Development of an anterior cruciate ligament injury prevention programme based upon the biomechanics of cutting activities.

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Thesis submitted for the award of a Ph.D

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Declaration

I hereby certify that this material, which I now submit for assessment on the programme of study leading to the award of PhD is entirely my own work, and that I have exercised reasonable care to ensure that the work is original, and does not to the best of my knowledge breach any law of copyright, and has not been taken from the work of others save and to the extent that such work has been cited and acknowledged within the text of my work.

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Dedicated to Rowan and Mam; *i gcónaí liomsa.*

Table of Contents

List of Abbreviations

ACL	Anterior cruciate ligament
ACSM	American college of sports medicine
AD	Adolescent
ADP	Adenosine diphosphate
AE	Athlete exposure
ANOVA	Analysis of variance
ANT	Anticipated
ASIS	Anterior superior iliac spine
ATP	Adenosine triphosphate
BW	Body weight
CAI	Chronic ankle instability
CI	Confidence interval
CON	Control
COP	Centre of pressure
DCS	Dynamic core stability
DPA	Discrete point analysis
EMG	Electromyography
ES	Effect size
GRF	Ground reaction force
HIIP	High intensity, intermittent exercise protocol
HR	Heart rate
IC	Initial contact
ICC	Intraclass correlation
INT	Intervention
IPP	Injury prevention programme
LESS	Landing error scoring system
MCL	Medial collateral ligament
MVIC	Maximal voluntary isometric contraction
NMT	Neuromuscular training
NS	Non significant

OR	Odds ratio
PAR-Q	Physical activity readiness questionnaire
PCL	Posterior cruciate ligament
PREAD	Preadolescent
RM	Repetition maximum
ROC	Receiver operating characteristic
ROM	Range of motion
RPE	Rate of perceived exertion
RTP	Return to play
SD	Standard deviation
SDMT	Symbol digit modalities test
SEBT	Star excursion balance test
SEM	Standard error of measurement
SPM	Statistical parametric mapping
SRC	Sports related concussion
TTS	Time to stabilisation
UNA	Unanticipated
VDJ	Vertical drop jump
Vs	Versus
WA	Weight acceptance

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Thesis Overview and Guidelines

This thesis has been written using the guidelines for PhD by publication. Therefore, the review of literature (Chapter Two) has been restricted to a summary of the relevant areas. Each study has been presented within its own chapter in the format of a journal paper (Chapters Three to Eight). The overall conclusion and future recommendations are included in Chapter Nine.

Abstract

Author: Enda F. Whyte

Title: Development of an anterior cruciate ligament injury prevention programme based upon the biomechanics of cutting activities.

Background: Anterior cruciate ligament injuries (ACL) frequently occur during cutting activities when fatigued or responding to the sporting environment. ACL injuries lead to profound short and long term consequences, making the prevention of ACL injuries critically important. However, the effect of fatigue and anticipation on cutting biomechanics is not well understood. Furthermore, there is limited research on the effects of interventions to improve the biomechanics of unanticipated cutting activities.
Aims: To determine the effects fatigue and anticipation on the biomechanics of side and crossover cutting. Secondly, to develop and assess the efficacy of an exercise programme designed to improve cutting technique.
Methods: Three studies determined the effect of a high intensity, intermittent exercise protocol (HIIP) on dynamic postural control, neurocognitive function and the biomechanics of the vertical drop jump, respectively. Two subsequent studies investigated the effects of the HIIP and anticipation on the biomechanics of crossover and side cutting, respectively. Finally, a randomised controlled trial assessed the effect of a dynamic core stability programme on the biomechanics of anticipated and unanticipated cutting.
Findings: The HIIP had detrimental effects on dynamic postural control and neurocognitive function but not on vertical drop jump biomechanics. The combination of the HIIP and unanticipation did not increase the magnitude of biomechanical risk factors for ACL injuries during side and crossover cutting. However, unanticipation increased the magnitude of certain risk factors for ACL injuries, particularly related to trunk kinematics, during cutting. Although the dynamic core stability programme did not alter trunk kinematics, it reduced the magnitude of a small number of risk factors for ACL injuries during cutting, particularly during anticipated side cutting.
Conclusion: The magnitude of biomechanical risk factors for ACL injuries is greater during unanticipated compared with anticipated cutting. A dynamic core stability programme has small, beneficial effects on cutting biomechanics.

List of Publications

Journal Publications

1. Whyte E, Burke A, White E, Moran K. A high-intensity, intermittent exercise protocol and dynamic postural control in men and women. *J Athl Train*. 2015. doi: 10.4085/1062-6050-49.6.08 [doi].
2. Whyte EF, Gibbons N, Kerr G, Moran KA. Effect of a high-intensity, intermittent-exercise protocol on neurocognitive function in healthy adults: Implications for return-to-play management after sport-related concussion. *J Sport Rehabil*. 2015; Technical Notes 16:2014-0201. doi: 10-1123/jsr.2014-0201 [doi].
3. Whyte EF, Kennelly P, Milton O, Richter C, O'Connor S, Moran KA. The effects of limb dominance and a short term, high intensity exercise protocol on both landings of the vertical drop jump: Implications for the vertical drop jump as a screening tool. *Sports Biomech*. 2017:1-13. doi: 10.1080/14763141.2017.1371215 [doi].
4. Whyte EF, Richter C, O'Connor S, Moran KA. The effect of high intensity exercise and anticipation on trunk and lower limb biomechanics during a crossover cutting manoeuvre. *J Sports Sci*. 2018; 36(8):889-900. doi: 10.1080/02640414.2017.1346270 [doi].
5. Whyte EF, Richter C, O'Connor S, Moran KA. Effects of a dynamic core stability program on the biomechanics of cutting maneuvers: A randomized controlled trial. *Scand J Med Sci Sports*. 2018; 28(2):452-462. doi: 10.1111/sms.12931 [doi].

Oral Presentations

1. Richter C, Moran K, Whyte E. (2015) Effects of fatigue and anticipation on the kinematics and kinetics during a side cut 25th Congress of the International Society of Biomechanics
2. Enda Whyte, Kieran Moran, Chris Richter. (2015) Cross cut biomechanics are affected to a greater extent when performed in an unanticipated condition compared to a fatigued condition 25th Congress of the International Society of Biomechanics

Author Contribution to Publications

This thesis includes five original, published manuscripts in line with the overall aims of the thesis which are to investigate (1) the effects of a high intensity, intermittent exercise protocol on the biomechanics of both landings of the vertical drop jump, (2) the effects of both anticipation and a high intensity, intermittent exercise protocol on the biomechanics of cutting manoeuvres and (3) to determine the effectiveness of an intervention programme designed to address the main detrimental biomechanical effects of both anticipation and a high intensity, intermittent exercise protocol on the biomechanics of cutting manoeuvres. The ideas, development and writing of the papers and overall thesis were my responsibility and completed by me, under the supervision of Dr. Kieran Moran in the School of Health and Human Performance. Co-authors have been included in the published papers to acknowledge the collaborative and team-based research. My contribution to the individual studies involved the following:

Chapter	Publication Title	Publication Status	Nature and Extent of candidate's contribution
3	A high-intensity, intermittent exercise protocol and dynamic postural control in men and women	Published in Journal of Athletic Training (2015). Apr;50(4):392-9 doi: 10.4085/1062-6050-49.6.08	First author, key ideas, assisted data collection and completed analysis, manuscript development and writing up.

4	Effect of a high-intensity, intermittent-exercise protocol on neurocognitive function in healthy adults: Implications for return-to-play management after sport-related concussion	Published in Journal of Sport Rehabilitation (2015); Technical Notes 16:2014-0201. doi: 10-1123/jsr.2014-0201	First author, key ideas, assisted data collection and completed analysis, manuscript development and writing up.
5	The effects of limb dominance and a short term, high intensity exercise protocol on both landings of the vertical drop jump: Implications for the vertical drop jump as a screening tool.	Published in Sports Biomechanics (2017):1-13. doi: 10.1080/14763141.2017.1371215 [doi].	First author, key ideas, assisted data collection and completed analysis, manuscript development and writing up.
6	The effect of high intensity exercise and anticipation on trunk and lower limb biomechanics during a crossover cutting manoeuvre.	Published in Journal of Sports Sciences (2018); 36(8):889-900. doi: 10.1080/02640414.2017.1346270	First author, key ideas, data collection and completed analysis, manuscript development and writing up.

7	An Investigation of the Effects of High Intensity, Intermittent Exercise and Unanticipation on Trunk and Lower Limb Biomechanics during a Side Cutting Manoeuvre using Statistical Parametric Mapping	Published in the Journal of Strength and Conditioning Research (2018); 32(6): 1583-1593. doi: 10.1519/jsc.00256	First author, key ideas, data collection and completed analysis, manuscript development and writing up.
8	Effects of a dynamic core stability program on the biomechanics of cutting maneuvers: A randomized controlled trial	Published in Scandinavian Journal of Medicine and Science in Sports (2018); 28(2):452-462. doi: 10.1111/sms.12931 [doi].	First author, key ideas, data collection and completed analysis, manuscript development and writing up.

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Signed: ________________________ Date: ________________________

Dr. Kieran Moran (supervisor)

Chapter 1 Introduction

1.1 Background and Justification

It has been estimated that up to 250,000 anterior cruciate ligament (ACL) injuries occur annually in the USA with a cost of over $2 billion (Silvers and Mandelbaum, 2007). Unfortunately, exact figures for Ireland do not currently exist. ACL injuries are one of the most common (Majewski et al., 2006) and detrimental knee injuries (Hewett et al., 2013) leading to profound short- and long-term consequences. These include an increased incidence of osteoarthritis by middle age regardless of surgical intervention (Oiestad et al., 2009; Oiestad et al., 2010), high medical costs (de Loes et al., 2000), and reduced likelihood to return to (Ardern et al., 2014; Shah et al., 2010) or continue (Rugg et al., 2014) pre-injury levels of physical activity. Therefore, ACL injury prevention programmes (IPPs) are of primary importance.

In order to develop ACL IPPs, the mechanisms of injury and risk factors for that injury should be understood before interventions are developed and rolled out (Finch, 2006). The majority of ACL injuries are noncontact in nature and typically occur during deceleration activities such as landing from a jump or cutting (Koga et al., 2010; Krosshaug et al., 2007). During the injury mechanism, the load applied to the ACL exceeds the integrity of the tissue itself, leading to rupture (Shultz, Schmitz, Benjaminse et al., 2015). Altered biomechanics of the trunk and lower limb are associated with ACL injuries and have been proposed to contribute to excessive ACL loading. As they are potentially modifiable, much of the research on the development of ACL IPPs has focussed on such modifiable biomechanical risk factors. Examples of such biomechanical factors include increased knee valgus angle and greater adductor moment[1] (Hewett et al., 2005), reduced trunk control (Zazulak et al., 2007), and altered muscle-firing patterns (Zebis et al., 2009). While it has been demonstrated prospectively that increased knee valgus angle (Hewett et al., 2005),

[1] An internal joint moment is a measure of the body's resistance (for example muscular contractions) to external moments generated by the ground reaction force and inertial forces. All joint moments are reported as "internal" moments in this thesis. When discussing findings from previous studies which originally reported results as "external" moments, the author has converted these results to "internal" moments to ensure consistency throughout.

deficits in lateral trunk control (Zazulak et al., 2007) and greater vertical ground reaction force (GRF) (Leppanen et al., 2017) during landing activities were strong predictors of noncontact ACL injury, our understanding of risk factors for ACL injuries specifically during side and crossover cutting activities is less well understood.

1.2 The Effects of Fatigue and Anticipation on ACL Injury Risk

Fatigue has been proposed to be a factor in sustaining ACL injuries as fatigue protocols have been shown to negatively affect biomechanical risk factors for ACL injuries (Kernozek et al., 2008; McLean et al., 2007; Zebis et al., 2010). However, the relationship between fatigue and the occurrence of ACL injuries is not clear as it has been found that ACL injuries occur more frequently in the first half compared with the second and more frequently in the first fifteen minutes of each half (Walden et al., 2011). Therefore, temporary fatigue observed in field sports following bouts of high intensity, intermittent exercise, may provide us with a greater understanding of the effects of fatigue on risk factors for ACL injuries and assist in the refinement of ACL IPPs.

The majority of noncontact ACL injuries occur when an athletes is responding to the sporting environment (guarding a player, receiving or passing a ball) at the time of noncontact ACL injuries (Boden et al., 2009). Furthermore, there is a much higher incidence of ACL injury in competition rather than training (Hootman et al., 2007). These observations have led to the suggestion that performance of high risk activities such as landing and cutting, when there is little time to anticipate and plan the task, increases the likelihood of adopting suboptimal biomechanics. This in turn can lead to excessive loading of the ACL and potential injury (Borotikar et al., 2008). Laboratory based studies have found that performance of cutting activities in the unanticipated condition results in the adoption of lower limb biomechanics that may increase the loading of the ACL (Kim et al., 2014; Mornieux et al., 2014).However, there is a deficiency in research investigating the effects on trunk kinematics during cutting, particularly crossover cutting.

ACL injuries can occur during sports where cutting tasks must be completed in the presence of fatigue and during unanticipated conditions. Therefore, the combined effects of fatigue and anticipation on biomechanical risk factors during side cutting have been investigated with equivocal results. The performance of unanticipated side cutting following exhaustive exercise protocols resulted in more pronounced detrimental biomechanics in terms of ACL injuries compared to the effect of fatigue or unanticipation alone (Borotikar et al., 2008; McLean and Samorezov, 2009). In contrast, this has not been supported during longer duration running protocols that are designed to replicate the physiological effects of soccer (Collins et al., 2016; Khalid et al., 2015). However, it is important to note that the combined effects of anticipation and fatigue, using a high intensity, intermittent exercise protocol, has not been investigated on cutting biomechanics. Furthermore, the effects on trunk kinematics have not been investigated during side cutting. Critically, there have been no investigations on the combined effects of anticipation and fatigue on biomechanical risk factors for ACL injuries during crossover cutting.

1.3 Exercise Intervention Programme to Address Biomechanical Risk Factors for Anterior Cruciate Ligament Injuries during Cutting Tasks

ACL IPPs have been found to successfully reduce injury rates in younger, female athletes (Myer et al., 2013; Taylor et al., 2015). Successful ACL IPPs contain lower limb strengthening exercises and balance exercises, and technique training, with an emphasis on landing technique (Sugimoto, 2015; Myer et al., 2013; Taylor et al., 2015). Interestingly, successful ACL IPPs that include strengthening and hip and trunk control exercises have greater efficacy than IPPs that do not include such exercises (Sugimoto et al., 2015). This may not be surprising given the association between trunk control (Zazulak et al., 2007), hip strength (Khayambashi et al., 2016) and ACL injury. Although ACL IPPs have been shown to ameliorate lower limb biomechanics that may increase ACL loading during landing tasks (Lopes et al., 2017), their effect is not clear during cutting tasks with limited understanding of their effects on trunk kinematics (Pappas et al., 2015). Indeed, the specific effects of core stability programmes on biomechanical risk factors for noncontact ACL injury is poorly understood (Norcross et al., 2016; Pappas et al., 2015). Only one study has examined this during cutting activities (Jamison, McNeilan et al., 2012). The study by Jamison et al. (Jamison et al., 2012) did not find any effect on trunk kinematics or knee joint loading, which may have been due to the absence of dynamic components to the programme. Currently, no randomised controlled trial has investigated the effect of a dynamic core stability programme on the biomechanics of unanticipated and anticipated side and crossover cutting manoeuvres.

1.4 Aims and Objectives

Overall study aims:

1. To investigate the effects of a high intensity, intermittent exercise protocol on the biomechanics of both landings of the vertical drop jump.

2. To investigate the effects of both anticipation and a high intensity, intermittent exercise protocol on the biomechanics of cutting manoeuvres.

3. To determine the effectiveness of an intervention programme designed to address the main detrimental biomechanical effects of both anticipation and a high intensity, intermittent exercise protocol on the biomechanics of cutting manoeuvres.

Overall study objectives

1. To determine the effects of a high intensity, intermittent exercise protocol on dynamic balance

2. To determine the effects of a high intensity, intermittent exercise protocol on measures of neurocognitive function

3. To determine the effects of a high intensity, intermittent exercise protocol on the biomechanics of the vertical drop jump

4. To determine the effects of both anticipation and a high intensity, intermittent exercise protocol on the biomechanics of crossover cutting manoeuvres

5. To determine the effects of both anticipation and a high intensity, intermittent exercise protocol on the biomechanics of side cutting manoeuvres

6. To develop and determine the effectiveness of an intervention programme in addressing any identified, major biomechanical deficits during cutting activities

Chapter 3 (Study 1)

Aim: To determine the effect of a high intensity exercise protocol on dynamic balance in males and females

Objectives:

1. To determine the test-retest reliability of the star excursion balance test

2. To determine the effect of a high intensity, intermittent exercise protocol on markers of exertion

3. To determine the effect of a high intensity, intermittent exercise protocol on the star excursion balance tests for males and females

Chapter 4 (Study 2)

Aim: To investigate the effect of a high intensity exercise protocol on vertical drop jump performance in male athletes

Objectives:

To determine the effects of

1. a HIIP (pre-HIIP versus post-HIIP) on the pattern of trunk and lower limb biomechanics during the VDJ,

2. landing (first versus second landing) on the pattern of trunk and lower limb biomechanics during the VDJ, and

3. limb dominance on the pattern of trunk and lower limb biomechanics during the VDJ.

Chapter 5 (Study 3)

Aim: To investigate the effect of a high intensity exercise protocol on neurocognitive function in male athletes

Objective:

1. To determine the effect of a high intensity, intermittent exercise protocol on neurocognitive function as assessed by the stroop and symbol digit modalities test compared with a control group.

Chapter 6 (Study 4)

Aim: To investigate the effect of both anticipation and a high intensity, intermittent exercise protocol on trunk and lower limb biomechanics during the weight acceptance phase of crossover cutting.

Chapter 7 (Study 5)

Aim: To determine the effect of both anticipation and a high intensity, intermittent exercise protocol on trunk and lower limb biomechanics during the weight acceptance phase of side cutting.

Chapter 8 (Study 6)

Aim: To determine the effect of a dynamic core stability programme on trunk and lower limb biomechanics during the weight acceptance phases of anticipated and unanticipated side and crossover cutting.

Objectives:

1. To determine if a dynamic core stability programme will improve trunk kinematics associated with biomechanical risk factors for ACL injury compared with a control group

2. To determine if a dynamic core stability programme will ameliorate knee biomechanics associated with risk factors for ACL injury compared with a control group

3. To determine if a dynamic core stability programme will have a greater effect in the unanticipated compared with the anticipated condition.

Chapter 2 Review of Literature

2.1 Introduction to the Review of Literature

The overall aim of this thesis is to develop and evaluate an exercise programme aimed at addressing biomechanical risk factors for ACL injury during cutting manoeuvres. This follows the method proposed by Finch (Finch, 2006) to translate research into injury prevention practice. The exercise programme developed in this thesis will be based upon correcting detrimental biomechanics identified during vertical drop jumps post fatigue and cutting manoeuvres post fatigue and in unanticipated conditions. Consequently, the literature review will focus on biomechanical risk factors for ACL injury, and the effect of both anticipation and fatigue and on these risk factors during vertical drop jumps and cutting manoeuvres. It will be generally shown that fatigue and anticipation individually, and potentially in combination, can increase potential ACL stress and the risk for injury during cutting activities. However, our understanding is restricted by research primarily focussing on precise biomechanical values (e.g. peak knee abduction angle) rather than the pattern of biomechanical variables. Also there is limited research investigating the effects of fatigue, resulting from high intensity intermittent exercise, and unanticipation on trunk and pelvic kinematics. In addition, the majority of research on research understandably focuses on female athletes as females have a higher incidence of ACL injury. However, this limits our understanding of the risk factors in males and the development of ACL IPPS for male athletes.

The fatigue protocol in this thesis will be a novel high intensity, intermittent exercise protocol. Therefore, a secondary function of the thesis is to investigate the effect of this fatigue protocol on dynamic balance and aspects of neurocognitive function, which are important components of the performing unanticipated cutting manoeuvres. It will generally be shown that dynamic balance and neurocognitive function are impaired by a high intensity, intermittent exercise protocol. The literature review will also investigate the efficacy of ACL injury prevention programmes and in particular their effect on biomechanical risk factors for ACL injuries. It will be established that strengthening and hip and trunk control exercises increase the efficacy of ACL IPPs. However, the effect of

ACL IPPs on biomechanical risk factors for ACL injuries during cutting activities is not well understood. The review will also highlight the dearth of knowledge on methods to improve trunk kinematics during side, and particularly cross, cutting, which may limit the ability to refine current ACL IPPs.

2.2 Epidemiology

ACL injuries account for 2.5% of all collegiate injuries (Hootman et al., 2007) and almost 50% of knee injuries (Majewski et al., 2006) with a 0.15 incidence rate per 1000 athlete exposures. Almost half of the injuries occur during match play (Stanley et al., 2016) with a match to training ratio of over 20 (Walden et al., 2011) despite a greater time being spent in practice. The sports with the highest incidences are male American football, soccer and basketball with incidences of 0.33, 0.28 and 0.23 per 1000 athlete exposures respectively (Hootman et al., 2007). When high school injuries are included, females have a 0.15 incidence rate per 1000 athlete exposures compared with 0.06 for males (Stanley et al., 2016). Despite this, there is a greater overall number of ACL injuries in males (Gornitzky et al., 2016). For these reasons, much of the research investigating risk factors for ACL injuries and developing ACL IPPs has focussed on females in an attempt to correct the gender disparity. This approach has led to the development of successful ACL IPPs for females (Hewett and Bates, 2017). However, this approach limits our understanding of risk factors for ACL injury and the subsequent development of ACL injury prevention plans for males (Sugimoto, Alentorn-Geli et al., 2015).

ACL IPPs are not only important for preventing first time injuries, but also for secondary prevention. Regardless of the post injury intervention (Buller et al., 2014), athletes who have ACL reconstructive surgery have a subsequent ACL injury incidence rate 6 times higher than an uninjured athletes (Paterno et al., 2014) and a high rate of patellofemoral and tibiofemoral osteoarthritis 10 -15 years after reconstruction (Oiestad et al., 2009; Oiestad et al., 2010). Understandably, outcomes after a second ACL injury and subsequent reconstruction are considerably worse (Spindler et al., 2011). These findings highlight the importance of developing effective ACL IPPs. This requires a detailed understanding of the risk factors for ACL injuries.

2.3 Aetiology of Anterior Cruciate Ligament Injuries

According to Finch's proposed framework to prevent injuries (Finch, 2006), the next stage following the determination of the extent of the problem is to identify the aetiology and mechanisms of injury. Noncontact ACL injury risk is multifactorial in nature (Hewett et al., 2010) resulting from excessive loading relative to ACL integrity (Shultz et al., 2015).This section of the review will initially investigate the common mechanisms of noncontact ACL injuries, demonstrating that ACL injuries tend to be noncontact in nature and result from modifiable, sub-optimal biomechanics. It will then discuss the functional anatomy of the ACL in relation to these mechanisms of injury. It will be found that the ACL ligament undergoes greatest loading when the knee is subject to high compressive loads, anterior tibial translation, knee adductor and knee rotator moments[2]. The final element of this section of the review will focus on biomechanical risk factors for ACL injury by examining (1) the relationship between loading and injury, (2) technique and injury and (3) technique and loading. It will be demonstrated that specific knee loading, hamstring activation patterns, altered trunk control and reduced hip strength and range of motion are associated with ACL loading and injury. However, ACL injuries are multifactorial and it is also important to have an understanding of the potential role of anatomical and genetic risk factors for ACL injury.

