compared to the control leg, may indicate that if these proprioceptive deficits are present, may have a larger effect on transverse and frontal plane knee motions.

The magnitude of the movement variability in the knee and hip joint rotations during landing and cutting, are within the range of values reported in Chapter 4. As discussed in Chapter 4 the impact of a  $1.45^\circ$  difference in transverse plane knee angle variability, as seen between the PI and control leg during the initial 40% of landing, may seem negligible. If considered as a percentage of average range of transverse plane knee angle displacement, it is equal to a 16% difference. The differences between the PI and control leg in the other knee and hip angles were of similar magnitude ( $9 \pm 4\%$ ), and when considered as a percentage of the average variability, the average difference was ranged from 13% to a 55% difference ( $37 \pm 14\%$ ). The variability of joint angles has not been analyzed and reported in this manner previously which makes comparison with previous research difficult, the percentage

differences are however consistent with previous investigations (Kiefer et al., 2008; Moraiti et al., 2010). The hip-rot\_knee-abad coupling variability values are within the range of values previously reported for nACL males and females during an unanticipated cutting manoeuvre (Pollard et al., 2005). The differences between the PI and NI leg (9% difference) are however relatively small when considered alongside the reductions in variability noted by Moraiti et al., and Kiefer et al., (2008; 2010) (~14% and 23% respectively). This level of variability is not thought to be sufficiently higher than the optimal reported by the control leg and is unlikely to result in the previously described inflexible movement system.

## **7.6 Conclusion**

The PI leg of ACLr individuals was shown to have less movement and coordination variability than the contralateral NI leg and more movement and coordination variability than the nACL control, during a drop-land and cut manoeuvre. The PI leg had less knee-rot\_knee-abad coupling and sagittal plane knee motion variability than the NI leg for during landing and cutting respectively. The PI leg had more frontal and transverse plane knee motion and the hip-rotation knee abduction adduction coupling variability than the control leg during landing, and more transverse plane hip motion and sagittal plane knee motion variability during cutting. This occurred during the majority regions of landing and cutting stance in the majority of cases.

Lower coordination variability on the PI leg, in comparison to the NI leg could result in a more repetitive loading pattern linked to cartilage degeneration and premature osteoarthritis in ACLr knees. Levels of coordination variability lower than what is considered optimal may render the system more predictable and decrease the potential to adapt to perturbations in game situations and increase the risk of repeated ACL injury in this population. The small differences reported however minimize these risks. The higher levels of movement variability present in the PI leg may also decrease the potential to adapt to perturbations in game situations via an unpredictable movement system with variability above what is considered optimal.

Further research is necessary to address whether levels of coordination or movement variability which are above or below what is considered optimal, actually result in a less adaptable movement system. This could be implemented by comparing groups who demonstrate low and high levels of coordination variability, and assess their response to an unanticipated perturbation such as landing on an uneven surface or being pulled off balance just prior to landing, this research may also help in identifying optimal levels of variability. Forward dynamics musculoskeletal modelling may also further inform the hypothesis on the role of movement variability and the potential occurrence of repeated micro trauma, thought to weaken the cartilage resulting in degenerative joint disease such as osteoarthritis.

## 7.7 References

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## **CHAPTER 8**

### **THESIS CONCLUSIONS**

## 8.1 Conclusions

As outlined in Chapter 1, ACL injury has become a constant adversary of those involved in dynamic team sports. There is a plethora of research in the area examining the biomechanical function of females and previously injured ACL injury subjects in a means to identifying the potential risk factors that increase their risk of ACL injury and re-injury or the development of OA respectively. Studies investigating the gender bias in ACL injury have generally been more advanced in the movement tasks utilised, some of which replicate match demands such as opposition players, movement intensity and level of anticipation required. Studies investigating rehabilitated ACL injured subjects however tend to utilise more basic movement tasks such as walking, and stair descent. There has been some development in this area however with recent investigations utilising combinations of landing and pivoting. There are few investigations in either research area however that adequately assesses the testing population via a movement task that is similar to what would occur in a game situation, unanticipated, at match intensity and producing data for both legs.