2.3.1 Mechanism of Anterior Cruciate Ligament Injuries

ACL injuries occur when there is sufficient load to overcome the ACL tissue integrity (Shultz et al., 2015). This can occur when there is excessive load on a healthy ACL or

[2] An internal joint moment is a measure of the body's resistance (for example muscular contractions) to external moments generated by the ground reaction force and inertial forces. All joint moments are reported as "internal" moments in this thesis. When discussing findings from previous studies which originally reported results as "external" moments, the author has converted these results to "internal" moments to ensure consistency throughout.

normal loading on a weakened ACL (Lipps et al., 2013). Between 53% and 85% (Boden et al., 2000; Dick et al., 2007; Walden et al., 2015) of ACL injuries are noncontact in nature during activities that athletes generally complete hundreds of times during a sporting event without any injury (Lipps et al., 2013). This suggests that there is considerable potential for preventing noncontact ACL injuries provided the risk factors are identified and modified where possible. ACL injuries generally occur during deceleration activities such as landing from a jump or changing direction (cutting) (Cochrane et al., 2007; Montgomery et al., 2016; Walden et al., 2015). Cutting activities are more likely to cause injuries when they are performed in response to the sporting environment (Boden et al., 2009; Krosshaug et al., 2007; Walden et al., 2015). This suggests that athletes who sustained a noncontact ACL injury may have implemented a sub-optimal neuromuscular control programme, due to the unanticipated nature of the activities, resulting in excessive loading of the ACL. Video analyses of noncontact ACL injuries have identified similar patterns of kinematics at the time of ACL injuries. ACL injury generally occurs in the weight acceptance phase of deceleration activities, typically between 40 and 50 ms following initial contact (Koga et al., 2010; Krosshaug et al., 2007). Therefore, the athlete's posture during early stance is of particular importance. Firstly, athletes who sustain noncontact ACL injuries tend to land with a relatively extended lower limb with a small amount of hip and knee flexion (Koga et al., 2010; Olsen et al., 2004; Walden et al., 2015), increased knee internal rotation (Cochrane et al., 2007; Koga et al., 2010; Olsen et al., 2004) and abduction (Hewett et al., 2009) and with a heel strike (Montgomery et al., 2016) (Figure 2.1). Decreased trunk flexion and increased trunk lateral flexion have also been observed at the time of ACL injury (Hewett et al., 2009). This overall posture may limit the musculature's ability to dissipate high ground reaction forces (GRFs) leading to potentially hazardous loading of passive structures such as the ACL (Hewett et al., 2005) and ultimate macrotraumatic or fatigue failure (Lipps et al., 2013). Also, injuries generally occur towards the end of matchplay (Ekstrand et al., 2011; Hawkins et al., 2001) leading to the suggestion that fatigue is a risk factor for ACL injury (Borotikar et al., 2008; McLean and Samorezov, 2009; Shultz et al., 2015). However, the relationship between fatigue and the occurrence of ACL injuries during match play is not clear as Walden et al. (Walden et al., 2011) found that slightly more than half of ACL injuries in occurred in the first half with 40% occurring during the first 15 minutes of the first and second halves. This suggests that if fatigue is related to ACL injury risk, then temporary fatigue which occurs in both halves of a soccer match following intermittent bouts of high intensity exercise may be more important than sustained fatigue which occurs towards the end of a match (Knicker et al., 2011). However, there is a dearth of research investigating the effects of this type of fatigue.

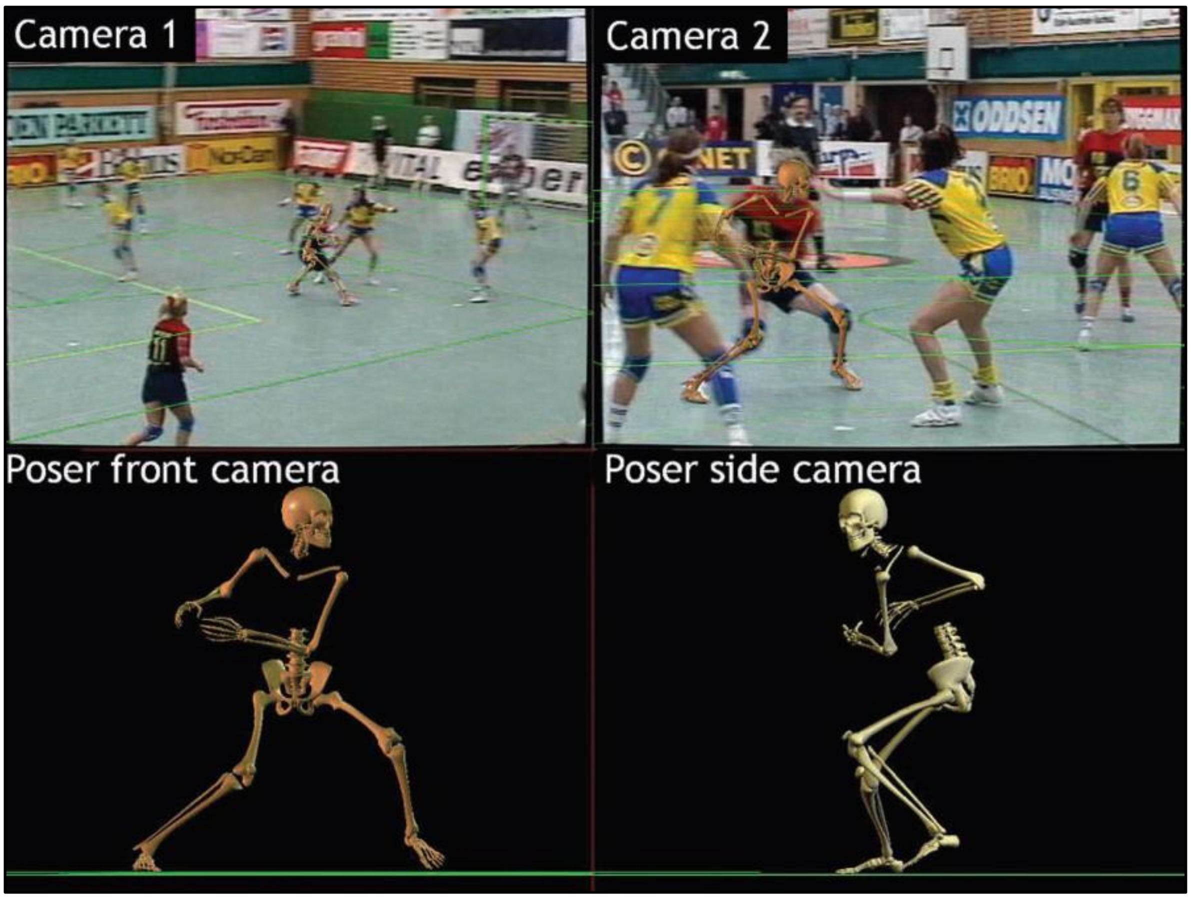

Figure 2.1 An example of an anterior cruciate ligament mechanism of injury from Koga et al., (2010)

In conclusion, noncontact ACL injuries occur during deceleration activities such as landing and cutting, particularly when athletes must react to the sporting environment and potentially when athletes experience temporary fatigue during sporting activities. As ACL injuries are noncontact in nature, it suggests that if their technique is modified, the relative excessive loading of the ACL may be avoided and therefore ACL injuries prevented. In order to achieve this, we must understand the interaction between the integrity and loading of the ACL (Shultz et al., 2015).

2.3.2 Functional Anatomy of the Anterior Cruciate Ligament

While the anatomy of the ACL is not being assessed as part of this thesis, it is essential to understand the function of the ACL in terms of biomechanical risk factors for ACL injury. It is also important to acknowledge the influence of anatomical and genetic risk factors on the tissue integrity of the ACL. The ACL passes anteriorly, medially and distally and rotates laterally from its attachment on the posterior aspect of the medial surface of the lateral femoral condyle to the tibial plateau, slightly anterior to the tibial spines (Petersen and Zantop, 2007). The ACL is thinnest in the mid portion (Harner et al., 1999) making it more susceptible to injury in this region in adults (McLean et al., 2015). Also, patients who sustain an ACL injury have smaller ACLs than controls (Chaudhari et al., 2009). Females have smaller ACLs compared with males when standardised for height and weight (Chandrashekar et al., 2005) and a lower modulus of elasticity. This results in less resistance to strain and rupture at lower stress levels (Chandrashekar et al., 2006) and may partially explain the observed gender bias in ACL injury incidence. In addition, patients with ACL injury have greater general laxity in their uninjured knee compared with the general population (Ramesh et al., 2005).

The tissue integrity of the ACL can also be negatively affected by differences in knee joint anatomy, namely the femoral intercondylar notch width inferior tibial plateau slope, and by genetic variations. Smaller intercondylar notch width has also been observed in patients who sustained ACL injuries compared with controls (Whitney et al., 2014). The smaller notch width is proposed to lead to impingement of the ACL, increasing shear force and contributing to macrotraumatic or fatigue failure (Simon et al., 2010). Also, it has been proposed that tibial morphology increases ACL stress. Greater lateral posterior inferior tibial plateau slopes are associated with a greater risk of non-contact ACL injury due to the resultant anterior tibial translation and internal tibial rotation (Beynnon et al., 2014) which increase ACL strain (Oh, Lipps et al., 2012; Shin et al., 2009). Although these anatomical factors have been identified as risk factors for ACL injury, they are not easy to modify and not addressed in ACL IPPs.

The integrity of the ACL can be affected by genetic variations which can lead to alterations in response to loading, (Posthumus et al., 2009). Variants have been observed in genes involved in the formation of collagen fibrils and cell signalling pathways involved in the remodelling process resulting in a structurally weaker ACL (Posthumus et al., 2009; Posthumus et al., 2012). Similarly, different sex hormones negatively affect the structural integrity of the ACL and partially explain the higher incidence of ACL injuries in females during the pre-ovulatory phase of the menstrual cycle (Beynnon et al., 2006). Hormone receptors on the ACL are proposed to regulate gene expression and collagen metabolism and reduce the mechanical strength of the ACL (Liu et al., 1996), placing the female athlete at risk of ACL injury at different times (Park et al., 2009). In conclusion, the structural integrity of the ACL may be affected by anatomical and genetic factors that may not be possible to modify. However, for ACL injuries to occur, an interplay must exist between the integrity of the ACL, which may not be modifiable, and the loading of the ACL, which can be modified (Shultz et al., 2015). It is therefore essential for us to understand the mechanical loading properties of the ACL.

The ACL is the primary restraint to anterior tibial translation, providing up to 86% of the total resistance (Butler et al., 1980), particularly at small knee flexion angles (Noyes and Grood, 1976). It stabilises the tibiofemoral joint throughout flexion and extension through the functional divisions, the anteromedial and posterolateral bundles. The posterolateral bundle provides stability in small flexion angles and the anteromedial bundle in greater flexion (Petersen and Zantop, 2007) (Figure 2.2). Given that quadriceps contraction at small knee flexion angles during functional activities is the primary cause of anterior tibial translation (DeMorat et al., 2004; Tsai et al., 2017) which can be resisted by hamstrings contraction, and noncontact ACL injuries typically occur during deceleration activities such as landing and cutting, it was felt that sagittal plane loading of the ACL was the primary cause of ACL rupture (McLean et al., 2015) (Figure 2.3).

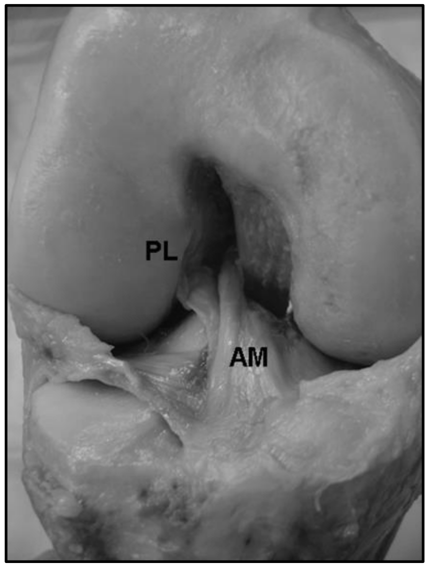

Figure 2.2.Anteromedial and posterolateral bundles of the anterior cruciate ligament (from Petersen et al., (2007)

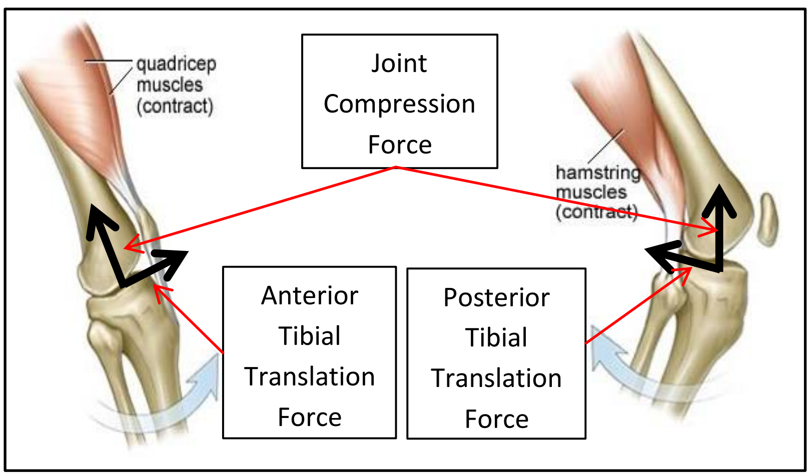

Figure 2.3 Tibial translation effect of quadriceps and hamstring contraction

Due to the distinct anatomical arrangement of the ACL, it also acts as a restraint to tibial adduction and abduction movements and rotation, in particular internal rotation (Oh et al., 2012; Shin et al., 2011). Cadaveric studies also demonstrate that the ACL becomes taut when an adductor moment is applied to the knee (McLean et al., 2015). It is the ACL, not the medial collateral ligament of the knee, which acts as the primary restraint to tibial adductor moments, particularly when it is combined with tibiofemoral compression and rotation (Quatman et al., 2014). Human-based computer models also suggest that sagittal plane loading alone cannot lead to ACL rupture (McLean et al., 2004). This may explain the tibial abduction is often observed at the time of ACL injury (Cochrane et al., 2007; Koga et al., 2010). In addition, video analysis of noncontact ACL injuries indicate that loading of the ACL occurs in all three planes at the time of injury (Hewett et al., 2009; Koga et al., 2010; Krosshaug et al., 2007). Cadaveric studies also demonstrate that it is the combination of tibiofemoral compression, rotation and valgus or varus moments that substantially increase ACL loading (Oh et al., 2012; Oh, Kreinbrink et al., 2012; Shin et al., 2007). The combination of a knee external tibial rotator and adductor moment results in a similar ACL strain compared with the coupling of an internal rotator and abductor moment (Oh et al., 2012). However, the steeper posterior slope of the lateral tibial plateau versus the medial tibial plateau will cause the lateral femoral condyle to slide posteriorly, increasing the knee external tibial rotator moment, which may increase the internal adductor moment further (Oh et al., 2012). Therefore, the combination of a knee external rotator moment to tibiofemoral compression and an extensor moment leads to greater ACL loading compared with an internal rotator moment (Oh et al., 2012; Oh et al., 2012).

In summary, ACL injuries occur as a result of the excessive loading relative to the tissue integrity. The ACL is placed under greatest stress when there is a combination of tibiofemoral joint compression, extensor, adductor and external rotator moments. This pattern of loading is similar to that suggested in video analyses of noncontact ACL injuries. These videos typically demonstrate the injury occurs during deceleration activities with a strong quadriceps contraction, a relatively extended knee (Boden et al., 2000; Krosshaug et al., 2007), internal tibial rotation (Cochrane et al., 2007; Koga et al., 2010; Olsen et al., 2004) and abduction (Hewett et al., 2009). However, tibiofemoral joint loading during jumping and landing activities is also influenced by other parts of the kinetic chain. It is therefore essential to understand the relationship between biomechanics of the kinetic chain and ACL loading.

2.4 Biomechanical and Neuromuscular Risk Factors for Anterior Cruciate Ligament Injuries

In order to develop ACL IPPs, it is essential that the relationship between biomechanical and neuromuscular factors and ACL injury is understood. The commonly used categorisation of biomechanical and neuromuscular risk factors for ACL injuries proposed by Hewett et al. (Hewett et al., 2010) is a useful system to understand the range of factors that may contribute to ACL injury. They theorised that patterns of movement of the kinetic chain can lead to (i) ligament dominant, (ii) quadriceps dominant, (iii) trunk dominant and (iv) leg dominant techniques which can increase ACL loading and the likelihood of ACL injury. The **ligament dominant theory** proposes that certain lower limb patterns of movement during high risk activities increase ACL stress and likelihood of injury. They also proposed the **quadriceps dominant theory** which suggests that excessive quadriceps forces relative to the hamstrings, results in increased anterior tibial translation and increased ACL strain and potential for injury. In addition, the **trunk dominant theory** proposes that a reduced ability to control the trunk during high risk activities increases the risk for ACL injury. Finally, the **leg dominant or asymmetrical theory** hypothesises that large differences in ACL loading between legs, during bilateral tasks such as landing or side to side differences during unilateral tasks such as cutting will predispose an athlete to ACL injury. While this is a useful categorisation of potential biomechanical and neuromuscular risk factors for ACL injuries, it does not clearly demonstrate the strength of the relationship between specific risk factors and ACL injuries. Therefore, this section of the review will attempt to demonstrate the strength of particular risk factors for ACL injuries in the following way. Initially, the relationship between loading and actual ACL injury will be examined. This section of the literature review will demonstrate that greater vertical GRF, knee valgus angle, knee adductor moment and decreased knee flexion angle during baseline vertical drop jumps are associated with subsequent noncontact ACL injury. Secondly, the factors that affect technique and their relationship between with ACL injury will be explored. It will generally be demonstrated that limited hip range of motion and strength and poorer trunk control predict ACL injury while reduced EMG activity of the hamstrings is associated with ACL injury. Finally, the relationship between technique and ACL loading will be reviewed. This section of the review will generally find that performing landing and cutting tasks with the knee in a more extended, abducted and internally rotated position, the hip in a more extended and internally rotated position and the trunk in less flexion, greater side flexion and rotation all act to increase ACL loading.

2.4.1The Relationship between Loading and Anterior Cruciate Ligament Injury

Four prospective studies have examined the relationship between loading during vertical drop jumps (VDJ) (Figure 2.4) and subsequent ACL injury with two studies finding biomechanical measures predictive of ACL injuries, whereas the other two studies did not find this. The vertical drop jump is a commonly used screening and risk factor identification tool (Bates et al., 2013b; Moran and Marshall, 2006). It consists of a drop from a height, a first landing followed by a maximal vertical jump and a second landing. It mimics a common mechanism of ACL injury (Bates et al., 2013b; Hewett et al., 2005) and has the benefit of high levels of reliability (Malfait et al., 2014). In a study of 205 females soccer and basketball players over 1 season using 3-d motion analysis, 9 of whom sustained a noncontact ACL injury, Hewett et al. (Hewett et al., 2005) found that knee adductor moment predicted ACL injury with 78% sensitivity and 73% specificity during the first landing of a vertical drop jump. A linear regression model analysing the most significant predictors of ACL injury in the study (knee abduction angles, internal knee adductor moments and side-side differences) demonstrated a predictive r squared value of 0.88. These findings support the categorisation of risk factors for ACL injuries proposed by Hewett et al. (Hewett et al., 2010), specifically the ligament dominant and limb asymmetry categorisations. Additionally, those who subsequently sustained an ACL injury had significantly greater knee abduction angles at initial contact (8.4°) and peak values (7.8°) ($p< 0.01$), peak vertical GRF (1266.1 ± 149.9 N versus 1057.8 ± 289.9; $p < 0.05$) than those who did not sustain an ACL injury. Greater peak knee adductor moments (45.3 ± 28.5 Nm versus 18.4 ± 15.6; $p <0.001$) were observed in athletes who subsequently sustained an ACL injury with 6.4 times greater side to side differences in knee adductor moments in the injured group compared with the uninjured group ($p < 0.001$). Although this study primarily demonstrated that higher levels of loading at the knee during vertical drop jumps predicted subsequent ACL injuries, it also established the relationship between hip loading, vertical GRF and knee joint loading in those who subsequently sustained ACL injuries. They found significant correlations between peak vertical GRF and knee abduction angle and hip adductor moment (R = 0.74 and R = 0.67 respectively, $p < 0.05$). Also, although there was no difference in hip adductor moment between the groups, it correlated to hip adductor moment in those who sustained ACL injuries (R = 0.69, $p < 0.05$). These findings demonstrate the relationship between vertical GRF, hip biomechanics and predictors of ACL injury.

Figure 2.4 Vertical drop jump
(from Paterno et al., (2010))

The results of the study by Hewett et al. (Hewett et al., 2005) are supported by a study examining biomechanical risk factors for ACL re-injury rates following ACL surgical reconstruction. In a study of 56 athletes (35 females and 21 males) who had a history of ACL reconstruction, 13 of whom subsequently reinjured the ACL, Paterno et al. (Paterno et al., 2010) found an ROC value of 0.94 for the combination of hip and knee rotator moments during landing, asymmetrical knee extensor moments at initial contact and deficits in dynamic stability. The hip and knee biomechanical variables alone predicted ACL re-injury with high sensitivity (0.92) and specificity (0.88). Athletes who subsequently reinjured their ACLs had greater knee valgus motion during landing (16.2° versus 12.1°) supporting the findings by Hewett et al. (Hewett et al., 2005). Athletes with greater frontal plane motion were over 3 times more likely to reinjure their ACL (OR = 3.5; 95% CI 1.3, 9.9) which is likely to be due to the increased loading of the ACL in this position (Markolf et al., 1995). As this movement pattern was identified during screening, it may be performed regularly during jumping and landing activities which may contribute to fatigue failure of the ACL graft.

Paterno et al. (Paterno et al., 2010) also found that differences in hip rotator moment impulse in the first 10% of landing in the uninvolved limb was significantly greater for those who subsequently sustained an ACL re-injury compared to those who did not (-2.4 x 10^{-3} Nms/kg versus 1.1 x 10^{-3}Nms/kg). Athletes with less external rotator moment impulse were over 8 times more likely to reinjure the ACL (OR = 8.4; 95% CI, 2.1, 33.3). Moment impulse during the first 10% of the landing represents the period of stance phase when ACL injuries are likely to occur. The fact that uninvolved hip kinetics predict ACL re-injury suggest the importance of bilateral hip control on knee loading during vertical drop jumps and the importance of assessing both limbs during vertical drop jumps. Also, side to side differences in the knee extensor moment at initial contact was identified as a predictor of ACL re-injury. Those who sustained ACL re-injuries had over 4 times more asymmetry in sagittal plane motions at initial contact and were over two times more likely to sustain an ACL re-injury (OR = 3.3; 95% CI = 1.2, 8.8). Athletes who sustained a second ACL injury had a smaller knee flexor moment at initial contact (-2.8 x 10^{-2} Nm/Kg) compared with the involved limb (-2.8 x 10^{-2} Nm/Kg) and with both the involved (-2.8 x 10^{-2} Nm/Kg) and uninvolved limbs (-2.8 x 10^{-2} Nm/Kg) in athletes who did not sustain a subsequent ACL re-injury. These findings suggest that suboptimal biomechanical patterns of the in both limbs during landing activities are associated with subsequent ACL re-injury and again demonstrates the importance of assessing both limbs during vertical drop jumps. To date, this has only been assessed in one cohort of athletes (Bates et al., 2013a; Bates et al., 2013b; Bates et al., 2013c).

In contrast a prospective study on have on 782 female handball and soccer players, 42 of which sustained noncontact ACL injuries, Krosshaug et al. (Krosshaug et al., 2016) did not find a predictive ability of any biomechanical measure examined [peak knee adductor moment (OR, 1.18; 95% CI 0.87 -1.60), knee valgus angle at initial contact (OR, 1.00; 95% CI 0.69 -1.45), peak vertical GRF (OR, 0.89; 95% CI 0.61 -1.29) and peak knee flexion angle (OR, 0.89; 95% CI 0.73 -1.36)]. While an association between medial knee displacement (resulting from knee and hip adduction) and injury (OR, 1.40; 95% CI 1.12-1.74, $p < 0.05$) was only observed in athletes who had a history of ACL injury, and it did not predict injury in those without a history of ACL injury (OR, 1.25; 95% CI 0.97 -1.61). This was still found to be the case when involvement in previous ACL IPPs was taken into consideration (Krosshaug et al., 2016). Furthermore, they did not find any difference in the injured groups versus the uninjured group in peak knee adductor moment (22.0 ± 12.0 versus 20.7 ±10.8 Nm; $p = 0.47$), knee valgus angle at initial contact (-2.2° ± 4.9 versus -1.8° ± 4.1; $p = 0.51$), peak vertical GRF (1311 ± 380 versus 1371 ± 429 N; $p + 0.36$) and peak knee flexion angle (91.6° ±14.2 versus 90.9° ± 14.8; $p = 0.75$). Overall, this comprehensive, prospective study with 42 ACL injuries did not support the findings of the study by Hewett et al. (Hewett et al., 2005) where there were just 9 ACL injuries. This has led to the conclusion that the vertical drop jump may not be an adequate test to identify biomechanical risk factors for noncontact ACL injuries (Krosshaug et al., 2016).

In a 3 year prospective study of 171 female basketball and floorball players, 15 of whom sustained a noncontact ACL injury, Lepannen et al. (Leppanen et al., 2017) investigated the relationship between knee valgus angle at initial contact, peak knee abduction moment, knee flexion angle at initial contact, peak knee flexion angle, peak vertical GRF, and medial knee displacement during baseline vertical drop jump screenings with subsequent noncontact ACL injuries. They found that the performing the vertical drop jump with greater sagittal plane loading was associated with noncontact ACL injuries. Specifically they found that peak knee flexion angle had a hazard ratio of 0.55 (95% CI, 0.34 – 0.88) for each 10° increase in peak knee flexion and higher peak vertical GRF had a hazard ratio of 1.26 (95% CI, 1.09 – 1.45) for each 100 N increase in peak vertical GRF. However, receiving operator characteristic analysis (which analyses the sensitivity and specificity of predictive factors) found a value of 0.6 for peak knee flexion angle and 0.7 for peak vertical GRF indicating poor to fair sensitivity and specificity. The authors of this study only found there to be a statistical difference in peak vertical GRF between the ACL injured and non-injured groups (1347 ± 403 N versus 1083 ± 321 N respectively; $p < 0.01$). Similar to the study by Krosshaug et al. (Krosshaug et al., 2016), they did not find any difference between the ACL injured group and the uninjured group in knee valgus angle at initial contact (0.9° ± 5.8 versus -1.8° ± 6.7; p – 0.12), knee flexion angle at initial contact (30.2° ± 11.7 versus 27.6° ± 9.0; $p = 0.29$), peak knee flexion angle (81.5° ± 10.0 versus 84.6° ± 10.3, p= 0.25), peak knee adductor moment (37.1 ± 24.9 Nm versus 31.2 ± 22.0; $p = 0.32$) and medial knee displacement (22.0 ± 18 mm versus 26 ± 20 mm; $p = 0.47$). In summary, the results of this study demonstrate that despite associations between biomechanical variables and injury, the vertical drop jump may not predict noncontact ACL injury with sufficient accuracy.

To conclude, biomechanical risk factors for ACL injury observed during assessment of baseline VDJs have limited ability to predict ACL injuries. However, there is some evidence to suggest that increased vertical GRF, knee valgus angle, knee adductor moment and decreased knee flexion angle observed during baseline vertical drop jumps are associated with noncontact ACL injury (Hewett et al., 2005; Leppanen et al., 2017) and ACL re-injury in particular (Paterno et al., 2010). Also, proximal biomechanical risk factors for ACL injuries have been identified at the hip joint (Hewett et al., 2005; Leppanen et al., 2017; Paterno et al., 2010). However, the effect of trunk biomechanics has not been investigated despite the proposal that they are significant risk factors for ACL injuries (Hewett et al., 2010). Also, as ACL injuries occur in the first 40-50 ms following initial contact, or weight acceptance phase, the peak kinetic and kinematic values analysed may not represent the vulnerable period for ACL injuries. Furthermore, despite video analysis demonstrating that the second landing is a common mechanism of non-contact ACL injuries in males (Cochrane et al., 2007; Walden et al., 2015), the prospective screening studies reviewed analyse the first landing of the VDJ only (Hewett et al., 2005; Krosshaug et al., 2016; Leppanen et al., 2017). Additionally, prospective studies analysing the relationship between ACL loading and injury have all been completed on females. Given the sex specific differences in the biomechanics of the vertical drop jump (McLean et al., 2007), our understanding of risk factors for ACL injuries in males is limited. Finally, as athletes demonstrate different knee biomechanics in high risk activities such as cutting (Kristianslund et al., 2014), investigations of activities such as cutting may increase our understanding of biomechanical risk factors for ACL injuries. Therefore, future studies should analyse the pattern of trunk, hip and knee biomechanical risk factors for ACL injuries in the first and second landing of the vertical drop in male participants and also during high risk activities such as cutting.

2.4.2 The Relationship Between Technique and ACL Injury

This section will explore the relationship between technique (and factors that may affect technique) and ACL injury. It will be generally found that trunk and hip biomechanical factors that affect technique and quadriceps and hamstring technique factors are associated with ACL injury whereas a landing technique scoring system is not predictive of ACL injury.

2.4.2.1 The relationship between trunk and hip biomechanics and ACL injury

One study demonstrated that poor trunk control predicts ACL injury while 5 out of 6 studies that investigated the relationship between hip range of motion and strength with ACL injury found a predictive or strong association. In relation to trunk control, a prospective study on 277 athletes, with 6 subsequent ACL injuries, found that a deficit in frontal plane control was the strongest predictor of ACL injury (OR = 2.32, $p = 0.02$) with deficits in control of trunk flexion (OR = 1.62, $p = 0.02$) also predicting ACL injuries in males and females combined (Zazulak et al., 2007). When divided by gender, frontal plane control was the only predictor of ACL in female athletes ($p = 0.024$) but not in male athletes. Deficits in trunk control were measured by positioning participants in a wooden apparatus that allowed them to produce isometric flexion, extension or lateral flexion contractions (Figure 2.5). The resistance to the isometric contraction was removed suddenly at random times during the isomeric contraction resulting in movement of the trunk, the degree of which was used for analysis. Despite the small number of ACL injuries in this study, the findings indicate that deficits in trunk control, particularly in the frontal plane, are predictive of ACL injury. Therefore, our understanding of the relationship between trunk control and ACL loading is important in developing ACL IPPs that effectively improve trunk control, reduce potential ACL loading and ultimately injury risk.

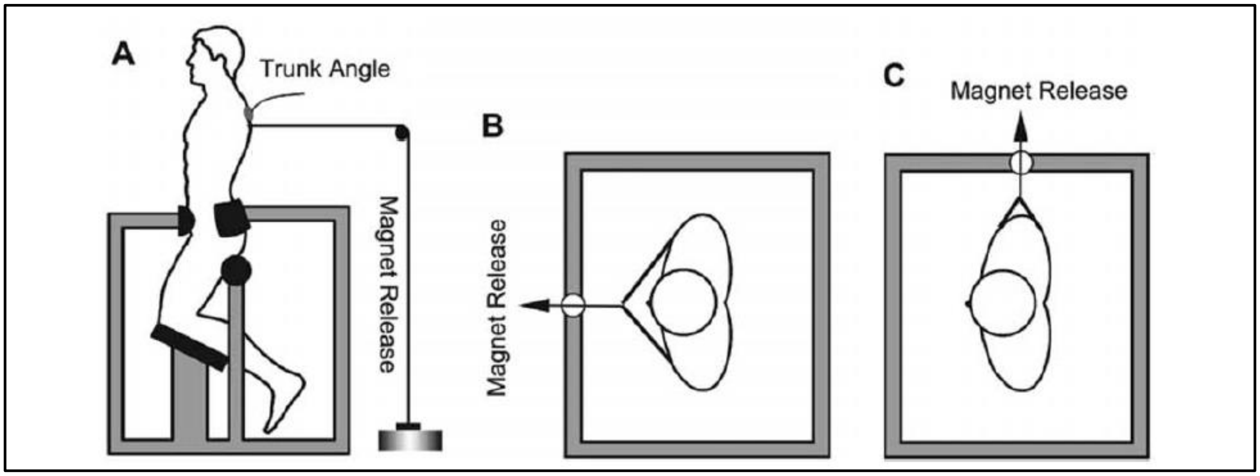

Figure 2.5. Method used to assess deficits in trunk control (from Zazulak et al., (2007))

Four retrospective studies have examined the relationship between range of hip rotation and ACL injury risk with all identifying an association between restricted hip range of rotation, in particular internal rotation, and noncontact ACL injury. Although these studies did not directly measure tibiofemoral biomechanics, reduced hip internal rotation has been found to result in a compensatory increase in internal rotation of the knee (Beaulieu et al., 2014), which itself increases ACL loading (Markolf et al., 1995; Oh et al., 2012; Shin et al., 2011) particularly during repetitive pivot loading (Beaulieu et al., 2015). In a retrospective study on 324 American footballers, 34 of whom had a unilateral history of ACL injury, Bedi et al. (Bedi et al., 2016) found that a restriction of left hip internal rotation was associated with a statistically significant increase in ACL injuries in the ipsilateral (OR = 0.95; 95% CI = 0.93, 0.98; $p = 0.001$) and contralateral knees (OR = 0.95; 95% CI = 0.93, 0.97; p <

0.001) while there was a non-significant trend towards an increase in ACL injuries with restricted right hip internal rotation and ipsilateral (OR = 0.95; 95% CI = 0.89, 1.01; $p > 0.05$) and contralateral (OR = 0.97; 95% CI = 0.92,1.02; $p > 0.05$) ACL injuries. They calculated that a 30° decrease in left hip internal rotation motion was associated with a 4.06 and 5.29 times greater odds of ipsilateral and contralateral ACL injuries whereas a similar restriction in right hip internal rotation motion was associated with a 5.19 and 2.71 times greater odds of ipsilateral and contralateral ACL injuries. However, this association may be overstated given that average hip internal rotation for all participants was 21.2° (95% CI = 20.1°, 22.4°) and 21.3° (95% CI = 20.1°, 22.4°). Nonetheless, this study demonstrates a clear link between the biomechanics of the hip and ACL injury, although a direct cause and effect relationship cannot be drawn from this.

In a retrospective, case control study on 50 soccer players with a noncontact ACL injury who were matched with 50 similar players without a history of ACL injury, Gomes et al. (Gomes et al., 2008) found an association between a reduction in hip rotation, in particular internal rotation, and ACL injuries. Specifically, it was found that athletes with a history of ACL injury had significantly smaller hip internal rotation (26.4° ± 7.7 versus 39.0° ± 7.1; $p < 0.001$), total (combined internal and external) right hip rotation (68.9° ± 13.8 versus 82.5° ± 7.4; $p < 0.001$), left hip rotation (68.0° ± 11.6 versus 82.1° ± 14.8; $p < 0.001$) and average total rotation (average total from left and right hips) (68.4° ± 12.3 versus 82.3° ± 13.8; $p < 0.001$) compared with the control group. It was found that by using an 70° or 80° cut off point, athletes with an ACL injury were more likely to have a decrease in hip range of motion compared with the control group (OR = 7.87; 95% CI = 3.07, 20.4 and OR = 11; 95% CI = 3.95, 30.3 respectively). Similarly, Ellera Gomes et al. (Ellera Gomes et al., 2014) found in a retrospective, case control study that athletes who suffered ACL re-injury had significantly lower total (combined internal and external) hip rotation when compared with health subjects (45.0° ± 7.9 versus 56.2° ± 10.3). Additionally, Vandenberg et al. (VandenBerg et al., 2017) found in a retrospective, case control study that those with a history of ACL injury had significantly lower internal (23.4 ° ± 7.6 versus 30.4° ± 10.4; $p = 0.009$) and total (combined internal and external) hip rotation (60.3° ± 12.4 versus 72.6°

± 17.2; $p = 0.006$) compared with the control group. They found that for every 10° increase in hip internal rotation there is a 0.419 factor decrease in the odds of having a history of ACL injury ($p = 0.015$) again demonstrating a relationship between reduced hip rotation and ACL injury.

The results of these studies show that reduced hip range of rotation, in particular internal rotation, is clearly associated with ACL injuries. Although a direct cause and effect relationship cannot be established given the retrospective nature of the studies, the association with injury may be due to a compensatory increase in internal rotation at the knee (Beaulieu et al., 2014). It is therefore essential that hip biomechanics are investigated as potential risk factors for ACL injuries so that ACL IPPs can also effectively correct suboptimal hip biomechanics.

It is proposed that a decrease in hip abductor and external rotator strength results in the decreased ability of athletes to prevent hip adduction and internal rotation during high risk activities such as landing and cutting, leading to increased knee valgus and subsequent ACL loading (Khayambashi et al., 2015; Steffen et al., 2016). In a prospective study by Khayambashi et al. (Khayambashi et al., 2016) over 1 season on 501 athletes (138 females and 363 males), 15 of whom subsequently suffered an ACL injury (6 females and 9 males), it was demonstrated that baseline isometric hip abductor (OR = 1.23, 95% CI = 1.08 – 1.39; $p = 0.001$) and external rotator strength (OR = 1.23, 95% CI = 1.08 – 1.39; $p = 0.001$) predicted future ACL injuries. They found that those who sustained ACL injuries had lower baseline hip external rotator (17.2 ± 2.9% body weight versus 22.1 ± 5.8% body weight; $p = 0.003$) and isometric abductor (30.8 ± 8.4% body weight versus 37.8 ± 7.6% body weight; $p < 0.001$) strength. They calculated that an external rotator strength of less than 20.3% body weight and abductor strength of less than 35.4% body weight predicted ACL injury with 93% and 87% sensitivity and 59% and 65% specificity, respectively. They also found that every percentage body weight decrease in isometric hip abductor and external rotator strength increased the odds of sustaining a noncontact ACL injury by 12% and 23%, respectively. In contrast to this, a prospective study over 8 years by Steffen et al. (Steffen et al., 2016) on 867 handball and soccer players, 57 of whom sustained a new ACL injury, did not find that reduced isometric hip abductor strength predicted ACL injury. Furthermore, they did not find any difference in isometric hip abductor strength in those who sustained ACL injuries compared to those who did not ($p = 0.10$). However, authors of this study suggested that the results of hip strength measurements should be interpreted with caution due to the low levels of reliability of the strength measurements observed in the study. They found ICC values of only 0.21 when examining the test-rest reliability of the isometric hip strength tests 1 to 5 years after baseline testing (average 2.2 years). Although Khayambashi et al. (Khayambashi et al., 2016) did not report test-rest reliability values, ACL injuries occurred within 1 season of the baseline measures, unlike the 8 seasons in the study by Steffen et al. (Steffen et al., 2016). Therefore, it suggests that deficiencies in hip strength abductor and external rotator strength are predictive of noncontact ACL injuries when assessed within 1 season. This may be due to a reduced ability to stabilise the femur, leading to a technique during landing and cutting activities that results in increased hip adduction and internal rotation (Claiborne et al., 2006; Lawrence et al., 2008) which increases ACL loading (Oh et al., 2012; Oh et al., 2012; Shin et al., 2007). Again, this reinforces the proposal that hip biomechanics should be investigated as potential risk factors for ACL injuries so that ACL IPPs can also target correction of any abnormal hip biomechanics.

2.4.2.2 The relationship between hamstring and quadriceps activation and ACL injury

A number of studies have examined the relationships between quadriceps and hamstring strength and activity and ACL injury. It is proposed that altered quadriceps and hamstring activity affects tibiofemoral stability during cutting and landing technique and can lead to increased ACL loading and subsequent injury risk (Hewett et al., 2010). It will be generally demonstrated that isokinetic strength of the hamstrings and quadriceps does not have an association with ACL injury whereas EMG activity of the hamstrings and quadriceps and ACL injury does.

Three prospective studies have examined the relationship between isokinetic strength and ACL injuries with only one demonstrating a weak link, while the others did not show any association. Firstly, a 4 year prospective study on 895 military recruits, 24 of who suffered an ACL injury, found that there was no association between isokinetic hamstring and quadriceps strength measurements and ratios (Uhorchak et al., 2003). Following on from this, a prospective, case control study by Myer et al. (Myer et al., 2009) examined isokinetic strength measurements between 22 athletes who sustained ACL injuries and 110 controls. They found that females who sustained a noncontact ACL injury had 15% (CI = 1-27%) lower hamstring strength compared to male, but not female, controls ($p = 0.04$), and no difference in quadriceps strength with any of the control groups. They suggested that a lower hamstring to quadriceps ratio may be a risk factor for ACL injury. In contrast to this, a larger prospective study by Steffen et al. (Steffen et al., 2016) over 8 years on 867 athletes, 57 of who sustained a new ACL injury, did not find any association between isokinetic hamstring and quadriceps strength ratios or absolute strength and ACL injury. They did not find any differences in isokinetic hamstring to quadriceps ratio (59.2 ± 8.6 versus 61.3 ± 9.3), hamstring strength (1.41 ± 0.25 Nm versus 1.43 ± 0.21 Nm) or quadriceps strength (2.41 ± 0.34 Nm versus 2.35 ± 0.33 Nm) between the groups who sustained ACL injuries and the uninjured group. The findings of these studies indicate that isokinetic hamstring and quadriceps strength and strength ratios are not related to subsequent ACL injuries.

On the other hand, one study has investigated the relationship between hamstrings and quadriceps EMG activity and noncontact ACL injury. A 2 year prospective study, Zebis et al. (Zebis et al., 2009) recorded the baseline hamstring and quadriceps EMG activity of 55 female soccer and handball players during side cutting. They found that the 5 athletes who subsequently sustained ACL injuries had significantly lower EMG pre-activity of the semitendinosus muscle (21% ± 6% versus 40% ± 17%; $p < .001$) and higher EMG pre-activity of the vastus lateralis muscle (69% ± 12% vs 35% ± 15%; $p < .01$) compared with the uninjured group. Pre-activity refers to the EMG activity 10 ms prior to foot contact. The decrease in hamstring pre-activity may lead to decreased tibiofemoral stability during the early phase of cutting, when ACL injuries tend to occur. Stability may be reduced further given that the overall difference between the vastus lateralis and semitendinosus EMG pre-activity was significantly greater ($p = 0.006$) in the injured (47% ± 14) compared with the uninjured group (2% ± 25). On the basis of these results they determined a 50% probability of sustaining an ACL injury if difference in vastus lateral and semitendinosus EMG pre-activity was greater than one standard deviation above the mean. The findings of this study demonstrate a strong association between the patterns of hamstring and quadriceps pre-activity and ACL injury.

In summary, differences in isokinetic strength measurements or strength ratios of the quadriceps and hamstrings are not associated with subsequent ACL injury however, the pre-activity EMG activity of the hamstrings and quadriceps are strongly associated with subsequent ACL injury.

2.4.2.3 The relationship between landing technique and ACL injury

The Landing Error Scoring System (LESS) has been developed as a clinician friendly, screening tool to identify errors in landing technique which may predispose athletes to ACL injuries. Two prospective studies have investigated this relationship and have not found it to be predictive of injury, although one of the studies did find a weak relationship between LESS scores and ACL injury. In a 3 year, prospective study on 5047 high school and collegiate athletes, 28 sustained noncontact ACL injuries (19 female, 9 male) and were matched with 64 controls (Smith et al., 2012). It was found that the LESS did not predict injury (OR = 1.04 per unit increase in LESS; 95%CI = 0.80 – 1.35; $p = 0.32$). This was still the case when analysing the entire group or subgroups based upon age and gender. There was no difference in LESS scores for those who sustained ACL injury (5.38 ± 1.85) and the matched controls (4.98 ± 2.00). Similarly, Padua et al. (Padua et al., 2015) did not find that LESS scores were predictive of ACL injury. However, in their study on 829 elite youth soccer players (481 females and 348 males), 7 of whom sustained ACL injuries (6 females, 1 in male), they found that those who sustained an ACL injury had a higher LESS score (6.24 ± 1.75; 95% CI = 4.62, 7.86) compared with those who did not sustain an ACL injury (4.43 ± 1.71; 95% CI = 4.34, 4.53) ($p < 0.005$). They also found that a LESS cut off score of 5 had optimal screening properties with 86% sensitivity (95% CI = 42% - 99%) and 64% specificity (95% CI = 62% - 67%). This suggests that the LESS has potential to identify adolescent athletes at risk of ACL injury. However, the cut-off point of 5 is very similar to the average values of those who did not sustain an injury in both studies. Therefore, a scoring system developed to assess landing technique has an equivocal relationship with ACL injury and requires further investigation.

Numata et al. (Numata et al., 2017) also examined the relationship between landing and ACL injury using 2 D video analysis to examine single leg drop landings.. In their prospective study, 27 athletes (out of 291 recruited) sustained a noncontact ACL injury over a three year period and were matched with 27 non injured athletes. They measured dynamic valgus as the distance from a line extending from the ASIS through the patella as far as the ground and the hallux (Figure 2.6). Although they did not report the predictive value of the dynamic valgus measurements, they found that dynamic valgus at initial contact and at maximum valgus were significantly greater in the injured compared with the non- injured group (2.1 ± 2.4cm versus 0.4 ± 2.2cm; $p = 0.006$ and 8.3 ± 4.3cm versus 5.1 ± 4.1cm; $p = 0.007$). This demonstrates the relationship between frontal plane loading of the knee and ACL injury which may be more apparent during single leg landing tasks compared with double leg landing. However, there are limitations to their study. Firstly they only used 2 D video analysis which has been shown to have poor to moderate correlation with 3 D motion analysis (Ortiz et al., 2016). Secondly, they only took dynamic valgus measurements at initial contact and maximum dynamic valgus despite the fact that ACL injuries tend to occur 40 to 50 milliseconds after initial contact (Koga et al., 2010; Krosshaug et al., 2007). Three D motion analysis throughout the weight acceptance phase of the landing may increase our understanding of the relationship between the biomechanics of single leg landing and ACL injury risk.

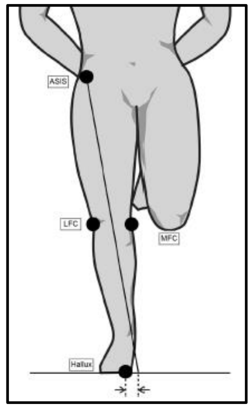

Figure 2.6 Method used to measure dynamic valgus (from Numata et al., (2017)

In conclusion, reduced trunk control and hip biomechanical factors that may affect technique are predictive of ACL injury. Also, the EMG activity of the hamstrings and quadriceps immediately prior to initial contact during a side cutting manoeuvre is strongly associated with ACL injury. Finally, there may be an association between ACL injury and scoring system for landing technique, in particular during single leg landing activities. The results demonstrate the importance of proximal biomechanics in particular on ACL injury risk. In order to develop ACL IPPs, these factors and the effectiveness of ACL IPPs in improving these factors should be investigated.