The aim of this thesis as outlined in Chapter 1 was twofold; firstly determine potential biomechanical risk factors that contribute to the disproportionate incidence of non-contact ACL injury in females and secondly determine potential biomechanical risk factors that contribute to the increased risk of ACL re-injury and the development of osteoarthritis in athletes following surgical reconstruction of the ACL, rehabilitation and return to sport. This was implemented via the measurement of joint kinematics and kinetics, joint coordination and both movement and coordination variability during a maximal drop-jump land and unanticipated cutting task.

In terms of the biomechanical risk factors that may contribute to the disproportionate incidence of non-contact ACL injury in females; Chapter 3 and 4 addressed this question.

- Chapter 3 compared the joint kinematics and kinetics of male and female athletes that were measured during a maximal drop-jump land and unanticipated cutting task. The only notable gender difference was at the hip joint where females had increased internal rotation at the hip when compared to the male athletes.
- Chapter 4 compared the variability of lower limb kinematics and coordination of male and female athletes that were measured during a maximal drop-jump land and unanticipated cutting task. Females were reported to have lower levels of both movement and coordination variability in comparison to the male athletes.

The gender differences in transverse plane hip kinematics and in the variability of lower limb coordination were only reported in the landing component of the maximal drop-jump land and unanticipated cutting task. This highlights the risk associated with the tasks

involving rapid deceleration. It may be the case that the increased internal rotation demonstrated in the female subjects limits the range of movement options available to the female athletes therefore, reducing the variability at which they coordinate their lower limb segments and joints. Aside from whether this relationship exists female subjects have decreased movement and coordination variability which if considered in line with the variability –injury hypothesis suggested by Stergiou and Bates, (1997) would increase their risk of injury.

The differences reported in movement and coordination variability which are present without any differences in discrete lower limb biomechanics indicates the movement and or coordination variability may be a key biomechanical parameter for future study

In terms of the biomechanical mechanisms that may contribute to the increased risk of ACL re-injury and the development of osteoarthritis in athletes following surgical reconstruction of the ACL, rehabilitation and return to sport; Chapter 5, 6 and 7 addressed this question.

- Chapter 5 compared the joint kinematics and kinetics of the previously injured leg of ACLr athletes against their contralateral non-injured leg and a nACL control leg during a maximal drop-jump land and unanticipated cutting task. No significant differences were reported between the previously injured and non-injured leg. The most notable differences reported between the previously injured and nACL control leg were increased hip flexion, increased internal knee abduction moment and increased transverse plane knee ROM on the previously injured leg.
- Chapter 6 compared the lower limb coordination of the previously injured leg of ACLr athletes against their contralateral non-injured leg and a nACL control leg during a maximal drop-jump land and unanticipated cutting task. No significant differences were reported between the previously injured and non-injured leg. All differences reported between the previously injured and nACL control leg emerged from the contribution of the hip and thigh in various couplings.
- Chapter 7 compared the variability of lower limb kinematics and coordination of the previously injured leg of ACLr athletes against their contralateral non-injured leg and a nACL control leg during a maximal drop-jump land and unanticipated cutting task. The previously injured leg of ACLr athletes was shown to have lower levels of coordination variability than the contralateral non-injured leg and higher levels of movement and coordination variability than the nACL control leg.

The only investigation to differentiate between the previously injured and contralateral non-injured leg of the ACLr subjects was that reported in Chapter 7. The variability of the knee rotation - knee abduction adduction coupling was significantly lower in the previously

injured leg when compared to the non-injured leg. This is interesting considering that no differences were reported in either the joint mechanics or lower limb coordination. The combination of frontal and transverse plane motion at the knee joint is also interesting given their association with the development of OA, and their presence as factors differentiating the ACLr and nACL populations in Chapter 5. It was suggested in Chapter 5 that the increased transverse plane ROM and internal abduction moment difference between the previously injured ACLr leg and nACL control leg could not be the sole contributor to the increased risk of OA in ACLr populations due to the fact that these differences were not reported between the ACLr subjects previously injured and non-injured legs. It was thought that the damage present in the joint due to the initial injury and surgery combined with these movement patterns which are present on both ACLr legs would increase the risk of OA on the previously injured leg. It may however be the case that the decreased variability in the knee rotation - knee abduction adduction coupling on the previously injured leg may be the factor that in combination with the increased transverse plane knee ROM and internal knee abduction moment increases the risk of OA development. Overall from all three chapters proximal control from the hip is highlighted as a factor differentiating ACLr and nACL groups.