2.4.3 The Relationship Between Technique and ACL Loading

This section will examine the effect technique can have on ACL loading. Specifically, it will demonstrate how landing with an extended posture and how altered trunk kinematics can increase potential ACL loading. Given that these techniques are modifiable, ACL IPPs have the potential to improve these techniques, reduce potential ACL loading and injury risk.

2.4.3.1 The relationship between extended lower limb posture during landing and ACL loading

A number of studies demonstrate that landing from a drop jump with less hip and knee flexion increases potential ACL loading. Specifically, Tsai et al. (Tsai et al., 2017) investigated the effects of a 15 minute programme to train athletes in performing a drop landing with greater knee and hip flexion. Prior to the intervention, athletes landed with significantly less hip (93.1 ± 15.6° versus 99.6 ± 18.7°; $p = 0.02$) and knee (100.8 ± 11.7° versus 109.1 ± 12.4°; $p = 0.002$) flexion. They found that landing in an extended position significantly increased peak anterior tibiofemoral shear force (11.1 ± 3.3 versus 9.6 ± 2.7° Nkg^{-1}; $p = 0.008$), up to 86% which is resisted by the ACL (Butler et al., 1980). They proposed this was due to greater anterior shear effect resulting from the greater knee extensor moment (13.2 ± 3.5 vs. 11.2 ± 3.4 Nkg^{-1}, $p = 0.010$). They also found greater tibiofemoral compressive force (68.4 ± 7.6 vs. 62.0 ± 5.5 Nkg^{-1}, $p = 0.015$) principally due to a greater vertical GRF (14.0 ± 3.7 vs. 11.6 ± 3.6 Nkg^{-1}; $p = 0.024$) when performing the drop jump with an extended hip and knee posture compared with a flexed posture. The findings of this study support the previous study by Cowling et al. (Cowling et al., 2003) when it was found that when instructing 24 athletes to perform a run, leap and single leg landing task with greater knee flexion resulted in significantly lower peak vertical (3.41 ± 0.77 BW versus 3.10 ± 0.93 BW; $p < 0.05$) and posterior GRFs (1.70 ± 0.34 BW versus 1.52 ± 0.39 BW; $p < 0.05$) compared with landing in the more extended position. The findings from these studies demonstrate that landing with a relatively extended hip and knee increase ACL loading. Importantly, they also demonstrated that this technique could be improved leading to a decrease in potential loading of the ACL.

Onate et al. (Onate et al., 2005) examined the effect of different forms of video feedback on jump landing performance in a randomised controlled trial and found that a short education session resulted in immediate and prolonged increases in knee flexion angle and decreases in vertical GRF during jump landing tasks. They found significantly greater knee flexion (p = 0.044) immediately post feedback and at a one week follow up following self-feedback (90.02 ± 7.4° and 93.14 ± 7.4°) and a combination expert and self-feedback (98.07 ± 9.3° and 99.06 ± 11.2°) compared with the control group (72.87 ± 7.61° and 70.47 ± 8.99°). They observed smaller ($p = 0.021$) peak vertical GRF immediately post feedback and at one week retention in the self-feedback (4.56 ± 1.13BW versus 3.33± 0.33 and 2.94 ± 0.51 respectively) and combination groups (4.79 ± 0.99 BW versus 3.43 ± 0.78 and 3.31 ± 0.64 respectively) compared with the control group (4.61 ± 0.76 BW versus 3.08 ± 0.59 and 3.18 ± 0.69 respectively). These results demonstrate that landing with less knee flexion increases peak vertical GRF and subsequent potential ACL loading. However, unlike the study by Tsai et al. (Tsai et al., 2017), Onate et al. (Onate et al., 2005) did not find any difference in anterior tibiofemoral shear force. Tsai et al. (Tsai et al., 2017) proposed that this additional information from their study was due to the use of a magnetic resonance imaging and EMG based modelling that also took into account individual muscle forces, unlike the inverse dynamics model used by Onate et al. (Onate et al., 2005). Similarly Irmischer et al. (Irmischer et al., 2004) found that a 9 week training programme on 28 athletes (14 in the intervention group and 14 controls) encouraging soft landing decreased peak vertical GRF (5.8 ± 0.16 BW versus 3.9 ± 0.6 BW; $p = 0.004$) and rate of force development (0.12 ± 0.06 $BWms^{-1}$ versus 0.08 ± 0.02 $BWms^{-1}$; $p = 0.02$) during drop landings. However, they did not report knee and hip flexion angles thus making it difficult to determine if the improvement was due to a more flexed posture in landing.

In a progression to these studies, Dai et al. (Dai et al., 2015) examined the effects of performing a stop jump and side cutting task with greater knee and hip flexion angles compared with a relatively extended limb. The stop jump and side cut tasks may be more ecologically valid as they involve a running and jump or change of direction activity which may more accurately replicate the ACL injury mechanism. Dai et al. (Dai et al., 2015) found that performing a stop jump with a relatively extended position of the hip and knee, compared to a flexed position, resulted in greater peak posterior GRF (-0.64 ± 0.26 BW and -0.65± 0.26 BW versus -0.52 ± 0.22 BW and -0.57 ± 0.21BW for males and females respectively; $p = 0.007$) and a concurrent greater knee extensor moment (-0.06 ± 0.04 BW.H and -0.07± 0.03 BW.H versus -0.06 ± 0..03 BW.H and -0.05 ± 0.02 BW.H for males and females respectively; $p = 0.023$). Peak posterior GRF was also greater in the extended posture compared to the flexed posture during the side cutting manoeuvre (-0.73 ± 0.28 BW and -0.67 ± 0.30 BW versus -0.60 ± 0.25 BW and -0.51 ± 0.20 BW for males and females respectively; $p < 0.001$) with a concurrent greater knee extensor moment (-0.07 ± 0.06 BW.H and -0.05 ± 0.04 BW.H versus -0.07 ± 0.05 BW.H and -0.04 ± 0.03 BW.H for males and females respectively; $p = 0.04$). As greater posterior GRF and knee extensor moments have been shown to increase ACL loading (Sell et al., 2007; Tsai et al., 2017), performing stop jump and side cutting manoeuvres with less knee flexion increases potential ACL loading during the stop jump and side cut.

In summary, performing jump landing and cutting tasks with a more extended hip and knee results in greater vertical and posterior GRFs and anterior tibiofemoral shear force which increases the loading of the ACL. This may be a contributory risk factor for ACL injuries as the extended lower limb posture is commonly observed during ACL injury plane (Hewett et al., 2009; Koga et al., 2010; Olsen et al., 2004; Walden et al., 2015). Importantly, this posture can be modified with short interventions and this modification can be maintained with a corresponding decrease in potential ACL loading. However, it must be noted that it might also result in a decrease in performance of athletic tasks. This presents the challenge to sports medicine clinicians of developing ACL IPPs which decrease ACL injury risk and have minimally negative effects on performance.

2.4.3.2 The relationship between trunk kinematics and ACL loading

A number of biomechanical studies have demonstrated the relationship between trunk position and knee joint loading during dynamic activities and are summarised in Table 2.1. It will generally be found that landing with a relatively extended trunk and side flexed trunk will increase potential ACL loading as will decreased trunk rotation in the direction of cutting.

Landing with a more extended trunk position during drop landings is likely to increase ACL strain due to the increased peak (Blackburn and Padua, 2009; Shimokochi et al., 2013) and decreased time to peak (Shimokochi et al., 2016) vertical GRF. This is because it leads to increased knee extensor moment (Shimokochi et al., 2013) and quadriceps activity (Blackburn and Padua, 2009) and subsequent anterior tibial shear force (Shimokochi et al., 2016). These findings may be, at least partially, explained by a relatively extended hip and knee posture (Blackburn and Padua, 2008; Shimokochi et al., 2013; Shimokochi et al., 2016) when performing drop landings with a more extended trunk posture. Although increasing trunk flexion (decreasing trunk extension) has been largely demonstrated to reduce potential sagittal plane ACL loading during drop landings (Blackburn and Padua, 2008; Blackburn and Padua, 2009; Shimokochi et al., 2013; Shimokochi et al., 2016), the effect during cutting activities is less clear. Frank et al. (Frank et al., 2013) found that increasing trunk flexion during side cutting was associated with increased knee external rotator moment ($r = 0.42$, $P = 0.020$). This may be due to the requirement to decelerate the momentum of the centre of mass in the sagittal plane and to re-direct it in the transverse plane during side cutting, which is not present in drop jumps or drop landings. It also suggests that the relationship between trunk kinematics and ACL injury risk factors are task dependent.

Altered trunk kinematics in the frontal and transverse planes may also affect biomechanical risk factors for ACL injury during side cutting activities (Table 2.1) although it has not been investigated during crossover cutting. In a study comparing different anticipated side cutting techniques, Dempsey et al., (2007) demonstrated that a performing the side cut with additional trunk side flexion away from the direction of cut resulted in greater peak knee adductor moment during the weight acceptance phase. Jamison et al. (Jamison et al., 2012) ($p = 0.002$) and Mornieux et al. (Mornieux et al., 2014) also found an association between increased trunk side flexion and greater knee adductor moment during unanticipated side cutting ($r = 0.41$, $P = 0.009$). Additionally, transverse plane trunk kinematics have also been demonstrated to affect knee biomechanics as less trunk rotation in the direction of travel during anticipated side cutting is associated with increased (internal) knee external rotator moments (Dempsey et al., 2007; Frank et al., 2013; Jamison, Pan et al., 2012).

In summary, a relatively extended trunk position during landings, a side flexed trunk posture away from the direction of cutting and decreased trunk rotation in the direction of cutting during side cutting results in increased potential ACL loading. However, this relationship has not been investigated in crossover cutting. Future research should investigate the relationship between trunk position and potential ACL loading. Additionally, the relationship between trunk kinematics and potential ACL loading should be investigated under fatigued and unanticipated conditions which are proposed to increase the likelihood of ACL injuries (Borotikar et al., 2008; McLean and Samorezov, 2009).

Table 2.1 The association between trunk kinematics and knee joint biomechanics

Study and participants	Activity	Phase analysed	Significant Findings	Magnitude of differences	Statistical value
Blackburn et al., (2009) 40 healthy, physically active participants (20 female, 20 male)	Drop landings 1. Less trunk flexion 2. Greater trunk flexion	1. Initial ground contact 2. Peak value for respective variable	Less trunk flexion resulted in 1. Increased peak vertical GRF (BW) 2. Increased quadriceps EMG activity (% MVIC) There was no trunk flexion by sex interaction or main effect for anterior tibiofemoral shear force	 3.98 ± 0.82 *vs* 3.42 ± 0.81 216 ± 197 *vs* 156 ± 146	 *P* <0.001 *P* <0.001
Shimokochi et al., (2013) 20 recreationally active participants (10 female, 10 male)	Single leg drop landings 1. Less trunk flexion 2. Greater trunk flexion	1. Peak, and time to peak kinematic and kinetic values 2. EMG values recorded in first 100ms following initial contact	Less trunk flexion resulted in 1. Greater peak vertical ground reaction force 2. Greater peak knee extensor moment 3. Greater dorsiflexor moment 4. Smaller hip extensor moments 5. Smaller knee flexion angle at peak knee extensor moment (°) 6. Less lateral gastrocnemius activity (%MVIC) 7. Less medial gastrocnemius activity (%MVIC) 8. Greater lateral quadriceps muscle activations activity (%MVIC)	 5.3 ± 0.8 *vs* 3.9 ± 0.7 0.14 ± 0.04 *vs* 0.09 ± 0.03 0.03 ± 0.07 *vs* -0.05 ± 0.03 -0.04 ± 0.12 *vs* 0.05 ± 0.08 26.7 ± 10.5 *vs* 37.5 ± 9.5 49.7 ± 20.1 *vs* 59.7 ± 22.3 49.8 ± 24.3 *vs* 62.7 ± 23.6 59.2 ± 26.1 *vs* 53.0 ± 19.7	 *P* <0.001 *P* <0.001 *P* <0.001 *P* <0.050 *P* <0.001 *P* <0.001 *P* <0.001 *P* <0.050
Blackburn et al., (2008) 40 healthy, physically active participants (20 female, 20 male)	Double leg drop landing 1. Normal landing 2. Flexed trunk	1. Initial ground contact 2. Peak value for respective variable in loading phase (i.e. to peak knee	Less trunk flexion resulted in 1. Decreased hip flexion angle (°) a. At initial contact b. Peak value 2. Decreased knee flexion angle (°) a. At initial contact	 14 ± 12 *vs* 20 ± 12 40 ± 20 *vs* 71 ± 19 6 ± 7 *vs* 9 ± 11	 *P* <0.001 *P* <0.001 *P* <0.001

		flexion)	b. Peak value No difference in hip and knee rotation or abduction/adduction angles	69 ± 16 *vs* 91± 16	*P* <0.001
Shimokochi et al., (2016) 20 recreationally active participants (10 female, 10 male)	Single leg drop landings 1. Less trunk flexion 2. Greater trunk flexion	Peak and time to peak values at time of peak tibial axial forces	Less trunk flexion resulted in 1. Shorter time to peak vertical tibial force (ms) 2. Greater peak tibial shear force (BW) 3. Greater peak proximal anterior tibiofemoral shear force (BW) 4. Greater vertical GRF (BW) 5. Smaller knee flexion angles at peak tibial vertical force (°)	37 ± 14 *vs* 67 ± 13 4.9 ± 0.7 *vs* 3.6 ± 1.3 0.5 ± 0.5 *vs* 0.3 ± 0.5 3.8 ± 0.6 *vs* 5.1 ± 0.8 18.0 ± 6.6 *vs* 28.7 ± 8.1	*P* <0.010 *P* <0.010 *P* <0.010 *P* <0.010 *P* <0.010
Dempsey et al., (2007) 15 male amateur footballers	4.5 ms^{-1} run and 45° cut manoeuvre with 10 different techniques 1. Trunk lean to and away from cutting direction 2. Trunk rotation away from direction of cut 3. Normal technique	Peak values during weight acceptance phase of side cut	1. Trunk lean away from direction of cut led to greater internal knee varus moment compare with normal technique ($Nmkg^{-1}m^{-1}$) 2. 2. Trunk rotation away from direction of cut led to greater internal knee external rotator moment compared with normal technique ($Nmkg^{-1}m^{-1}$)	0.65 ± 0.36 *vs* 0.45 ± 0.32 0.29 ± 0.1 *vs* 0.19 ± 0.1	*P* <0.050 *P* = 0.001
Jamison et al., (2012) 29 healthy participants (14 female, 15 male)	Unanticipated run and 45° cut manoeuvre at self-selected pace	Peak values during weight acceptance phase of side cut	1. Trunk side flexion away from direction of cut positively associated with peak internal knee adductor moment 2. Trunk rotation away from direction of cut negatively associated with peak internal knee external rotator moment	*Not applicable*	*P* = 0.002 *P* = 0.021

Mornieux et al. (2014) 13 male amateur soccer players	Run and 45° cut manoeuvre under following conditions 1. anticipated 2. instructed 850 ms before cut 3. instructed 600 ms before cut 4. instructed 850 ms before cut	Values at time of peak internal knee adductor moment	1. Lateral trunk flexion positively correlated with increased internal knee adductor moment 2. Greater trunk side flexion away from direction of cut in 500ms and 600 ms conditions compared with a. anticipated condition b. 850 ms condition 3. Greater trunk side rotation away from direction of cut in 500ms and 600 ms conditions compared with a. anticipated condition b. 850 ms condition	*not applicable* 12.2 ± 5.7° and 9.8 ± 5.4° *vs* 1.9 ± 5.2° 3.6 ± 4.7° 11.1 ± 7.3 and 10.8 ± 9.3 *vs* 7.1 ± 8.4 7.0 ± 8.5	r = 0.41 $P = 0.009$ $P < 0.001$ $P < 0.001$ $P < 0.026$ $P < 0.026$
Frank, Bell et al. (2013) 30 participants (15 females, 15 males)	Two footed standing jump 50% of body height over a 17cm hurdle and then performed a side cut at 60° angle	Peak values during first 50% of side cutting task	Internal knee adductor moment 1. negatively associated with trunk rotation in direction of cut 2. positively associated with hip adductor moment 3. 81% of the variance of internal knee adductor moment can be explained by the combination of trunk rotation and hip adductor moment Internal knee external rotator moment 1. positively associated with trunk flexion 2. positively associated with internal hip internal rotator moment 3. 48% of the variance of internal knee external rotator moment can be explained by the combination of trunk flexion and hip rotator moment	*Not applicable*	$r = -0.46$, $P = 0.011$ $r = 0.83$, $P < 0.050$ $R^2 = 0.81$ $P < 0.05$ $r = 0.42$, $P = 0.020$ $r = 0.59$, $P = 0.001$ $R^2 = 0.48$ $P < 0.05$
Demspey et al.,	Investigated the	Peak internal knee	1. Significantly less trunk side flexion (°)		

(2009) 9 male, non-elite team sport players	effects of a 6 week side cutting technique improvement programme on side cutting technique in planned and unplanned conditions	adductor and external rotator moments, and mean knee flexor and extensor moments during weight acceptance phase.	away from direction of cut following programme for a. planned side cuts b. unplanned side cuts 2. Significantly smaller peak internal knee adductor moments ($Nmkg^{-1}$) following programme for a. planned side cuts b. unplanned side cuts	 0.24 ± 0.22 *vs* 0.38 ± 0.26 0.24 ± 0.22 *vs* 0.38 ± 0.26 0.24 ± 0.22 *vs* 0.38 ± 0.26 0.26 ± 0.11 *vs* 0.40 ± 0.23	 $P = 0.005$ $P = 0.034$
Imwalle et al., (2009) 19 female soccer players	Standing jump (0.4 m) , unanticipated side cut and run task	Kinematics at peak vertical GRF	1. Hip adduction angle was a significant predictor of knee abduction angle 2. Hip internal rotation angle not a predictor of knee abduction angle. 3. Did not investigate the effect on knee internal rotation angle	*Not applicable*	$R^2 = 0.49$

GRF = Ground reaction force, EMG = Electromyography, MVIC = Maximal voluntary isometric contraction

2.5 The Relationship between Fatigue and Biomechanical and Neuromuscular Risk Factors for Anterior Cruciate Ligament Injuries

Fatigue is proposed to increase the risk of ACL injury by resulting in the adoption of altered neuromuscular patterns (Kernozek et al., 2008; McLean et al., 2007; Zebis et al., 2010) which potentially increase ACL loading (McLean et al., 2008). Fatigue can be defined as an “exercise induced decline in performance” due to a complex interaction of multiple processes (Knicker et al., 2011). A decline in performance occurs both towards the end of a sporting event and temporarily following periods of high intensity exercise during a sporting event (Knicker et al., 2011; Mohr et al., 2005). Fatigue can generally be described as a deterioration of exercise or competition performance (Knicker et al., 2011) which may be contributed to by a decrease in muscular force output (Gandevia 2001). The physiological processes involved in fatigue are generally divided into central and peripheral fatigue. Central fatigue refers to an exercise induced reduction in voluntary muscle activation due to altered processes proximal to the neuromuscular junction, whereas peripheral fatigue refers to alterations distal to the neuromuscular junction (Gandevia, 2001; J. L. Taylor and Gandevia, 2008).

2.5.1 Physiology of Neuromuscular Fatigue

There are several central processes at supraspinal and spinal level that lead to reduced neuromuscular activation. An increased perception of exertion is influenced by afferent information from the exercising muscle, cardiovascular and respiratory systems and the biochemical and temperature by-products of exercising muscles. This increased perception of exertion coupled with inter-related altered levels of motivation and cortical output can reduce the motor cortical output (Knicker et al., 2011). These central factors can lead to a 25% reduction in muscle force output (Gandevia, 2001; J. L. Taylor and Gandevia, 2008). During fatiguing exercise, there is a general reduction in the excitatory impulse to the lower motor neuron at spinal level, an increased inhibitory affect at supraspinal level and an input into the perception of exertion from the muscle afferents (Gandevia, 2001; J. L. Taylor and

Gandevia, 2008). This negatively affects cortical motor drive or result in inhibition of the motor pathway at the spinal level, ultimately leading to a decrease in muscle force output. The decreased muscle force output may be contributed to by "late adaptation" (Gandevia, 2001; J. L. Taylor and Gandevia, 2008). Late adaptation is a decrease in the motor neurones' responsiveness rate due to a decrease in responsiveness to synaptic input following prolonged stimulation. It may contribute to temporary fatigue observed following bouts of high intensity exercises during sport and generally recovers after 1-2 minutes rest (Gandevia, 2001; J. L. Taylor and Gandevia, 2008).

It has been proposed that these central adaptations to exercise are a protective mechanism, preventing exercising beyond the capacity of the neuromuscular, cardiovascular and respiratory systems (Gandevia, 2001) and may allow a reserve for maximal tasks (Gandevia, 2001). Therefore, if exercise is submaximal then fatigue may occur without a deterioration of performance as the body can instigate compensatory mechanisms such as recruitment of other motor units or muscles (Knicker et al., 2011; J. L. Taylor and Gandevia, 2008). Furthermore, central fatigue can be mediated during maximal or supra maximal efforts, suggesting that the manifestations of fatigue in maximal and submaximal activities may be influenced by levels of motivation and neurocognitive function.

Peripheral fatigue results from alterations in the contractile or transmission mechanisms (Boyas and Guevel, 2011). Neuromuscular transmission is the transformation of a nerve action potential into a muscle action potential. Sustained motor nerve stimulation leads to a decrease of presynaptic neurotransmitter release and the amplitude of motor end plate potentials and subsequent reduction in muscle force output (Reid et al., 1999; Wu and Betz, 1998). This may be due to a decrease in the amount of neurotransmitter released, or a decrease in the number of exocytotic vesicles (Reid et al., 1999; Wu and Betz, 1998), or an increase in the length of time required for acetylcholine to bind to post synaptic receptors (Magleby and Pallotta, 1981). During repeated muscular depolarisations, the sodium potassium pumps do not adequately restore the correct potassium and inhibition of concentration leading to a decrease in propagated action potentials and subsequent muscular force output (Lannergren and Westerblad, 1987).

Peripheral fatigue may also be contributed to by metabolic changes within the muscle, which negatively affect muscular contractile mechanisms in particular the increases in intramuscular hydrogen ion and inorganic phosphate concentrations (Boyas and Guevel, 2011). The increased inorganic phosphate concentration, due hydrolysis of ATP to ADP and inorganic phosphate that occur in the phosphagen energy supply system, impairs cross bridge formation in early exercise, and reduce myofibril sensitivity to depolarisation in later exercise (Allen et al., 2008). The increase in hydrogen ions during anaerobic glycolysis leads to a drop in cellular acidosis which negatively affects the cross bridge in the myofibril and membrane conductance. It also inhibits the release of calcium and subsequent depolarisation of the sacrolemmas (Allen et al., 2008; Boyas and Guevel, 2011; Noakes, 2000). Finally, reactive oxygen species, which are produced in the exercising muscle, are proposed to negatively affect the contractile proteins, the sodium potassium pump and stimulate inhibitory muscle afferents (Allen et al., 2008; Knicker et al., 2011). The combined effects of these peripheral processes lead to a decrease in muscle force output and muscular fatigue.

In conclusion, there are several central and peripheral processes during exercise that can lead to fatigue. These processes are influenced by motivation and neurocognitive function. No single process is strongly associated with the decline in performance during activities requiring rhythmic, moderate-high intensity exercise (Gandevia, 2001). Furthermore, given that fatigue is task dependent, i.e. the exercise that is performed dictates the mechanisms that cause fatigue (Enoka and Duchateau, 2008), the ecological validity of exercise protocols is critical to our understanding of the effects of fatigue on biomechanical and neuromuscular risk factors for ACL injury.

2.5.2 The Effects of Fatiguing Protocols on Biomechanical and Neuromuscular Risk Factors for Anterior Cruciate Ligament Injuries

This section examines the effects of differing fatiguing protocols on the biomechanics of jumping and landing activities such as drop jumps, stop jumps and cutting. It will generally be found that a variety of fatigue protocols result in biomechanical changes along the kinetic chain which increase potential ACL loading during ecologically valid tasks. For example, activities that require a horizontal deceleration component, such as cutting or stop jumping, result in the adoption of a more joint-extended landing posture which can increase ACL loading. On the other hand, activities that predominantly require vertical deceleration, such as drop jumps and drop landings, lead to a more joint-flexed landing posture post fatigue which will generally result in decreased loading of the ACL. Therefore, this section will demonstrate that the effects of fatigue depend upon the task undertaken (Barber-Westin and Noyes, 2017).

Fatigue results in smaller hip flexion (Borotikar et al., 2008; Cortes et al., 2012; Cortes et al., 2013; J. H. Kim et al., 2014; Lucci et al., 2011; McLean and Samorezov, 2009; Potter et al., 2014; Quammen et al., 2012) and knee flexion (Borotikar et al., 2008; Chappell et al., 2005; Cortes et al., 2013; Khalid et al., 2015; J. H. Kim et al., 2014; Lucci et al., 2011; McLean and Samorezov, 2009; O'Connor et al., 2015; Potter et al., 2014; Quammen et al., 2012; Raja Azidin et al., 2015) angles during cutting manoeuvres (Table 2.2). A more extended hip and knee posture during landing results in decreased hip extensor moments and greater knee extensor moments (Shimokochi et al., 2016) and quadriceps activity (Blackburn and Padua, 2009). This suggests that a neuromuscular strategy is adopted post fatigue which increases knee joint loading and decreases hip joint loading. Although a decrease in hip extensor moment and an increase in knee extensor moments may be expected, the results are not consistent. The studies of Khalid et al (Khalid et al., 2015) and Kim et al. (Kim et al., 2014) found an increased knee extensor moment post fatigue in contrast with other studies which found a decrease in extensor moment (Lucci et al., 2011; McLean and Samorezov, 2009; O'Connor et al., 2015). These contradictory findings may be explained by the timing of the measurements. In contrast to the pre-selection of the timing of knee extensor moment measurements (e.g. initial contact, peak vertical GRF), Kim et al. (Kim et al., 2015) reported the kinetic changes during phases of the cutting manoeuvre. They found that there was an initial increase in knee extensor moment followed by a decrease in knee extensor moment. As ACL injuries typically occur during early stance, the increased knee extensor moment and decreased hip extensor moment may be of greater practical importance. Interestingly, they found that the increase in knee extensor moment corresponded with a decrease in hip extensor moment during the early stance phase of cutting. A subsequent decrease in knee extensor moment also corresponded with an increase in hip extensor moments. This inverse relationship in hip and knee moment is in line with the findings of Shimokochi et al. (Shimokochi et al., 2016) and was reflected in the EMG activity of the hip and knee extensor muscles (Kim et al., 2015). The findings of Kim et al. (Kim et al., 2015) are important for a number of reasons. Firstly, they can explain the contradictory findings on the effect of fatigue on knee extensor moments. Secondly they demonstrate the inverse relationship between hip and knee extensor moments. Critically, they also demonstrate the importance of examining the phases of kinetic and kinematics rather than the selection of predetermined, discrete points. The findings of Kim et al. (Kim et al., 2015) also show that fatigue does not only affect the net joint moments but also the timing of the activation of the gluteus maximus and quadriceps muscles which may influence the dynamic stability of the joints of the lower limb.

Table 2.2 The effect of fatigue on neuromuscular and biomechanical risk factors for anterior cruciate ligament injuries during cutting tasks

Study and participants	Activity	Fatiguing methodology	Effect of Fatigue	Magnitude of differences	Statistical value
Cortes et al., (2012) 15 female collegiate soccer players	Unanticipated stop jump (2-legged) Unanticipated side cutting manoeuvres	Running, step-ups, counter movement jumps and agility drill Protocol ceased after 4 repetitions	**Kinematic findings**		
			1. Greater knee internal rotation angle at initial contact (°)	11.4 ± 7.5 *vs* 7.9 ± 6.5	*P* = 0.011
			2. Less knee flexion at peak stance (°)	36.6 ± 6.2 *vs* 40.0 ± 6.3	*P* = 0.003
			3. Less hip flexion at initial contact (°)	35.5 ± 8.7 *vs* 43.2 ± 9.5	*P* = 0.002
			Kinetic findings	*Not assessed*	
Cortes et al., (2013) 18 female, collegiate soccer players	Unanticipated side cutting manoeuvres	Running, step-ups, counter movement jumps and agility drill Protocol ceased when participant (a) could not attain 90% of maximal jump height on 3 consecutive jumps or (b) heart rate plateaued at 90% max heart rate for 3 consecutive circuits	**Kinematic findings**		
			1. Less knee flexion angle (°) at initial contact at 50% and 100% fatigue	17 ± 5 *vs* 16 ± 6 (@50% fatigue) and 14 ± 4 (@100% fatigue)	*P* = 0.004
			2. Less knee flexion angle (°) at peak stance contact at 50% and 100% fatigue	52.9 ± 5.6 *vs* 56.1 ± 7.2 (@50% fatigue) and 50.5 ± 7.1 (@100% fatigue)	*P* = 0.001
			3. Greater hip flexion angle (°) at initial contact at 50% and less at 100% fatigue	45.4 ± 10.9 *vs* 46.2 ± 11.2 (@50% fatigue) and 40.9 ± 11.3 (@100% fatigue)	*P* = 0.004
			4. Greater hip flexion angle (°) at peak stance at 50% and less at 100% fatigue	49.8 ± 9.9 *vs* 52.9 ± 12.1 (@50% fatigue) and 46.3 ± 12.9 (@100% fatigue)	*P* = 0.001
			5. Smaller hip abduction angle (°) at peak stance at 50% and at 100% fatigue	13.8 ± 6.6 *vs* 9.1 ± 6.5 (@50% fatigue) and 7.8 ± 6.5 (@100% fatigue)	*P* < 0.001
			Kinematic findings		
			6. Internal knee adductor moment ($Nm.kgm^{-1}$) smaller at 50% fatigue and greater at 100% fatigue	0.49 ± 0.23 *vs* 0.55 ± 0.23 (@50% fatigue) and 0.37 ± 0.24 (@100% fatigue)	*P* = 0.030

			7. Smaller hip adductor moment (Nm.kgm^{-1}) at peak stance at 50% and at 100% fatigue	0.14 ± 0.13 *vs* 0.08 ± 0.13 (@50% fatigue) and 0.06 ± 0.05 (@100% fatigue)	*P* = 0.007
Iguchi et al., (2014) 23 active participants (11 females and 12 males)	Unanticipated side cutting manoeuvres	Repeated counter movement jumps until participants could not jump 70% of baseline countermovement jump height for 2 consecutive jumps.	**EMG findings** (% of EMG activity in pre fatigue, anticipated side cuts) 1. Increased EMG activity of gluteus medius during first 50 ms of stance 2. Decreased EMG activity of semitendinosus 50ms before and after initial contact 3. Decreased EMG activity of semitendinosus during first 50ms of stance phase	 0.2 ± 22* *vs* 21.5 ± 48.3 1.9 ± 22* *vs* -6.2 ± 20.1 1.9 ± 22* *vs* -7.9 ± 26.6 (*approximate values from graph)	 *P* = 0.03, *d* = 0.49 *P* = 0.03, *d* = 0.51 *P* = 0.01, *d* = 0.58
			Kinematic findings	*Not significant*	
			Kinetic findings	*Not significant*	
Khalid et al., (2015) 12 collegiate soccer players (6 females and 6 males)	Anticipated and unanticipated side cutting manoeuvres	Yo-yo shuttle test. Discontinued when unable to complete two consecutive shuttles in the correct time	**Kinematic findings** 1. Smaller knee flexion angle (°) at initial contact **Kinetic findings** 2. Greater peak knee extensor moment 3. Smaller internal knee external rotator moment 4. Greater peak vertical GRF 5. Greater peak posterior GRF	 31.37 ± 11.72 *vs* 28.09 ± 9.33) 1.56 ± 4.2 *vs* 1.67 ± 0.34 0.23 ± 0.12 *vs* 0.17 ± 0.01 2.13 ± 0.58 *vs* 2.28 ± 0.60 0.75 ± 2.7 *vs* 0.83 ± 0.30	 *P* < 0.001 η^2 = 0.239 *P* = 0.038, η^2 = 0.083 *P* = 0.009 *P* = 0.020, η^2 = 0.104 *P* = 0.030, η^2 = 0.090

Borotikar et al., (2008) 25 female collegiate athletes	Single leg side jump/cutting manoevure	5 two legged squats and repeated trials until unable to complete 3 two legged squats unassisted	**Kinematic findings***		
			1. Greater peak stance knee abduction angle (°)at 50% and 100% fatigue	3.5 ± 3.2 *vs* 3.6 ± 3.3 (@50% fatigue) and 4.3 ± 3.1 (@100% fatigue)	$P < 0.001$
			2. Greater peak stance knee internal rotation angle (°) at 50% and 100% fatigue	13.1 ± 4.2 *vs* 14.1 ± 5.1 (@50% fatigue) and 14.4 ± 5.0 (@100% fatigue)	$P < 0.001$
			3. Smaller hip flexion angles (°) at 50% and 100% fatigue	31.0 ± 3.0 *vs* 28.9 ± 2.8 (@50% fatigue) and 27.2 ± 2.4 (@100% fatigue)	$P < 0.001$
			4. Greater hip internal rotation angles (°) at 50% and 100% fatigue	8.2 ± 2.1 *vs* 10.0 ± 1.9 (@50% fatigue) and 10.0 ± 2.2 (@100% fatigue)	$P < 0.001$
			5. Greater peak stance ankle supination angle (°) at 50% and 100% fatigue	9.1 ± 3.6 *vs* 13.0 ± 3.1 (@50% fatigue) and 13.0 ± 3.2 (@100% fatigue)	$P = 0.010$
			Kinetic findings * Kinematic values for dominant limb reported	Not investigated	
Lucci et al., (2011) 15 female collegiate soccer players	Unanticipated side cutting manoeuvres	Two fatigue protocol; Long – VO_2 peak test and 30 min interval run Fast – 4 circuits of running, step-ups, counter movement jumps and agility drill	**Kinematic findings** Both protocols:		
			1. Smaller knee flexion angle (°) at initial contact	25.8 ± 6.7 *vs* 22.6 ± 9.7 (Long) 25.5 ± 8.0 *vs* 22.3 ± 7.1 (Fast)	$P = 0.003$, $d = 0.65$
			2. Smaller knee flexion angle (°) at peak vertical GRF	41.9 ± 8.2 *vs* 40.1 ± 7.3 (Long) 41.6 ± 7.7 *vs* 37.4 ± 6.8 (Fast)	$P = 0.017$, $d = 0.31$
			3. Smaller knee flexion angle (°) at peak stance	54.5 ± 5.1 *vs* 51.9 ± 6.4 (Long)	$P = 0.001$, $d = 0.59$

		53.1 ± 7.0 *vs* 48.3 ± 7.4 (Fast)	
	4. Greater knee internal rotation angle (°) at initial contact	14.1 ± 6.2 *vs* 10.9 ± 5.6 (Fast) 12.8 ± 5.9 *vs* 14.1 ± 5.2 (Long)	$P < 0.001$, $d = 0.78$
	5. Greater knee internal rotation angle (°) at peak posterior GRF	11.4 ± 5.2 *vs* 15.1 ± 4.8 (Fast) 36.2 ± 8.7 *vs* 32.6 ± 9.3 (Long)	$P = 0.037$, $d = 0.45$
	6. Smaller hip flexion angle (°) at initial contact	36.5 ± 8.0 *vs* 28.1 ± 9.3 (Fast) 38.2 ± 9.2 *vs* 33.9 ± 9.5 (Long)	$P = 0.022$, $d = 0.38$
	7. Smaller hip flexion angle (°) at peak stance	38.3 ± 8.8 *vs* 29.3 ± 9.7 (Fast) 13.2 ± 9.4 *vs* 10.1 ± 8.3 (Long)	$P = 0.001$, $d = 0.73$
	8. Smaller hip internal rotation angle (°) at initial contact	9.7 ± 6.6 *vs* 5.7 ± 9.4 (Fast) 7.8 ± 9.4 *vs* 3.8 ± 10.9 (Long)	$P = 0.031$, $d = 0.33$
	9. Smaller hip internal rotation angle (°) at peak vertical GRF	4.4 ± 7.5 *vs* 0.3 ± 8.9 (Fast) 10.5 ± 9.5 *vs* 6.3 ± 10.5 (Long)	$P = 0.018$, $d = 0.40$
	10. Smaller hip internal rotation angle (°) at peak posterior GRF	9.2 ± 6.5 *vs* 3.1 ± 9.1 (Fast) 3.1 ± 9.9 *vs* -0.75 ± 11.4 (Long)	$P = 0.01$, $d = 0.49$
	11. Smaller hip internal rotation angle (°)	-0.48 ± 7.0 *vs* -4.5 ± 8.2	$P = 0.007$,

			at peak stance	(Fast)	*d* = 0.20
			Kinetic findings		
			12. Smaller peak knee extensor moment ($Nm.kg^{-1}$)	2.1 ± 0.3 *vs* 1.95 ± 0.27 (Long) 1.92 ± 0.3 *vs* 1.90 ± 0.29 (Fast)	*P* = 0.015, *d* = 0.34
McLean and Samorezov (2009) 20 female collegiate athletes	Anticipated and unanticipated single leg side jumps	Single leg squats and single leg side jumps until unable to perform squats	**Kinematic findings**		
			1. Smaller knee flexion angle (°) at initial contact	16.0 ± 2.0 *vs* 12.1 ± 2.3 (@50% fatigue) and 10.0 ± 2.2 (@100% fatigue)	*P* < 0.01
			2. Greater knee abduction angles (°) at peak stance	5.1 ± 3.6 *vs* 4.6 ± 3.9 (@50% fatigue) and 4.5 ± 3.2 (@100% fatigue)	*P* < 0.01
			3. Greater hip internal rotation angles (°) at peak stance	8.8 ± 4.7 *vs* 10.1 ± 3.8 (@50% fatigue) and 9.6 ± 4.9 (@100% fatigue)	*P* < 0.01
			Kinematic findings		
			4. Smaller knee extensor moments at peak stance (Nm)	142.9 ± 18.1 *vs* 119.0 ± 21.0 (@50% fatigue) and 99.9 ± 23.8 (@100% fatigue)	*P* < 0.01
			5. Smaller knee adductor moments at peak stance (Nm)	40.1 ± 7.5 *vs* 41.3 ± 5.0 (@50% fatigue) and 38.0 ± 5.1 (@100% fatigue)	*P* < 0.01
			6. Greater hip external rotator moments at peak stance (Nm)	8.7 ± 3.4 *vs* 15.0 ± 6.3 (@50% fatigue) and 15.9 ± 5.2 (@100% fatigue)	*P* < 0.01
O'Connor et al., (2015) 11 female volunteers	Single leg counter movement jumps and vertical or lateral stride	Isokinetic fatiguing protocol for hamstrings. Protocol ceased when torque decreased to 25%	**Kinematic findings**	*Principal component analysis of stance phase completed. Discrete pre and post values not*	
			1. Smaller knee flexion angles		*P* = 0.002
			Kinetic findings		
			2. Smaller knee peak extensor moment		*P* = 0.011

	landings.	of baseline	3. Smaller knee peak power	*provided.*	*P* = 0.020
Potter et al., (2014) 19 female, collegiate soccer players	Cutting activity during agility run	Repeated agility runs. Protocol ceased when circuit time increased by over 1 standard deviation above baseline time.	**Kinematic findings**		
			1. Smaller knee flexion angles (°) at 32 ms post initial contact	Side cut; pre 36.9 ± 12.1 *vs* post 32.3 ± 10.5 Crossover cut; pre 51.9 ± 20.9 *vs* post 44.9 ± 19.6	*P* = 0.016
			2. Smaller knee abduction angles (°) at 32 ms post initial contact	Side cut; pre -3.7 ± 5.0 *vs* post -1.3 ± 6.5 Crossover cut; pre 0.7 ± 4.5 *vs* post 2.6 ± 6.5	*P* = 0.017
			3. Smaller hip flexion angles (°) at 32 ms post initial contact	Side cut; pre 38.1 ± 10.1 *vs* post 34.3 ± 12.3 Crossover cut; pre 46.4 ± 11.1 *vs* post 40.4 ± 18.0	*P* = 0.010
			Kinetic findings	Not investigated	
Raja Azidin et al., (2015) 19 male, collegiate soccer players	Side cutting manoeuvres	Soccer specific protocols lasting 45 mins with one on a treadmill and one overground	**Kinematic findings*** Smaller peak knee flexion angle (°) during weight acceptance phase		
			a. post 60 mins (45 mins exercise and 15 minute rest) v baseline	Overground; baseline 14.9 ± 5.0 *vs* 60 mins 13.1 ± 4.5 Treadmill; baseline 14.7 ± 4.7 *vs* 60 mins 13.2 ± 4.8	*P* = 0.027
			b. post 45 mins exercise versus post 60 mins (additional 15 minute rest following exercise)	Overground; 45 mins 13.5 ± 4.5 *vs* 60 mins 13.1 ± 4.5 Treadmill; 45 mins 14.9 ± 5.1 *vs* 60 mins 13.2 ± 4.8	*P* = 0.009
			Kinetic findings	Not significant	
			* Approximate values taken from graph		
Sanna and O'Connor, (2008)	Side cutting manoeuvres	Soccer specific protocol lasting 60 minutes	**Kinematic findings**		
			1. Greater knee internal rotation range of motion (°) during side cut	13.7 2.4 *vs* 17.0 ± 4.2	*P* = 0.017

12 female, collegiate soccer players			**Kinetic findings**	Not significant	
Tsai et al., (2009) 15 female, recreational athletes	Eccentric phase of side cutting manoeuvre	Circuits of sprints and vertical jump squats. Ceased when unable to jump to 50% of baseline	**Kinematic findings** 1. Greater peak knee valgus angle (°) 2. Greater peak knee internal rotation angle (°) **Kinetic findings** 1. Greater peak knee adductor moment ($Nm.kg^{-1}$)	2.4 ± 3.0 greater post fatigue 4.9 ± 2.7 greater post fatigue 0.3 ± 0.3 greater post fatigue	$P < 0.05$ $P < 0.05$ $P < 0.05$
Chappell et al., (2005) 20 recreationally active participants (10 females and 10 males)	Forward stop jump Vertical stop jump Backward stop jump	Circuit of 5 vertical jumps followed by 30 metre sprint. Circuit ceased at volitional exhaustion	**Kinematic findings** 1. Smaller knee flexion angle (°) **Kinetic findings** 2. Greater peak anterior tibiofemoral shear force (BW) 3. Greater knee adductor moment ($Nm.kgm^{-1}$)	29.9 *vs* 25.7 0.24 *vs* 0.29 0.026 *vs* 0.051 * standard deviations not provided	$P = 0.030$ $P = 0.010$ $P = 0.030$
Kim et al., (2015) 21 physically active participants (7 females and 14 males)	Forward side jump	Circuit of 5 minute runs, 20 seconds of lateral counter movement jumps and 20 vertical counter movement jumps. Circuits ceased when participants reported 17 on RPE scale and the vertical counter movement jump height fell below 80% of baseline	**Kinematic findings** 1. Smaller knee flexion angle (0-10% of stance) 2. Smaller hip flexion angle (0-10% of stance) 3. Smaller plantarflexion angle (0-15% of stance) **Kinetic findings** 4. Greater knee extensor moment (5-15% of stance) 5. Smaller knee extensor moment (20-30% of stance)	A functional analysis of variance was conducted displaying differences over percentages of the stance phase and did not report specific magnitude differences	$P < 0.05$ $P < 0.05$ $P < 0.05$ $P < 0.05$ $P < 0.05$

			6. Smaller hip extensor moment (5-10% of stance)		*P* < 0.05
			7. Greater hip extensor moment (20-30% of stance)		*P* < 0.05
			EMG findings		
			8. Increased vastus lateralis EMG activity (15% of stance)		*P* < 0.05
			9. Decreased hamstring EMG activity (25-60% of stance)		*P* < 0.05
			10. Decreased gluteus maximus EMG activity (0-5% of stance)		*P* < 0.05
			11. Decreased tibialis anterior EMG activity (30-40% of stance)		*P* < 0.05
Quammen et al., (2012) 15 collegiate, female soccer players	Running stop jump	Two fatigue protocol; Long – VO_2 peak test and 30 min interval run Fast – 4 circuits of running, step-ups, counter movement jumps and agility drill	**Kinematic findings**		
			1. Smaller hip flexion angle (°) at initial contact	50.1 ± 9.5 *vs* 44.7 ± 8.1	*P* = 0.001 d = 0.57
			2. Smaller hip flexion angle (°) at peak vertical GRF	50.4 ± 10.3 *vs* 44.7 ± 8.4	*P* = 0.001 d = 0.55
			3. Smaller hip flexion angle (°) at peak posterior GRF	51.1 ± 10.8 *vs* 45.2 ± 8.6	*P* = 0.001 d = 0.55
			4. Smaller hip flexion angle (°) at peak knee flexion angle	45.1 ± 11.6 *vs* 38.7 ± 8.7	*P* = 0.001 d = 0.55
			5. Smaller peak hip flexion angle (°)	53.3 ± 10.9 *vs* 47.3 ± 8.2	*P* = 0.001 d = 0.55
			6. Smaller knee flexion angle (°) at peak vertical GRF	56.8 ± 52.5 *vs* 44.7 ± 8.4	*P* < 0.001 d = 0.52
			7. Smaller knee flexion angle (°) at peak posterior GRF	38.8 ± 5.03 *vs* 35.9 ± 6.5	*P* = 0.001 d = 0.48
			8. Smaller peak knee flexion angle (°)	38.4 ± 5.6 *vs* 35.8 ± 7.2	*P* = 0.009 d = 0.39
			9. Greater hip adduction angle (°) at peak knee flexion in fast compared with long protocol	3.8 ± 4.6 *vs* 1.8 ± 3.9	*P* = 0.03

			Kinetic findings 10. Greater knee adductor moment ($Nm.kg^{-1}$) at initial contact in fast compared with long protocol 11. Smaller peak knee adductor moment ($Nm.kg^{-1}$) in fast compared with long protocol	0.064 ± 0.09 *vs* 0.024 ± 0.06 2.01 ± 0.32 *vs* 1.9 ± .36	d = 0.41 *P* = 0.03 d = 0.44 *P* = 0.02 d = 0.31
Cortes et al. (2014a) 18 female, collegiate soccer players	Unanticipated crossover cutting	Running, step-ups, counter movement jumps and agility drill Protocol ceased when unable to jump 90% of baseline jump height	**Kinematic variables** 1. Decreased knee flexion angle at initial contact (°) 2. Decreased knee adduction angles at initial contact	32 ± 9 *vs* 29 ± 11 (@50% fatigue) and 22 ± 9 (@100% fatigue) 9 ± 5 *vs* 8 ± 4 (@50% fatigue) and 6 ± 4 (@100% fatigue)	*P* < 0.001 *P* = 0.015 *P* = 0.006 *P* = 0.015
Collins et al. (2012) 13 female, collegiate, soccer players	Anticipated and unanticipated side cutting	60 minute running protocol with varying levels of intensity	**Kinematic Variables** 1. Increase in peak knee abduction angles (°)* * Values reported for anticipated side cutting	ANT 2.2 ± 4.7 *vs* UNA 3.8 ± 5.0	*P* = 0.020