The increased movement and coordination variability in the ACLr subject group was suggested to increase the risk of ACL re-injury by decreasing the ability to adapt to movement perturbations in match situations, this combined with the altered coordination patterns and joint kinematics and kinetics present in the ACLr subject group indicate that the ACLr group decelerate from a landing and change direction in a different manner from the nACL group. It is unclear if these differences will increase their risk of re-injury, but they are alterations that must be further investigated to identify if they were a precursor to the injury or the result of it and its treatment.

### **8.3 Practical Applications**

The potential benefits of movement and coordination variability for the reduction of injury risk have been identified in females and previously ACL injured subjects. This merits consideration in the planning of sports fitness and skill training, injury prevention prehabilitation programmes and rehabilitation programmes post ACL reconstruction. Coaches should reduce the emphasis on exact repetition of movement and sport skills, as variability in the repeated execution of a skill may decrease the risk of injury and repeated stress on then lower limbs. The inclusion of variability over a training cycle may also limit repeated stress and reduce injury risk. Finally including perturbations and variability into training, prehabilitation or rehabilitation programmes may increase an athlete's movement pattern repertoire and therefore their ability to react to perturbations in match situations.

Both the female and ACLr populations had altered kinematics at the hip joint. Therefore increasing proximal control at the hip joint has the potential to reduce the incidence of ACL injury re-injury and the development of OA. This could be implemented through strength training of the core, and gluteal musculature on both dominant and non-dominant sides.

#### **8.4 Future Research Recommendations**

A number of future research recommendations in terms of ACL injury risk factors for females can be drawn from the findings of chapters three and four.

In Chapter three, females were shown to have increased internal rotation angles at the hip during landing. This finding adds to the current body of evidence reporting gender differences in transverse plane hip kinematics. In terms of future research the source of this gender difference in internal hip rotation merits further investigation. If the source of the increased internal rotation in females can be identified it could be targeted in ACL injury prevention programs. Two such potential sources of the increased internal rotation in females are muscular weakness or tightness in the gluteal region. Forward dynamics musculoskeletal modeling investigations could be utilised to assess the role of muscle strength or tightness on hip kinematics.

Chapter four, reports that females have lower levels of movement and coordination variability when compared to males. These findings support the previous work of Pollard et al., (2005). Levels of movement and coordination variability has received limited research attention in the ACL injury risk factor literature and the findings of this chapter highlight a key risk factor for ACL injury that strongly merits further research and dissemination to surgeons and practitioners. Identifying a prospective link between movement and/or coordination variability and the incidence of ACL injury would strengthen the argument for including movement and/or coordination variability as a risk factor for ACL injury. Future research investigating the effect of the practical applications outlined in the previous section as part of a prevention program in a female population would be another logical progression of the findings of chapter five. The use of perturbation training as a tool for increasing movement and coordination variability and concurrently increasing ability to deal with movement perturbations in game situations may be an interesting avenue of initial investigation. Investigating the relationship between transverse plane rotation at the hip and associated levels of movement and coordination variability may also be useful and may highlight a potential avenue for intervention for increasing levels of movement and coordination variability.

Further research recommendations for an ACL rehabilitated population can be drawn from the findings of chapters five, six and seven.

In Chapter five and six, there was no difference between the previously injured and contralateral leg of ACLr subjects. The ACLr subjects did have increased hip flexion knee abduction moment and knee rotation range of motion and different coordination patterns when compared to non-injured control subjects however. These findings are an important addition to previous research in an ACLr population, as the task utilised was unique and more demanding than what has been utilised previously. Development of this line of research where ACLr subjects are tested in demanding and match specific situations is essential. This needs to be done for the purpose of research investigations, but also for patient screening prior to returning to sport. The successful completion of a rehabilitation program is not a sufficient indicator that a patient is ready to return to sport. The addition of isometric strength testing and some functional hopping tests is still inadequate as these tests do not accurately represent the demands that will be placed on the individual in a game situation. It seems that the femur and hip are the source of the majority of the ACLr nACL differences. Future research investigations should address hip control and strength as potential training methodology for decreasing transverse plane knee motion and knee abduction moment and potentially the occurrence of OA and re-injury.