GRF = Ground reaction force, EMG = Electromyography, MVIC = Maximal voluntary isometric contraction, RPE = Rate of perceived exertion

Table 2.3 A summary of the biomechanical effects of fatigue on cutting manoeuvres

Variable	Sagittal	Frontal	Transverse
Hip Kinematics	↓ Flexion	↑ Abduction	↑ Internal rotation
Hip Moments	↓ Extensor	↑ Adductor	↑ External rotator
Knee Kinematics	↓ Flexion	↑ Abduction	↑ Internal rotation
Knee Moments	↑ Extensor	↑ Adductor	

Internal joint moments reported

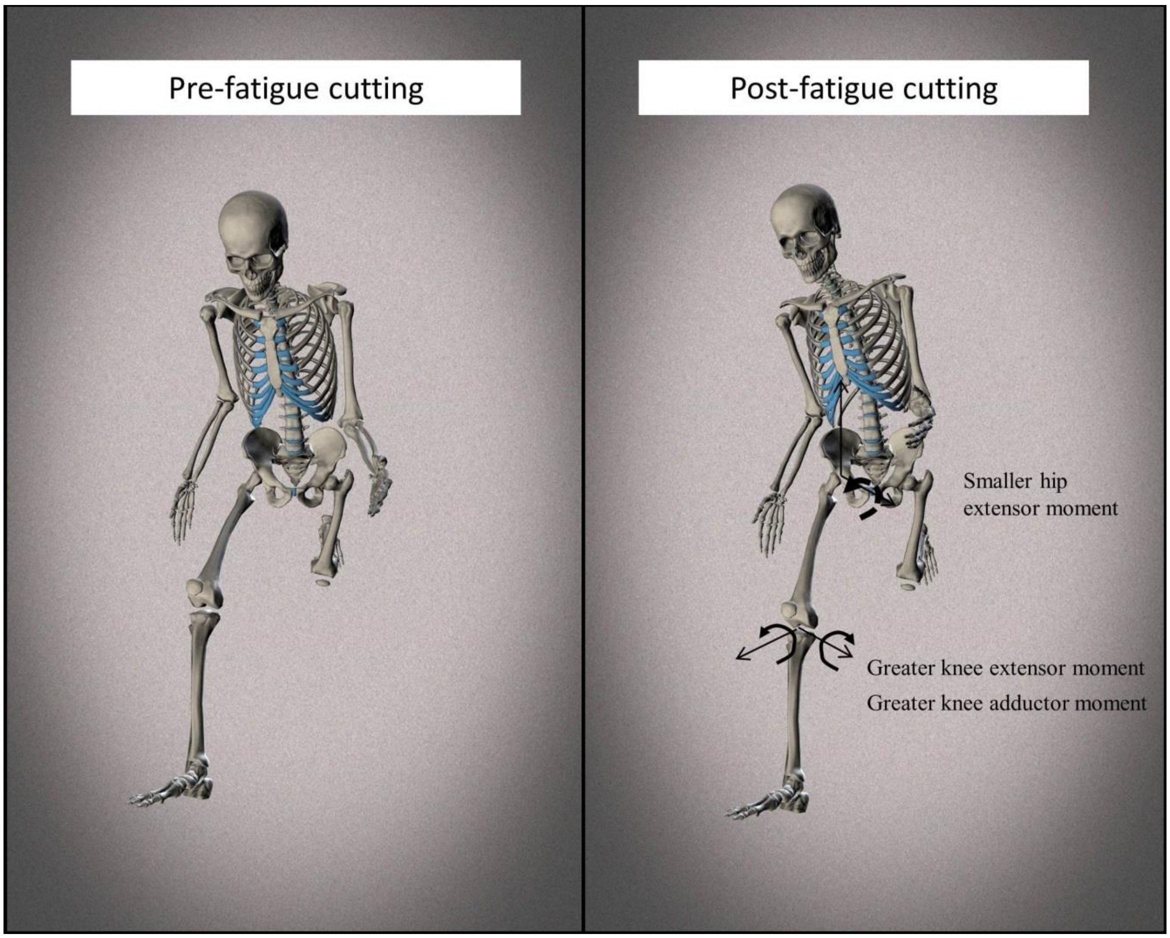

Figure 2.7 The biomechanical effects of fatigue on cutting manoeuvres

These findings suggest that fatigue leads to an extended hip and knee posture during cutting and stop jumping, a neuromuscular pattern which causes the initial decrease in hip extensor activity and moment and a corresponding increase in knee extensor activity and moment. This potentially increases ACL loading further as greater GRFs are evident post fatigue in cutting activities (Bell et al., 2016; Khalid et al., 2015). Landing with smaller hip and, in particular, knee flexion angles and the subsequent increase in GRF lead to an increase in knee extensor moment (Blackburn and Padua, 2008; McLean and Samorezov, 2009; Myers et al., 2012). Greater knee extensor moments lead to a greater anterior tibiofemoral shear force (Sell et al., 2007), which directly loads the ACL (Butler et al., 1980; Markolf et al., 1995). This effect is greatest at small knee flexion angles in particular (Chappell et al., 2005; Yu et al., 2006) as the hamstrings are at a mechanical disadvantage and therefore have a reduced proximal posterior tibial shear effect (Pandy and Shelburne, 1997). Furthermore, fatiguing protocols have been shown to lead to a decrease in hamstring EMG pre-activity and activity levels during stance (Gehring et al., 2009; Iguchi et al., 2014; H. Kim et al., 2015) and an increase in quadriceps EMG activity (Kim et al., 2015). Therefore fatigue may alter the dynamic stability of the knee joint during landing and cutting activities. These sagittal plane, fatigue-induced negative effects on landing posture and muscle activity patterns may therefore combine to increase ACL strain and subsequent injury risk. However, concurrent sagittal, frontal and transverse plane loading is required to sufficiently load the ACL and place it at risk of injury (McLean et al., 2004; Oh et al., 2012; Oh et al., 2012; Shin et al., 2009).

In contrast, the effect of fatigue on sagittal plane kinematics in drop jumps and drop landings is less consistent (Table 2.4). Although some studies demonstrated that fatigue results in decreased hip and knee flexion during drop jumps (Benjaminse et al., 2008; Chappell et al., 2005; O'Connor et al., 2015; Quammen et al., 2012), other studies have found that it results in increased knee and/or hip flexion (Gehring et al., 2009; Kernozek et al., 2008; Liederbach et al., 2014) and smaller hip and knee extensor moments and tibiofemoral joint forces (Kernozek et al., 2008). These findings were reported in studies that examined the effect of fatigue on drop landings (Gehring et al., 2009; Kernozek et al., 2008; Liederbach et al., 2014) which require participants to decelerate the centre of mass only. On the other hand, during drop jump activities, participants are required to accelerate the centre of mass following the initial deceleration which may explain the inconsistencies in the findings of different studies. This shows that the effects of fatigue are likely to be task specific and therefore the tasks being investigated should be as ecologically valid as possible.

Fatigue has a greater effect on knee biomechanics in the frontal and transverse planes during change of direction activities such as cutting (Table 2.2) compared with drop jumping and landing activities (Table 2.4). In contrast to studies that a functional fatigue protocol resulted in greater peak internal rotation (McLean et al., 2007) and abduction angles and greater external rotator and adductor moments (Collins et al., 2016; McLean et al., 2007), the majority of other studies either found no effect on frontal and transverse plane knee kinematics or a decrease in potential ACL loading during drop jump and drop landing activities (Table 2.4). On the other hand, fatigue tends to have a greater effect on frontal and transverse plane biomechanics during cutting activities as seen by the fatigue-induced increase in knee abduction angles (Borotikar et al., 2008; McLean and Samorezov, 2009; Shultz, Schmitz, Cone et al., 2015) adductor moments (Chappell et al., 2005; Cortes et al., 2013; McLean et al., 2007; McLean and Samorezov, 2009; Tsai et al., 2009) and internal rotation angles (Borotikar et al., 2008; Cortes et al., 2012; Lucci et al., 2011; Sanna and O'Connor, 2008; Schmitz et al., 2015; Tsai et al., 2009). These changes, in conjunction with the aforementioned sagittal plane alterations, may combine to increase potential ACL loading and injury risk. The fatigue-induced biomechanical alterations in the frontal and sagittal planes may be due to a number of factors. Firstly, fatigue results in decreased semitendinosus EMG activity during side cutting (Iguchi et al., 2014) which may reduce rotational and medial knee stability. In addition, a large proportion of the variance in knee frontal and transverse plane loading can be explained by hip and trunk biomechanics (Frank et al., 2013). Therefore, the influence of fatigue on hip and trunk biomechanics in the frontal and transverse planes also needs to be investigated.

The effect of fatigue on hip frontal and transverse biomechanics during activities with horizontal (e.g. cutting) and vertical deceleration (e.g. drop landing and drop jumping) activities is unclear, partially due to the fact that hip frontal and transverse plane biomechanics are not routinely investigated (Table 2.4). Where it has been investigated, fatigue has been found to lead to an increase in hip adduction angles during cutting manoeuvres (Cortes et al., 2013; Quammen et al., 2012), which may contribute to increased knee abduction angles, or not to have any effect (Borotikar et al., 2008; Cortes et al., 2012; Lucci et al., 2011; McLean and Samorezov, 2009). Similarly, the effect of fatigue on transverse plane hip biomechanics is inconsistent as it has been found to increase hip internal rotation angle (Borotikar et al., 2008; McLean and Samorezov, 2009) and peak hip external rotator moments (McLean and Samorezov, 2009) during side cutting activities. However, Lucci et al. (Lucci et al., 2011) found that both a long and short duration fatigue protocol resulted in decreased hip internal rotation angles during side cutting. The reason for these conflicting results may be that McLean et al. (McLean and Samorezov, 2009) terminated the fatigue protocol when the athlete reached volitional exhaustion and was unable to continue whereas Lucci et al. (Lucci et al., 2011) terminated the fatigue protocols after a distinct time or number of repetitions (30 minutes for the long protocol and 4 repetitions for the fast protocol). Given that restrictions in hip range of motion (Bedi et al., 2016) and strength (Khayambashi et al., 2015) are associated with, and predict ACL injuries respectively, a greater understanding of the effect of fatigue on hip biomechanics is warranted in order to optimise ACL IPPs.

Although the effects of fatigue on the frontal and transverse biomechanics of the hip are unclear, fatigue has been found to alter the EMG activity of the hip abductors during landing and cutting activities. Patrek et al. (Patrek et al., 2011) also found that a local hip abductor fatiguing protocol lead to a decrease in activation latency of the gluteus medius during single leg drop landings without a decrease in integrated or peak EMG measurements. The increased latency is effectively a reduction in the anticipatory activation of the gluteus medius which may be due to a central fatiguing process (Patrek et al., 2011) and may affect the an athlete's ability to stabilise the lower limb during anticipated drop landings. It must be noted that although the fatiguing protocol delayed the EMG activity of the gluteus medius and led to a decrease in peak hip abductor moment, it did not alter knee joint biomechanics. This may be due to the fact that the drop landing task is not sufficiently demanding of frontal hip control. On the other hand, Iguchi et al. (Iguchi et al., 2014) found that fatigue resulted in greater integrated EMG activity of the gluteus medius muscle during the first 50ms of cutting manoeuvres, where greater frontal plane control is required. This suggests that the frontal plane activity of the hip increases post fatigue during cutting manoeuvres and that the hip may play an important protective role during such conditions. However, the effect of anticipation and fatigue combined on muscle activity during cutting is not well understood and should be investigated further.

Table 2.4 The effect of fatigue on neuromuscular and biomechanical risk factors for anterior cruciate ligament injuries during landing

Study and Participants	Activity	Fatiguing methodology	Effect of Fatigue	Magnitude of differences	Statistical value
Bell et al., (2016) 40 recreationally active (20 females and 20 males)(Bell et al., 2016)	Drop jump	High intensity running protocol with numerous changes of direction, squatting and jumping exercises. Protocol ceased when 17 reported on RPE scale	**Kinematic findings** 1. Increase in LESS for males and females due to decreased hip and knee flexion and increased trunk side flexion	Females; pre – 5.33 ± 2.11 *vs* post – 6.69 ± 2.61 Males; pre – 4.70 ± 2.33 *vs* post – 6.34 ± 2.26	$P < 0.001$
			Kinetic findings Significant time (pre vs post) sex interaction effects 2. Increase in peak vGRF (BW) in males post exercise compared with females	Females; pre – 2.66 ± 0.58 *vs* post – 2.75 ± 0.45 Males; pre – 2.40 ± 0.61 *vs* post 29.7 ± 0.75	$P = 0.020$
			3. Increased rate of vGRF loading (BWs^{-1}) in males post exercise compared with females	Females; pre – 80.43 ± 25.31 *vs* post – 93.52 ± 30.86 Males; pre – 73.39 ± 29.49 *vs* post 104.93 ± 33.58	$P = 0.008$
Mclean et al., (2007) 20 collegiate soccer players (10 female and 10 male)	Drop jump (first landing)	4 minutes of repeated circuit consisting of 20 step ups and repeated plyometric bounds	**Kinematic findings *** 1. Greater peak stance knee abduction angle (°)	Females; pre 3.4 ± 4.4 *vs* post 10.2 ± 5.0 Males; pre 1.6 ± 1.5 *vs* post 6.2 ± 1.9	$P < 0.001$
			2. Greater peak stance knee internal rotation angle (°)	Females; pre 12.5 ± 7.1 *vs* post 19.7 ± 9.4 Males; pre 4.9 ± 1.9 *vs* post 12.1 ± 3.6	$P < 0.001$
			3. No effect on hip kinematics		

			Kinetic findings 4. Greater peak knee adductor moment ($Nm.kgm^{-1}$) 5. Greater peak knee external rotator moment ($Nm.kgm^{-1}$) 6. No effect on hip kinetics * Kinematic and kinetic values for dominant limb reported	Females; pre 0.13 ± 0.06 *vs* post 0.34 ± 0.11 Males; pre 0.10 ± 0.08 *vs* post 0.10 ± 0.06 Females; pre 0.11 ± 0.06 *vs* post 0.28 ± 0.20 Males; pre 0.08 ± 0.03 *vs* post 0.16 ± 0.06	$P < 0.001$ $P < 0.001$
O'Connor et al., (2015) 11 female volunteers	Single leg counter movement jumps and vertical or lateral stride landings. Principal component analysis	Isokinetic fatiguing protocol for hamstrings. Protocol ceased when hamstring torque decreased to 25% of baseline	**Kinematic findings** 1. Smaller knee flexion angles **Kinetic findings** 2. Smaller knee peak extensor moment 3. Smaller knee peak power	*Principal component analysis of stance phase completed. Discrete pre and post values not provided.*	$P = 0.002$ $P = 0.011$ $P = 0.020$
Benjaminese et al., (2008) 30 recreationally active participants (15 females and 15 males)	Single leg standing jump	Astrand protocol – progressive running, treadmill protocol to volitional exhaustion	**Kinematic findings** 1. Smaller knee flexion angle (°) at initial contact 2. Smaller peak knee valgus angle (°) 3. No effect on hip kinematics **Kinetic findings**	13.6 ± 5.3 *vs* 11.5 ± 5.8 3.5 ± 3.7 *vs* 2.7 ± 3.5 *Not investigated*	$P = 0.009$ $P = 0.038$
Moran and Marshall (2006) 15 physically active males	30 cm and 50 cm drop jump	Progressive treadmill running protocol. Protocol ceased when 17 reported on RPE scale	**Kinematic findings** 1. Greater knee joint angular velocity at 30cm height **Kinetic findings** Greater peak tibial accelerations at 30cm height	21% greater post fatigue 24% greater post fatigue	$P < 0.001$ $P = 0.004$
Shultz et al., (2015)	Drop jump (30cm) (analysis from initial	Intermittent exercise running protocol (90	1. Greater internal knee rotation motion in females associated with	Discrete baseline and post-fatigue values were not	$P = 0.002$

59 soccer players (29 females and 30 males)	contact to peak centre of mass displacement)	minutes). Principal component analysis and hierarchical linear modelling was used to analyse patterns of motion and their relationship with fatigue.	anterior posterior laxity from exercise 2. Greater hip flexion motion rotation in males associated with anterior posterior laxity from exercise 3. Greater knee valgus and dorsiflexion motion in females associated with rotational laxity from exercise 4. Greater hip external rotation, knee internal rotation and valgus motion in males associated with rotational laxity from exercise 5. Greater knee extensor moment in males associated with anterior posterior laxity from exercise	presented	$P = 0.006$ $P < 0.050$ $P = 0.030$ $P = 0.040$
Patrek et al. (2011) 20 physically active females	Single leg drop landing from 40cm. Hip and knee frontal and sagittal plane kinematics at initial contact and 60ms after initial contact. Peak hip and knee kinetics and glut medius EMG in this phase	Local hip abductor fatiguing exercise. Repeated hip abduction at 60 repetitions per second until 19 reported on RPE scale or unable to complete 2 consecutive repetitions	**Kinematic findings** 1. Pelvic, hip and knee kinematics unchanged **Kinetic findings** 2. Decreased hip internal rotator moments (Nm) 3. Decreased hip abductor moment (Nm) **EMG findings** 4. Delayed activity of gluteus medius	Not significant 40.2 ± 20.0 *vs* 29.8 ± 17.2 41.7 ± 20.4 *vs* 31.6 ± 16.7 32 millisecond delay	$P < 0.001$ ES = 0.56 $P < 0.001$ ES = 0.54
Liederbach et al., (2014) 40 ballet dancers (20 female, 20 male)	Single leg drop landings (30cm)	Series of 50 step ups, 15 maximal effort single leg jumps Protocol ceased when unable to jump 90% of	**Kinematic findings** 1. Increased peak knee flexion angles (°) **Kinetic findings** 2. Smaller peak knee extensor	54.3 ± 1.4 *vs* 57.6 ± 1.7 2.7 ± 0.1 *vs* 2.4 ± 0.1	$P < 0.001$ $P = 0.003$

and 40 field sport athletes (20 female, 20 male)		maximal vertical jump height	moments (Nm.kg^{-1})		
Kernozek et al. (2008) 14 female and 16 male athletes	Single leg drop landing (50cm)	Repeated parallel squat exercises at 60% repetition maximum until failure	**Kinematic findings**		
			1. Greater peak knee flexion (°), particularly in males	Females; pre 64.3 ± 10.5 *vs* post 64.2 ± 10.5 Males; pre 67.2 ± 11.8 *vs* post 73.8 ± 10.9	$P = 0.025$
			2. Greater peak hip flexion angle (°)	Females; pre 40.7 ± 9.6 *vs* post 48.0 ± 14.4 Males; pre 26.7 ± 1.4 *vs* post 31.7 ± 12.5	$P = 0.012$
			3. Greater ankle dorsiflexion angle (°)	Females; pre 23.6 ± 4.7 *vs* post 26.0 ± 5.1 Males; pre 24.3 ± 8.0 *vs* post 25.7 ± 8.0	$P = 0.007$
			Kinetic findings		
			4. Smaller knee extensor moment (Nm.kg^{-1})	Females; pre 1.44 ± 0.64 *vs* post 1.05 ± 0.31 Males; pre 1.39 ± 0.33 *vs* post 1.13 ± 0.31	$P < 0.001$
			5. Smaller knee abductor moment (Nm.kg^{-1})	Females; pre 1.66 ± 0.46 *vs* post 1.13 ± 0.64 Males; pre 1.55 ± 0.53 *vs* post 1.33 ± 0.4	$P = 0.014$
			6. Smaller tibiofemoral compression force (% BW)	Females; pre 2.14 ± 0.28 *vs* post 2.0 ± 0.25 Males; pre 2.08 ± 0.21 *vs* post 1.8 ± 0.19	$P < 0.001$
			7. Smaller anterior tibiofemoral shear force (%BW)	Females; pre 0.95 ± 0.2 *vs* post 0.76 ± 0.15 Males; pre 1.0 ± 0.13 *vs* post	$P < 0.001$

				0.62 ± 0.11	
			8. Smaller hip extensor moment ($Nm.kg^{-1}$)	Females; pre 2.7 ± 2.6 *vs* post 2.0 ± 0.8 Males; pre 2.1 ± 0.8 *vs* post 1.5 ± 0.7	*P* = 0.019
Gehring et al. (2009) 13 female athletes and 13 male athletes	Two legged drop landing from 52cm	Leg press machine with 50% repetition maximum until failure	**Kinematic differences**		
			1. Greater peak knee flexion angle (°)	Females; pre 87.6 ± 2.7 *vs* post 90.0 ± 2.0 Males; pre 71.8 ± 3.1 *vs* post 76.3 ± 3.7	*P* = 0.004
			Kinetic differences		
			2. Decreased GRF impulse during first 50ms after initial contact	Females; pre 2.5 ± 0.1 *vs* post 2.3 ± 0.1 Males; pre 2.6 ± 0.1 *vs* post 2.4 ± 0.1	*P* = 0.004
			3. Decreased GRF impulse during first 100ms after initial contact	Females; pre 2.8 ± 0.1 *vs* post 2.8 ± 0.1 Males; pre 3.0 ± 0.1 *vs* post 2.8 ± 0.1	*P* = 0.010
			EMG differences		
			4. Decreased pre-activation of the biceps femoris (%MVIC)	22% reduction	*P* < 0.001
			5. Decreased initial activity of the biceps femoris	11% reduction	*P* = 0.045
			6. Decreased pre-activation of the semitendinosus	21% reduction	*P* = 0.020
			7. Decreased pre-activation of the medial gastrocnemius	10% reduction	*P* = 0.030

EMG: electromyography, RPE: Rate of perceived exertion, LESS: Landing error scoring system

Table 2.5 A summary of the biomechanical effects of fatigue on landing

Variable	Sagittal	Frontal
Trunk Kinematics	↑ Flexion	↑ Lateral flexion
Hip Kinematics	↑ Flexion	
Knee Kinematics	↑ Flexion	

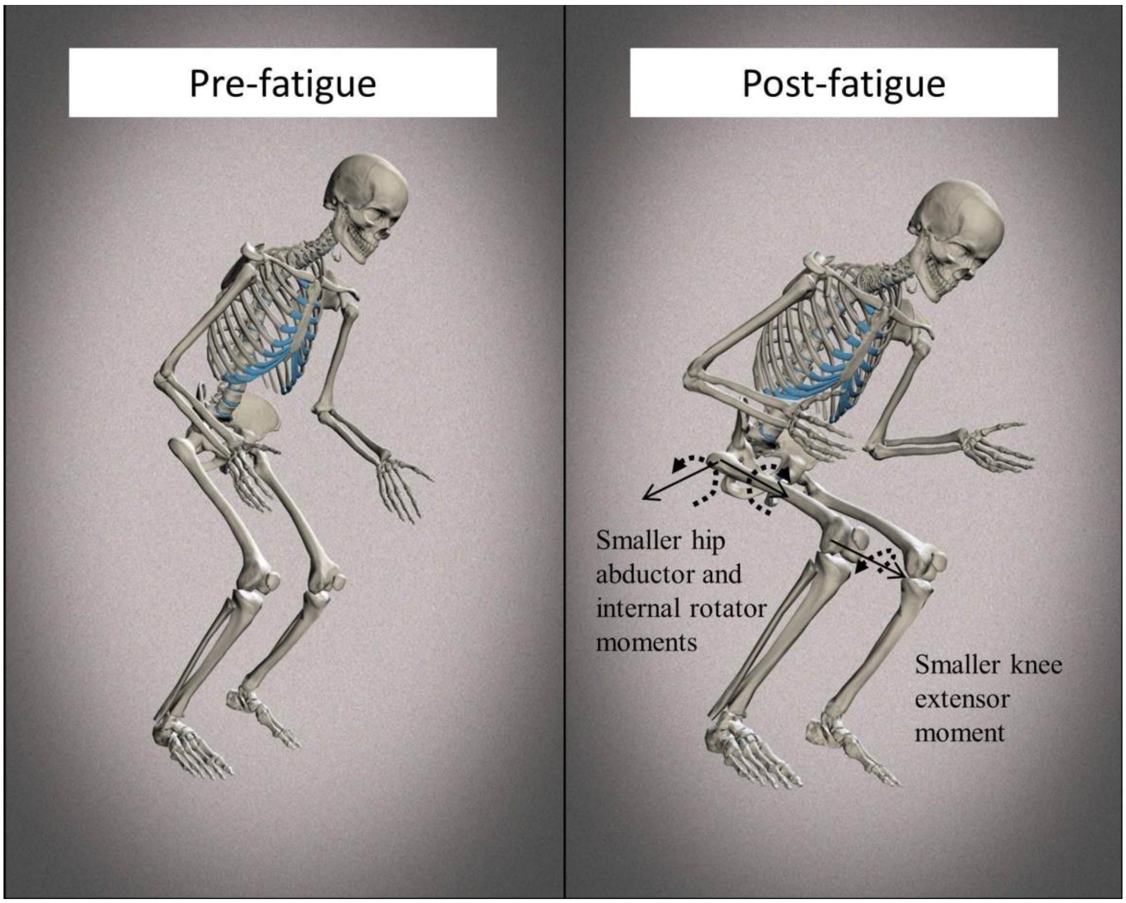

Figure 2.8 The biomechanical effects of fatigue on landing

Although there have been a number of studies examining the effect of fatigue on trunk kinematics during drop landing and jumping activities (Table 2.6), no studies have examined the effect during change of direction activities such as cutting. This is a significant limitation to our understanding of the effects of fatigue on the biomechanics of cutting, given the observed relationship between suboptimal trunk control and ACL loading (Hewett et al., 2009; Koga et al., 2010; Zazulak et al., 2007). Consequently, our understanding of the effects of fatigue on trunk kinematics must be garnered from studies examining the effect of fatigue on trunk kinematics during tasks such as running, single leg squat and landing despite the fact that the effect of fatigue is task dependent. Fatigue has generally been found to increase trunk flexion during drop jumps, drop landings (Lessi and Serrao, 2017; Liederbach et al., 2014), treadmill running (Maas et al., 2017), counter movement jumps (McNeal et al., 2010), and single leg squatting (Weeks et al., 2015) although it was not found during landing in Irish dancing (Wild et al., 2017). Increased trunk flexion results in a decrease in internal knee extensor moment (Frank et al., 2013; Shimokochi et al., 2013), EMG activity (Blackburn and Padua, 2009) and subsequent ACL strain (Markolf et al., 1995; Weinhandl et al., 2013). This suggests that fatigue may act to decrease knee loading by increasing trunk flexion with subsequent greater hip loading as trunk flexion results in greater internal hip extensor moment (Frank et al., 2013; Shimokochi et al., 2013) and subsequent gluteal activity (Lessi et al., 2017). In direct contrast to these findings, Bell et al. (Bell et al., 2016) found that a functional fatiguing protocol resulted in decreased trunk flexion during drop jumps which may at least partially explain the greater vertical GRF and increased rate of vertical GRF loading observed in a fatigued state. This indicates that fatigue leads to decreased trunk flexion and a subsequent increase in ACL loading during activities with higher loading that may more closely replicate sporting activities. However more study is required on this possible effect of fatigue particularly during cutting activities

Functional fatigue protocols resulted in increased lateral trunk flexion during drop jumps (Bell et al., 2016), single leg landings (Liederbach et al., 2014) and single leg squats (Weeks et al., 2015). This is particularly important given that a deficit in frontal plane control is a strong predictor of ACL injury (Zazulak et al., 2007). It has also been found that fatigue leads to increased contralateral pelvic drop or side flexion during single leg stance activities such as drop landing (Lessi and Serrao, 2017; Liederbach et al., 2014), running (Maas et al., 2017) and single leg squatting (Weeks et al., 2015). Changes in pelvic frontal plane kinematics are likely to lead to changes in trunk frontal kinematics although this was not specifically investigated in these studies and therefore should be explored in the future. Similarly, fatigue has been found to induce increases in trunk and pelvic rotation during single leg squatting (Weeks et al., 2015) and running (Maas et al., 2017). Increased trunk rotation has been found to increase knee external rotator moments which increase potential ACL loading (Dempsey et al., 2007; Frank et al., 2013; Jamison et al., 2012). In conclusion, there is a scarcity of research investigating the effects of fatigue on trunk kinematics during high loading activities that mimic high risk sporting activities in terms of ACL injury. The majority of research investigated the effects of fatigue on trunk kinematics drop landings, running or single legs squats and has found alterations in trunk sagittal and frontal plane control and pelvic frontal and transverse control. The understanding of fatigue on trunk and pelvic kinematics during cutting manoeuvres where rotational control of the centre of mass is severely limited and should be investigated.

Fatigue protocols that are terminated at volitional exhaustion or when an athlete is unable to complete a certain task may be excessive when compared with sport competition. Therefore, they should simulate aspects of the sport as closely as possible with a particular focus on temporary fatigue that occurs following bout of high intensity exercise (Knicker et al., 2011). This approach is supported by the fact that studies have found that biomechanical changes from fatigue protocols are evident after 50% of the protocol (Borotikar et al., 2008; Cortes et al., 2013) and short, intense protocols have similar effects when compared with long duration protocols (Lucci et al., 2011; Quammen et al., 2012). Therefore, the effect of high intensity fatiguing protocols that mimic aspects of the sport under investigation may affect the biomechanics of cutting and may therefore improve our understanding of the effects of fatigue on biomechanical risk factors for ACL injuries.

In conclusion, fatigue protocols may result in an increase in potential ACL loading as a result of altered trunk, hip and knee biomechanics, depending on the task being investigated. Therefore, the task being investigated should be ecologically valid. Also, the nature of the fatiguing protocol should closely mimic the demands of the sport being investigated. Our understanding of the effects of fatigue is limited due to differing fatiguing protocols and a dearth of investigations of hip and trunk biomechanics during cutting manoeuvres. In addition, there is a limited understanding of the combined effects of fatigue and decision making on neuromuscular and biomechanical risk factors for ACL injuries.

2.6 The Relationship between Anticipation and Biomechanical and Neuromuscular Risk Factors for Anterior Cruciate Ligament Injuries

Non-contact ACL injuries often occur during high risk sporting activities such as landing and cutting, particularly when responding to the sporting environment such as guarding an opponent, receiving or passing a ball (Boden et al., 2009; Walden et al., 2015). Performance of these activities in the unanticipated condition requires greater neurocognitive function to select and adjust the appropriate neuromuscular programme (Swanik, 2015). This section will demonstrate how performance of cutting activities in unanticipated compared with anticipated conditions alter trunk, hip and knee biomechanics leading to greater potential loading of the ACL.

The successful completion of sporting activities such as cutting is achieved through a combination of feed-forward and feedback control. Feedforward movement patterns are completed from an internal programme based upon experience of performing the activity. It is an anticipatory action with the neuromuscular pattern implemented in advance of the task to be completed (Kandel, 1999). The feed-forward motor programing system can be seen up to 500ms before sporting tasks (Bouisset and Do, 2008). Feedback motor control involves the modification of an ongoing motor task based upon feedback from the sensory receptors such as the eyes, the vestibular apparatus, joint and muscle receptors. Feedback control allows for the modification of technique and is necessary during unanticipated tasks

Table 2.6 The effect of fatigue on trunk kinematics

Study and participants	Activity	Fatiguing methodology	Effect of Fatigue	Magnitude of differences	Statistical value
Bell et al., (2016) 40 recreationally active (20 females and 20 males)	Drop jump	High intensity running protocol with numerous changes of direction, squatting and jumping exercises. Protocol ceased when 17 reported on RPE scale	**Kinematic findings** 1. Increase in LESS for males and females for A. Increased lateral trunk flexion at initial contact B. Decreased trunk flexion	 Pre-fatigue 13% reported *vs* 59% post-fatigue Pre-fatigue 18% reported *vs* 44% post-fatigue	 $P < 0.001$ $P = 0.002$
Lessi et al., (2017) 40 recreationally active (20 females and 20 males)	Single leg drop landing	Series of 10 bilateral squats, 2 bilateral vertical jumps, 20 step ups. Protocol ceased when unable to hop 80% of maximal hop distance	Increased peak trunk flexion (°) Increased contra lateral pelvic drop (°) Increased hamstring EMG activity (% MVIC) Increased gluteus maximus activity (% MVIC)	5.1° (95% CI = 2.7–7.4) 1.1° (95% CI = 0.5–1.6 3.5 (95% CI = 0.2–6.7) 4.4% (95%CI = 1.0–7.6)	$P < 0.001$ $P < 0.001$ $P = 0.037$ $P = 0.013$
Liederbach et al., (2014) 40 ballet dancers (20 female, 20 male) and 40 field sport athletes (20 female, 20 male)	Single leg drop landing	Series of 50 step ups, 15 maximal effort single leg jumps Protocol ceased when unable to jump 90% of maximal vertical jump height	Increased peak trunk flexion (°) Increased peak trunk side flexion (°)	Female athletes; pre -0.5 ± 1.2 *vs* post 0.4 ± 1.4 Male athletes; pre 0.6 ± 1.5 *vs* post 5.3 ± 2.2 Female athletes; pre 5.2 ± 0.7 *vs* post 6.7 ± 0.7 Male athletes; pre 5.9 ± 0.6 *vs* post 7.7 ± 1.0	$P = 0.002$ $P < 0.001$
Maas et al., (2017) 15 competitive athletes (5 female, 10 male), 15	10 second treadmill running trial	3200 metre run to exhaustion	Increased peak trunk flexion (°) Increased peak trunk rotation (°) Increased anterior pelvic tilt (°) Increased contra lateral pelvic drop	3.0 ± 4.2 (novice) 3.5 ± 4.7 (novice) 2.0 ± 2.1 (novice) 0.9 ± 0.6 (competitive) 1.6 ± 1.5 (competitive)	$P = 0.021$ $P = 0.034$ P = 0.01 $P < 0.001$ $P = 0.002$

novice athletes (6 female, 9 male)			(°) Increased pelvic rotation (°)	2.4 ± 0.3 (novice) 2.3 ± 1.6 (competitive)	P = 0.026 *P* < 0.001
McNeal et al., (2010) 20 athletes (9 female, 11 male)	Counter movement jump	Continuous counter movement jumps for 60 seconds	Increased peak trunk flexion	Values not reported	*P* < 0.001
Weeks et al., (2015) 60 moderately active adults (30 female, 30 male)	Single leg squat	Continuous lunging Protocol ceased when unable to continue lunging or jump 80% of maximal counter jump height	Increased peak trunk flexion (°) Increased peak trunk side flexion (°) Increased peak trunk rotation (°) Increased peak anterior pelvic tilt (°) Increased peak contra lateral pelvic drop (°) Increased peak pelvic rotation away (°)	24.5 ± 13.7 *vs* 29.8 ± 11.8 -7.0 ± 3.9 *vs* -3.3 ± 13.0 6.8 ± 5.7 *vs* 10.1 ± 7.9 30.4 ± 10.8 *vs* 31.8 ± 8.7 -5.2 ± 3.3 *vs* -1.9 ± 6.9 -4.0 ± 3.0 *vs* -5.4 ± 5.0	*P* = 0.001 *P* = 0.030 *P* = 0.001 *P* = 0.050 *P* = 0.001 *P* = 0.040
Wild et al., (2017) 14 female Irish dancers	Jump (dance specific)	Dance specific fatigue protocol Ceased when participants reported 17 on Borg RPE scale	No increase in trunk flexion No increase in trunk side flexion		*P* = 0.257 *P* = 0.081

MVIC = Maximal voluntary isometric contract, RPE = Rate of perceived exertion

(Seidler et al., 2004). The interaction between feed-forward and feedback control is particularly important in sports that require participants to formulate neuromuscular strategies in response to unanticipated external stimuli. However, this interaction requires a certain amount of time to account for the delay in feedback loops and this delay may be a factor in sustaining ACL injuries. It has been calculated that the pathological loading of the ACL occurs in under 40 milliseconds during single limb loading (Koga et al., 2010) which is shorter than the minimum of 333ms required for feedback motor programming (McLean and Samorezov, 2009). This temporal constraint in the feedback control system may limit the ability of neurocognitive faculties to process the somatosensory information, modify the ongoing motor task (Swanik, 2015) and avoid excessive loading of tissues, such as the ACL.

Three stages of information processing are required when modifying neuromuscular programmes in response to unanticipated stimuli: stimulus identification; response selection; and response programming (Schmidt, 2008). During the first stage, athletes identify the stimulus through the visual, vestibular and/or proprioceptive systems. During the response selection phase, athletes must integrate the information regarding the unanticipated stimulus with the existing, pre-programmed neuromuscular programme and related somatosensory feedback. In the final response programming phase, athletes must then retrieve the motor programme for task to be completed and prepare for its implementation (Brown et al., 2014). In addition to this system of motor control, rapid, reflex motor corrections based in the brainstem and spinal cord can occur with little or no conscious control. These reflexive modulations involve the M1 and M2 responses, triggered reactions and voluntary reaction time (M3) response (Schmidt, 2008). The M1 response (also known as the monosynaptic, stretch reflex) occurs where there is a sudden stretch of the muscle spindles resulting a in a muscle contraction with a latency of 30 – 50 milliseconds. Numerous M1 responses can occur simultaneously and contribute to postural corrections. The M2 response is also triggered by muscle spindle activity. Unlike the M1 response, signals from the M2 response continue past the spinal cord and up to the motor cortex and/or cerebellum. This explains the longer latency period (50 – 80 milliseconds) and how this reflex can be modulated (increased or decreased) by conscious processes. The triggered reaction has a latency of 80 - 120 milliseconds, generates coordinated muscle contractions away from the initial sensory receptors (e.g. muscle spindles). Although this is a reflex response, it can become a learned response. Finally, the M3 (voluntary reaction) response has a latency of 120 180 milliseconds. It can affect all the muscles of the body, not just those being stretched. It can be quite a powerful and sustained response. Although it is classified as a reflex response, it requires the person's attention and can be modified by anticipation, instruction and sensory feedback (Schmidt, 2008).

Although many reflexive and peripheral pathways contribute to the selection of a neuromuscular programme in response to a stimulus, the neurocognitive function of the cerebral cortex is ultimately responsible for their selection and modification (Swanik, 2015). A clear association between neurocognitive function and noncontact ACL injury has been demonstrated (see section 2.11 The Effect of Fatigue on Neurocognitive Function). Although the effect of unanticipation on knee biomechanics during high risk activities such as side cutting manoeuvres has been investigated (Brown et al., 2014), there is limited understanding of its effect on trunk and pelvic biomechanics during side cutting and its effect on crossover cutting in general (Kim et al., 2014). Furthermore, there is limited understanding of the interplay between fatigue and unanticipation on biomechanics of high risk activities.

2.6.1 The Effects of Anticipation on Biomechanical and Neuromuscular Risk Factors for Anterior Cruciate Ligament Injuries during Side Cutting

Using musculoskeletal modelling based on 20 athletes Weinhandl et al. (Weinhandl et al., 2013) determined that ACL loading was significantly increased by 13% ($p = 0.02$) when completing side cuts in the unanticipated condition compared with the anticipated condition. They found that sagittal plane loading was the largest contributor to ACL loading in both the anticipated (62%) and unanticipated (67%) conditions and it was responsible for the increased ACL loading in the unanticipated condition. Frontal plane loading accounted for 26% and 24% of ACL loading during anticipated and unanticipated side cutting respectively whereas transverse plane loading accounted for 12% and 9% respectively. The reason for the increase in overall ACL loading is likely due to the adoption of different biomechanics during unanticipated side cutting, namely increased knee extensor moments (Table 2.7). It was generally found that performance of the side cut in the unanticipated condition generally leads to greater hip (Kim et al., 2014; McLean and Samorezov, 2009; Weinhandl et al., 2013) and knee flexion (Besier et al., 2001; Collins et al., 2016; Cortes et al., 2011; Dempsey et al., 2009; Donnelly et al., 2012; Khalid et al., 2015; Kim et al., 2014; McLean and Samorezov, 2009) and ankle dorsiflexion (Weinhandl et al., 2013) compared with the anticipated condition. However, some studies found that there was no effect on knee sagittal plane kinematics (Borotikar et al., 2008; Cochrane et al., 2010; Lee et al., 2013; Weinhandl et al., 2013), or even a decrease in hip (Borotikar et al., 2008; McLean and Samorezov, 2009) and knee (Meinerz et al., 2015) flexion. The conflicting results in hip and knee kinematics may be due to the different cutting techniques analysed. Most studies that found an increased flexed posture during unanticipation used a run and cut at 45° whereas others that did not find this posture used a jump forwards followed by a lateral jump technique (Borotikar et al., 2008; McLean and Samorezov, 2009)[3]. Additionally, Meinerz et al. (Meinerz et al., 2015) positioned participants on a step prior to a jump forwards, landing and side cutting which may account for the differences in knee flexion found. Conflicting results may also be contributed to by the analysis of variables at different pre-selected time points such as initial contact, during the weight acceptance phase (Besier et al., 2001; Dempsey et al., 2009; Donnelly et al., 2012; Khalid et al., 2015; Lee et al., 2013), at peak ACL loading (Weinhandl et al., 2013), during the first 50% of stance (Cortes et al., 2011) or during 100% of stance (Borotikar et al., 2008; Kim et al., 2014; McLean and Samorezov, 2009).

The flexed lower limb posture was accompanied by greater hip extensor moment (Kim et al., 2014) and gluteus maximus activity (Meinerz et al., 2015), greater knee extensor

[3] While it could be argued that this is not cutting, these studies are generally considered as cutting activities in the literature and will considered as such in this thesis.

(Cortes et al., 2011; Kim et al., 2014; McLean and Samorezov, 2009; Meinerz et al., 2015) and ankle plantarflexor (Kim et al., 2014) moments. This may be a strategy to provide an athlete with more time to select and implement the appropriate neuromuscular strategy. This is supported by the fact that smaller peak posterior (Khalid et al., 2015) and vertical GRFs (Khalid et al., 2015; Kim et al., 2014) are generally evident during unanticipated, compared with anticipated, side cutting. In contrast, Meinerz et al. (Meinerz et al., 2015) found an increase in vGRFs during unanticipated side cutting which may have been contributed to by landing and cutting from a height. Despite these general decreases in GRFs, Weinhandl et al. (Weinhandl et al., 2013) estimated that maximal peak ACL loading occurred in the first 30ms following initial contact, which is similar to the timing of ACL injuries (Koga et al., 2010; Krosshaug et al., 2007). They found that this was due to increased quadriceps activity which results in increased tibiofemoral compressive and anterior shear force due to the inclination of the patellar tendon and the posterior tibial slope. Therefore, although performance of side cutting in the unanticipated condition results in smaller GRFs and increased hip and knee flexion, which may at least partially offset an increase in ACL loading, the greater resultant knee extensor moment leads to a significant increase in the estimated ACL loading in the critical weight acceptance phase of side cutting.

Our understanding of the sagittal plane effects of unanticipation are restricted by the fact that trunk and pelvic kinematics have not been investigated. In addition, the effects of unanticipation have been investigated on different discrete points at predetermined time points rather than phases of cutting. Furthermore, if an athlete’s ability to employ a flexed hip and knee posture during unanticipated side cutting is restricted due to fatigue for example, it suggests that a combination of fatigue and unanticipation may lead to greater ACL loading and warrants further investigation.

In the frontal plane, performing unanticipated compared to anticipated side cuts resulted in greater trunk side flexion away from the direction of cut (Dempsey et al., 2009; Lee et al., 2013) and increased hip abduction (Kim et al., 2014; Lee et al., 2013; Weinhandl et al., 2013). This posture is due to the combination of lateral foot positioning and a trunk roll technique (combination of side flexion away from direction of cut, trunk flexion and trunk rotation in the direction of cut) when there is restricted time to plan the task (Mornieux et al., 2014; Patla et al., 1999). Trunk side flexion has been demonstrated to be associated with greater knee adductor moment (Dempsey et al., 2007; Jamison et al., 2012; Mornieux et al., 2014), which in turn can increase subsequent ACL loading and risk of injury (Hewett et al., 2005). The trunk side flexed and hip abducted position may, at least partially, necessitate the greater hip adductor moment (Kim et al., 2014), increased knee abduction angle (Collins et al., 2016; Cortes et al., 2011; McLean and Samorezov, 2009) and knee adductor moment (Besier et al., 2001; Collins et al., 2016; Lee et al., 2013; McLean and Samorezov, 2009) generally observed in unanticipated compared with anticipated side cutting. These altered frontal plane biomechanics may increase ACL loading (Oh, Lipps et al., 2012; Shin et al., 2011) and risk of injury (Hewett et al., 2005) particularly if coupled with increased sagittal and transverse plane loading.

Unanticipation has been found to have a number of effects in the transverse plane along the kinetic chain during side cutting. Despite the fact that reduced trunk rotation in the direction of travel during side cutting is associated with increased knee external rotator moments (Dempsey et al., 2007; Frank et al., 2013; Jamison et al., 2012), only one study examined the effect on transverse plane trunk kinematics and found there to be no effect (Dempsey et al., 2009). In relation to the hip joint, unanticipation was generally found to increase internal hip rotation angle (Borotikar et al., 2008; McLean and Samorezov, 2009; Weinhandl et al., 2013) and moment (Kim et al., 2014; McLean and Samorezov, 2009), knee internal rotation angle (Collins et al., 2016; Cortes et al., 2011; Kim et al., 2014; McLean and Samorezov, 2009) and knee external rotator moment (Besier et al., 2001; Kim et al., 2014; McLean and Samorezov, 2009) which can all act to increase ACL loading and potential for injury. Again, in contrast to this, Meinerz et al. (Meinerz et al., 2015) found that anticipation resulted in a decrease in knee external rotator moment which could be due to the different cutting technique. An increase in knee external rotator moment, as found in three of the studies (Besier et al., 2001; Kim et al., 2014; McLean and Samorezov, 2009), in combination with greater adductor and extensor moments has been found to create the greatest ACL loading (Oh, Kreinbrink et al., 2012; Oh et al., 2012). This demonstrates the potentially negative effect of completing side cuts in the unanticipated condition.

In conclusion, performance of side cutting in the unanticipated condition, compared with the anticipated condition, results in greater ACL loading due to a combination of greater sagittal, frontal and transverse plane loading. This results from alterations in trunk kinematics, and hip and knee biomechanics. Given the association between reduced hip internal rotation range of motion and ACL injury, (Bedi et al., 2016; Ellera Gomes et al., 2014; Gomes et al., 2008; VandenBerg et al., 2017), a thorough understanding of hip joint biomechanics during unanticipated side cutting is of paramount importance in order to develop effective ACL IPPs. Adequate hip control seems to be particularly important in the unanticipated condition as net hip energy absorption is significantly higher than in the anticipated condition (Meinerz et al., 2015). Our understanding of the effects of unanticipation is limited by the dearth of research investigating the effect on trunk kinematics, particularly in the frontal and transverse planes. Furthermore, as unanticipation and fatigue individually lead to altered biomechanics which may increase ACL loading, an understanding of the combined effects of fatigue and unanticipation on the biomechanics of the kinetic chain will enhance our understanding of biomechanical risk factors for ACL injuries and facilitate the development of ACL IPPs.