Chapter seven was the only chapter to report differences between the previously injured and contralateral legs in an ACLr population. This is very important when considering that in the majority of cases OA only occurs in the previously injured leg. The lower levels of movement and coordination variability present in the previously injured leg is therefore a likely risk factor for the progression of OA in this population. Tracking of movement and coordination variability during a match specific task as it progresses post-surgery and rehabilitation may give some novel information on the progression of OA and may indicate potential avenues for intervention. Differences were also reported between the ACLr and nACL population. The ACLr population was reported to have higher levels of movement and coordination variability than the control subjects. It is important that future research address the scale of movement and coordination variability present in a variety of populations. There are currently no guidelines for what is high, low or optimal in terms of the amount of variability in the repetition of movement or coordination of a given task. This information would inform other research investigations which aim to modify levels of variability toward a more healthy state. The investigation of movement or coordination variability measured from data collected via methods such as high-speed biplane radiography where a more accurate measure of bone motion is quantified, would provide further endorsement of injury variability hypothesis defined in this chapter and chapter three. The differences present between the nACL and ACLr groups in movement and coordination variability, lower limb coordination and joint kinematics and kinetics need to be further investigated. It is important to establish whether these differences were the cause or consequence of the ACL injury and

its treatment. Prospective studies and musculoskeletal modelling as detailed for Chapter three may be a useful means of investigating this concept.

In terms of the biomechanical risk factors that may contribute to the increased risk of ACL re-injury and the development of osteoarthritis in athletes following surgical reconstruction of the ACL, rehabilitation and return to sport. Increased transverse plane knee ROM and internal knee abduction moment in combination with decreased variability in the knee rotation - knee abduction adduction coupling on the previously injured leg is thought to, increase the risk of OA in the ACLr population. As mentioned for the female population above programmes, whether prevention or rehabilitation programmes, designed to increase the movement and coordination variability may be useful for decreasing the risk of OA development in this population.

#### **8.4 Reference**

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## **CHAPTER 9**

## **APPENDICES**

## APPENDIX A

Appendix A is also available to download at:

<http://dl.dropbox.com/u/32326398/Appendices.zip>

*Appendix A1 - Video of maximal drop-jump land and unanticipated cutting task.*

Also available online at <http://bit.ly/w4uVLJ>

*Appendix A2 – Pictures of experimental Set-up.*

*Appendix A3– Testing Documentation*

*Appendix A4 – SPSS Statistical Output Files*



## **APPENDIX B - Related Publications**

### *International Society of Biomechanics in Sports Conferences*

- Breen Sarah, Hanlon Michael, 'A COMPARISON OF VARIABILITY IN GROUND REACTION FORCE AND KNEE ANGLE PATTERNS BETWEEN MALE AND FEMALE ATHLETES' XXVII International Symposium on Biomechanics in Sports, Limerick, Ireland. 2009
- Breen Sarah, Harrison Drew, Kenny Ian, 'HIP ROTATION RANGE OF MOTION AND ITS IMPACT ON LOWER LIMB ALIGNMENT ON LANDING' XXVIII International Symposium on Biomechanics in Sports. Marquette, Michigan, USA. 2010
- Breen Sarah, Harrison Drew, Kenny Ian, 'JOINT KINEMATICS AND KINETICS OF LANDING IN ACL REHABILITATED VOLUNTEERS' XXIX International Symposium on Biomechanics in Sports. Porto, Portugal. 2011
- Breen Sarah, Harrison Drew, Kenny Ian, 'THE ROLE OF PREVIOUS ANTERIOR CRUCIATE LIGAMENT INJURY ON THE VARIABILITY OF JOINT KINEMATICS AND COORDINATION DURING A MATCH SPECIFIC LAND-CUT TASK' XXX International Symposium on Biomechanics in Sports. Melbourne, Australia. 2012

### *PE PAYS - A Shared vision for Physical Education, Physical Activity and Youth Sport*

- Breen Sarah, Harrison Drew, Kenny Ian 'Injury Incidence in Juvenile Ladies Gaelic Football 2008-2009' PE PAYS - A Shared vision for Physical Education, Physical Activity and Youth Sport 2010, Limerick, Ireland

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