Table 2.7 The effects of unanticipation on biomechanical and neuromuscular risk factors for anterior cruciate ligament injuries during side cutting

Study and participants	Activity	Effect of Unanticipation	Magnitude of differences	Statistical value
Kim et al., (2014) 37 male, adolescent soccer players	Light indicating either a 45° side cutting or 45° crossover cutting manoeuvre was triggered at 90% stride distance before the centre of the force plate. Approach speed was 3.5 ms^{-1}. Biomechanical variables examined over 100% of stance phase.	**Kinematic findings**		
		1. Increased peak knee flexion angles (°)	ANT 45.7 ± 7.5 *vs* UNA 57.8 ± 7.6	$P < 0.001$
		2. Increased knee varus angles (°)	ANT 0.7 ± 6.8 *vs* UNA -0.7 ± 9.6	$P = 0.011$
		3. Increased peak knee internal rotation in side cutting (°)	ANT 10.9 ± 11.1 *vs* UNA 13.6 ± 10.6	$P = 0.011$
		4. Increased peak hip flexion (°)	ANT 39.7 ± 8.0 *vs* UNA 48.4 ± 7.8	$P < 0.001$
		5. Increased peak hip abduction (°)	ANT 17.7 ± 6.1 *vs* UNA 23.1 ± 5.8	$P < 0.001$
		Kinetic findings		
		6. Greater peak knee extensor moment ($Nm.kg^{-1}$)	ANT 2.41 ± 2.54 *vs* UNA 5.33 ± 2.81	$P < 0.001$
		7. Greater peak knee adductor moment ($Nm.kg^{-1}$)	ANT 0.10 ± 1.00 *vs* UNA 1.44 ± 1.16	$P < 0.001$
		8. Smaller peak knee internal rotator moment ($Nm.kg^{-1}$)	ANT 1.03 ± 0.25 *vs* UNA 0.84 ± 0.31	$P < 0.001$
		9. Greater peak hip extensor moment ($Nm.kg^{-1}$)	ANT 4.18 ± 6.74 *vs* UNA 10.48 ± 7.26	$P < 0.001$
		10. Greater peak hip adductor moment ($Nm.kg^{-1}$)	ANT 1.12 ± 2.14 *vs* UNA 4.26 ± 3.24	$P < 0.001$
		11. Greater peak hip internal rotator moment ($Nm.kg^{-1}$)	ANT - 0.23 ± 0.50 *vs* UNA 0.07 ± 0.49	$P = 0.008$
		12. Greater peak ankle plantar flexor moment ($Nm.kg^{-1}$)	ANT 3.37 ± 0.96 *vs* UNA 3.86 ± 1.00	$P = 0.016$
		13. Smaller peak ankle inversion moment ($Nm.kg^{-1}$)	ANT 1.48 ± 0.78 *vs* UNA 0.95 ± 0.90	$P = 0.001$
		14. Longer time to peak mediolateral GRF (s)	ANT 0.55 ± 0.09 *vs* UNA 0.60 ± 0.09	$P < 0.001$
		15. Smaller peak mediolateral GRF (%BW)	ANT 0.80 ± 0.13 *vs* UNA 0.58 ± 0.16	$P < 0.001$
		16. Smaller peak vertical GRF(%BW)	ANT 2.76 ± 0.39 *vs* UNA 2.32 ± 0.32	$P < 0.001$
		EMG findings		
		17. Lower vastus lateralis EMG activity between 50-60% of stance	UNA - magnitude not provided	$P < 0.001$
		18. Lower vastus medialis EMG activity between 40-50% of stance	UNA - magnitude not provided	P <0.01 - = 0.04

McLean and Samorezov (2009) 20 female collegiate athletes	Single leg side jump/cutting manoeuvre indicated by light 350ms prior to ground contact. Biomechanical variables examined over 100% of stance phase.	**Kinematic findings*** 1. Decreased hip flexion at initial contact (°) 2. Increased hip internal rotation at initial contact(°) 3. Increased peak hip internal rotation (°) 4. Increased peak knee flexion angles (°) 5. Decreased peak knee abduction angles (°) 6. Increased peak knee internal rotation angles (°) **Kinetic findings*** 7. Greater knee extensor moment at initial contact (Nm) 8. Greater peak hip internal rotator moment (Nm) 9. Greater peak knee adductor moment (Nm) 10. Greater peak knee adductor moment (Nm) * Values for non-fatigued limb reported	ANT: 30.6 ± 7.2 *vs* UNA: 28.2 ± 6.3 ANT: 4.6 ± 3.2 *vs* UNA: 6.6 ± 2.7 ANT: 8.8 ± 4.7 *vs* UNA: 10.8 ± 4.7 ANT: 57.9 ± 8.6 *vs* UNA: 55.3 ± 7.6 ANT: 5.1 ± 3.6 *vs* UNA: 3.3 ± 2.8 ANT: 13.1 ± 6.1 *vs* UNA: 14.6 ± 6.5 ANT: 16.5 ± 3.4 *vs* UNA: 16.9 ± 2.0 ANT: 8.0 ± 3.5 *vs* UNA: 10.3 ± 4.7 ANT: 3.8 ± 2.9 *vs* UNA: 3.9 ± 2.6 ANT: 13.2 ± 5.4 *vs* UNA: 14.7 ± 5.7	$P < 0.01$ $P < 0.01$ $P < 0.01$ $P < 0.01$ $P < 0.01$ $P < 0.01$ $P < 0.01$ $P < 0.01$ $P < 0.01$ $P < 0.01$
Khalid et al., (2015) 12 collegiate soccer players (6 females and 6 males)	Anticipated and unanticipated side cutting manoeuvres at 45° during a run of 4.5 – 5.5 ms^{-1} with an average reaction time of 290 ms for the unanticipated condition. Biomechanical variables examined during weight acceptance and peak push off phases.	**Kinematic findings** 1. Increased knee flexion angle (°) at initial contact **Kinetic findings** 2. Smaller peak knee extensor moment ($Nmkgm^{-1}$) 3. Smaller peak knee valgus moment ($Nm.kg.m^{-1}$) 4. Greater peak knee varus moment ($Nm.kg.m^{-1}$) 5. Greater peak knee external rotator moment ($Nm.kg^{-1}.m^{-1}$) 6. Smaller peak vertical GRF ($N.kg^{-1}$) 7. Smaller peak posterior GRF ($N.kg^{-1}$)	ANT: 27.09 ± 9.51 *vs* UNA: 32.38 ± 11.98 ANT: 1.74 ± 0.38 *vs* UNA: 1.49 ± 0.37 ANT: 0.37 ± 0.17 *vs* UNA; 0.22 ± 0.12 ANT: 0.41 ± 0.14 *vs* UNA: 0.52 ± 0.22 ANT: 0.15 ± 0.09 *vs* UNA: 0.25 ± 0.12 ANT: 2.35 ± 0.66 *vs* UNA: 2.06 ± 0.50 ANT: 0.84 ± 0.28 *vs* UNA: 0.73 ± 0.25	$P < 0.001$ $P < 0.001$ $P = 0.019$ $P = 0.009$ $P < 0.001$ $P < 0.001$ $P < 0.001$
Borotikar et al., (2008) 25 female collegiate athletes	Single leg side jump/cutting manoeuvre indicated by light 350ms prior to ground contact	**Kinematic findings*** 1. Decreased hip flexion at initial contact (°) 2. Increased hip internal rotation at initial contact(°) 3. Increased peak knee abduction angles (°) 4. Greater hip internal rotation angles (°) **Kinetic findings**	ANT: 31.3 ± 3.0 *vs* UNA: 27.2 ± 2.4 ANT: 8.2 ± 2.1 *vs* UNA: 10.2 ± 1.5 ANT: 3.5 ± 3.2 *vs* UNA: 3.9 ± 2.8 ANT: 13.1 ± 4.2 *vs* UNA: 15.2 ± 5.0 Not investigated	$P < 0.001$ $P < 0.001$ $P < 0.001$ $P < 0.001$

		* Kinematic values for dominant limb reported		
Cortes et al., (2011) 13 female soccer players	Anticipated and unanticipated side cuts at 45° following a run at 4.4 ms^{-1} (anticipated) and 3.7 ms^{-1} unanticipated ($p < 0.001$). Stop jump data not analysed. Biomechanical variables examined over 50% of stance phase.	**Kinematic findings***		
		1. Increased knee flexion at initial contact (°)	ANT: 15.4 ± 4.5 *vs* UNA: 20.7 ± 4.7	$P < 0.001$
		2. Increased knee abduction at initial contact (°)	ANT: 0.8 ± 3.9 *vs* UNA: 1.5 ± 3.9	$P = 0.039$
		3. Increased knee internal rotation initial contact (°)	ANT: 5.2 ± 6.5 *vs* UNA: 8.1 ± 4.7	$P = 0.031$
		4. Increased hip adduction at initial contact (°)	ANT: 12.7 ± 4.8 *vs* UNA: 8.8 ± 7.6	$P = 0.015$
		5. Increased peak knee flexion angles (°)	ANT: 45.2 ± 4.5 *vs* UNA: 52.4 ± 5.6	$P < 0.001$
		Kinetic findings		
		6. Greater knee extensor moment at initial contact ($Nm.kg^{-1}.m^{-1}$)	ANT: 0.16 ± 0.14 *vs* UNA: 0.014 ± 0.11	$P = 0.003$
		7. Smaller peak knee adductor moment ($Nmkg^{-1}m^{-1}$)	ANT: 0.52 ± 0.40 *vs* UNA: 0.37 ± 0.36	$P = 0.035$
Demspey et al., (2009) 9 male, non-elite team sport players	Anticipated and unanticipated side cutting at 45° following an approach run of 5.7 ms^{-1} (anticipated) and 5.1 ms^{-1} (unanticipated) ($p < 0.05$). Visual trigger provided 400ms prior to foot contact. Biomechanical variables examined during weight acceptance phase.	**Kinematic variables**		
		1. Increased mean knee flexion during weight acceptance phase (°)	ANT: 29.7 ± 4.8 *vs* UNA: 31.2 ± 2.8	$P = 0.038$
		2. Increased trunk side flexion away from direction of cut (°)	ANT: 7.4 ± 3.2 *vs* UNA: 12.2 ± 4.9	$P = 0.003$
		Kinetic variables		
		No significant differences for knee moments		
Weinhandl et al., (2013) 20 healthy female recreational athletes	Anticipated and unanticipated side cutting at 45° following an approach run at 4.5-5.0 ms^{-1}. Visual stimulus was presented 600ms prior to initial contact. Three options of run, side cut and stop. Biomechanical variables reported during weight acceptance phase.	**Kinematic variables at peak ACL loading**		
		1. Increased hip flexion (°)	ANT: 36.3 ± 9.0 *vs* UNA: 38.6 ± 8.0	$P = 0.006$
		2. Increased hip abduction (°)	ANT: 6.3 ± 6.6 *vs* UNA: 11.0 ± 6.8	$P < 0.001$
		3. Increased hip internal rotation (°)	ANT: 6.6 ± 7.6 *vs* UNA: 8.2 ± 7.0	$P = 0.016$
		4. Increased ankle dorsiflexion(°)	ANT: 3.5 ± 4.4 *vs* UNA: 5.3 ± 3.0	$P = 0.015$
		Kinematic variables at peak ACL loading		
		5. Smaller hip adductor moment ($Nm.kg^{-1}$)	ANT: 0.88 ± 1.12 *vs* UNA: 0.36 ± 1.06	$P = 0.023$
		6. Greater peak ACL loading ($N.kg^{-1}$)	ANT: 11.02 ± 4.65 *vs* UNA: 12.40 ± 3.79	$P = 0.022$
		7. Greater sagittal planar peak ACL loading(Nkg^{-1})	ANT: 6.79 ± 3.43 *vs* UNA: 8.27 ± 2.68	$P = 0.007$

Besier et al., (2001) 11 male subjects	Run at 3 ms^{-1} and 4 possibilities: straight run, crossover cut at 60°, side cut at 30° and side cut at 60°. Biomechanical variables reported during weight acceptance phase.	**Kinematic findings** 1. Increased knee flexion during weight acceptance phase (°) Knee transverse and frontal plane kinematics not investigated **Kinetic findings** 2. Knee extensor moment 3. Greater knee adductor moment 4. Greater external rotator moment	30: ANT 31.9 *vs* UNA 35.2 60: ANT 32.3 *vs* UNA 34.3 50% greater during UNA side cut at 60° in weight acceptance phase 129% greater during UNA side cut at 30° in weight acceptance phase 49% greater during UNA side cut at 60° in weight acceptance phase	$P = 0.005$ $P < 0.001$ $P < 0.001$ $P < 0.001$ $P < 0.001$
Cochrane et al., (2010) 50 healthy, male subjects	Run at 4.0- 4.5 ms^{-1} and 4 possibilities: straight run, crossover cut at 60°, side cut at 30° and side cut at 60°. Biomechanical variables examined during weight acceptance phase.	**Knee kinematic variables** **Knee kinetic variables**	Not significant Not significant	
Donnelly et al., (2012) 34 male, Australian rules footballers	Anticipated and unanticipated side cutting at 45°. Biomechanical variables examined during weight acceptance phase.	**Kinematic Variables** 1. Increased knee flexion range of motion during the weight acceptance phase **Knee kinetic variables**	ANT: 33.0 ± 6.2 *vs* UNA: 35.3 ± 6.4 no difference at baseline	$P < 0.01$
Lee et al., (2013) 15 high level male soccer players and 15 low level, male soccer players	Anticipated and unanticipated side cutting to 45° with visual cueing 453ms prior to initial contact. Biomechanical variables examined during weight acceptance phase.	**Kinematic Variables** 1. Increased trunk side flexion away from cut direction during weight acceptance phase (°) 2. Decreased hip flexion during weight acceptance phase (°) 3. Increased hip abduction during weight	ANT 6.5 ± 6.0 *vs* UNA 10.3 ± 4.0 ANT 47.5 ± 7.3 *vs* UNA 41.4 ± 6.6 ANT 11.2 ± 5.9 *vs* UNA 20.2 ± 5.0	$P < 0.01$ $P < 0.001$ $P < 0.01$

		acceptance phase (°)		
		Kinetic Variables		
		4. Greater peak knee adductor moment during weight acceptance phase ($N.kg^{-1}$)	ANT 0.39 ± 0.27 *vs* UNA 0.66 ± 0.30	$P < 0.01$
		5. Smaller peak knee external rotator moment	ANT 0.21 ± 0.10 *vs* UNA 0.15 ± 0.11	$P < 0.01$
Meinerz et al., (2015) 18 female collegiate soccer players	Anticipated and unanticipated side cutting to 45° with visual cueing 453ms prior to initial contact. Biomechanical variables examined at initial contact and at peak values during stance.	**Kinematic Variables**		
		1. Less knee flexion at initial contact (°)	ANT 19.6 ± 4.1 *vs* UNA 17.4 ± 4.1	$P = 0.001$
		Kinetic Variables		
		2. Greater knee extensor moment (Nm)	ANT 163.0 ± 25.0 *vs* UNA 155.6±22.8	$P = 0.041$
		3. Greater knee external rotator moment (Nm)	ANT 9.8 ± 4.6 *vs* UNA 7.0 ± 3.9	$P = 0.015$
		4. Greater hip abductor moment (Nm)	ANT 57.5 ± 17.7 *vs* UNA 72.1 ± 18.5	$P = 0.028$
		5. Greater hip external rotator moment (Nm)	ANT 49.0 ± 12.4 *vs* UNA 59.9 ± 11.7	$P = 0.001$
		6. Greater ankle invertor moment (Nm)	ANT 12.7 ± 6.0 *vs* UNA 19.0 ± 7.9	$P = 0.002$
		7. Greater net hip mechanical absorption in the landing phase ($W.kg^{-1}$)	ANT -0.84 ± 0.39 *vs* UNA -1.06 ± 0.49	$P = 0.001$
		8. Smaller net ankle mechanical absorption in the landing phase ($W.kg^{-1}$)	ANT –0.77 ± 0.31 *vs* UNA -0.62 ± 0.27	$P = 0.001$
		9. Greater gluteus maximus EMG activity in pre landing phase (%MVIC)	ANT 34 ± 19 *vs* UNA 49 ± 27	$P = 0.005$
		10. Greater gluteus maximus EMG activity in pre landing phase (%MVIC)	ANT 34 ± 19 *vs* UNA 49 ± 27	$P = 0.016$
Collins et al., (2016) 13 female, collegiate, soccer players	15 m at 4.5 – 5.0 ms^{-1} with possibilities of side cut, stop and straight run. Anticipated and unanticipated side cuts analysed. Variables normalised to 100% of stance. Approximately 600ms notification for unanticipated side cut	**Kinematic Variables**		
		2. Increase in peak knee flexion angles (°)	ANT 49.3 ± 7.2 *vs* UNA 53.1 ± 7.4	$P = 0.001$
		3. Increase in peak knee abduction angles (°)	ANT 2.2 ± 4.7 *vs* UNA 2.8 ± 4.4	$P = 0.030$
		4. Increase in peak knee internal rotation angles (°)	ANT 10.6 ± 5.0 *vs* UNA 11.2 ± 4.9	$P = 0.020$
		Kinetic Variables		
		5. Greater peak knee adductor moments ($Nm.kg^{-1}$)	ANT 0.3 ± 0.2 *vs* UNA 0.4 ± 0.2	$P = 0.016$

ANT = Anticipated, UNA = Unanticipated, GRF = Ground reaction force, EMG = Electromyography

Table 2.8 A summary of the biomechanical effects of anticipation on side cutting manoeuvres

Variable	Sagittal	Frontal	Transverse
Trunk Kinematics		↑ Ipsilateral flexion	
Hip Kinematics	↑ Flexion	↑ Abduction	↑ Internal rotation
Hip Moments	↑ Extensor	↑ Adductor	↑ External rotator
Knee Kinematics	↑ Flexion		↑ Internal rotation
Knee Moments	↑ Extensor		↑ External rotator
Ankle Kinematics	↑ Dorsiflexion		
Ankle Moments	↑ Plantarflexor		

Internal joint moments reported

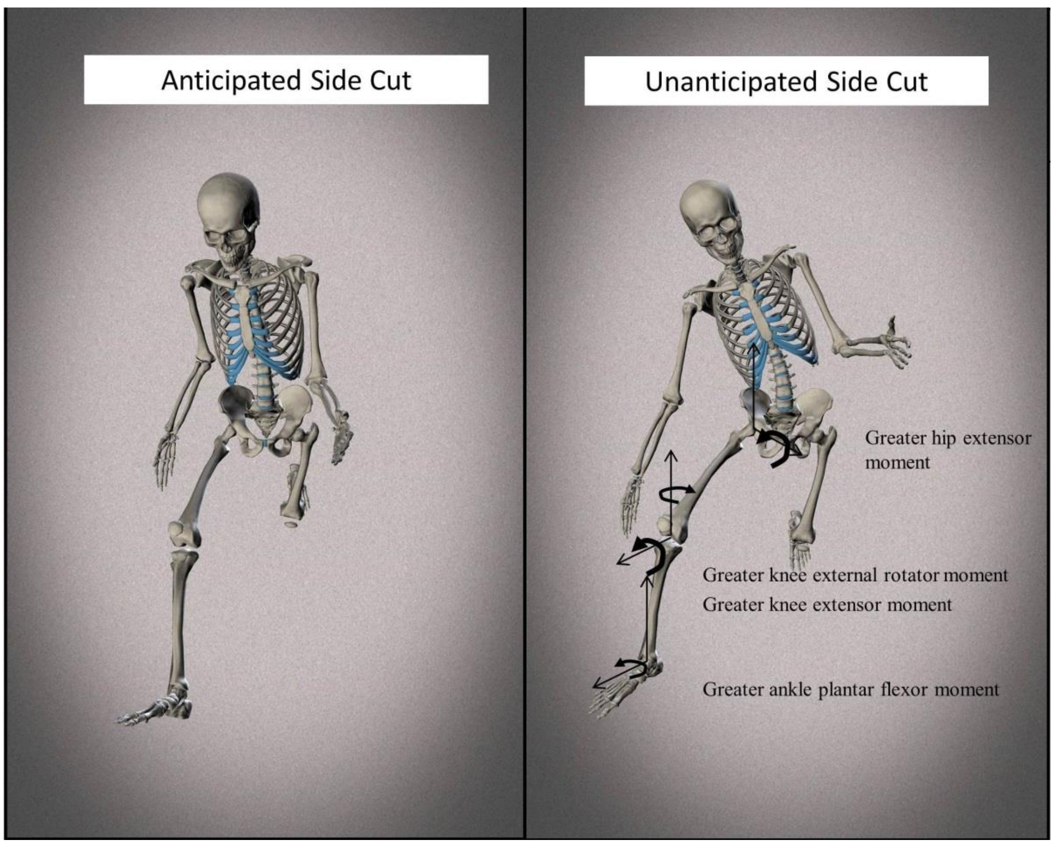

Figure 2.9 The biomechanical effects of anticipation on side cutting manoeuvres

2.6.2 The Effects of Anticipation on Biomechanical and Neuromuscular Risk Factors for Anterior Cruciate Ligament Injuries during Crossover Cutting

Despite the fact that crossover cutting is regularly used to change direction in field sports (Andrews et al., 1977; Potter et al., 2014), the effect of unanticipation on crossover cutting has not been investigated as frequently as on side cutting (Cortes et al., 2014; Whyte et al., 2018). Only three studies (Table 2.9) have investigated the effects of unanticipation on the biomechanics of crossover cutting. Therefore, the suggestion that crossover cutting is a safer change of direction technique than side cutting (McGovern et al., 2015; Potter et al., 2014), as non-contact ACL injuries occur less frequently during crossover cutting (Cochrane et al., 2007), may be premature. The lower observed incidence of ACL injuries during crossover cutting may be due to a lower frequency of crossover cutting manoeuvers used as a change of direction technique compared with side cutting (Potter et al., 2014). The three studies that have investigated the effects of anticipation on crossover cutting have provided some contradictory findings (Besier et al., 2001; Cochrane et al., 2010; Kim et al., 2014). Although Cochrane et al. (Cochrane et al., 2010) did not find any knee kinematic or kinetic effect, other studies found that unanticipation resulted in increased hip (Kim et al., 2014) and knee (Besier et al., 2001; Kim et al., 2014) flexion and ankle dorsiflexion (Kim et al., 2014), which is a similar flexed posture observed during unanticipated side cutting. Kim et al. (Kim et al., 2014) also found that unanticipation altered the pattern of joint loading with a greater knee extensor moment, contributed to by an increase in vastus lateralis activity, and a decrease in hip extensor and ankle plantar flexor moments in the early stance phases of crossover cutting. These findings, in conjunction with the observations that there were longer times to peak mediolateral and vertical GRF and smaller peak values, suggest that deceleration during unanticipated crossover cutting was achieved mainly by the quadriceps activity. These alterations, in combination with an increase in gastrocnemius activity (Kim et al., 2014) during the critical weight acceptance phase of crossover cutting, may increase ACL loading (Fleming et al., 2001). Therefore, the increased knee extensor loading and decreased ankle and hip loading suggest that athletes adopt a knee dominant strategy in the sagittal plane during unanticipated crossover cutting which may increase ACL loading and subsequent injury risk. However, our overall understanding is limited as no studies have investigated the effect on trunk kinematics.

In the frontal plane, unanticipation led to increased knee adduction and ankle inversion angles (Kim et al., 2014) and a greater knee valgus moment (Besier et al., 2001). These alterations in frontal plane knee biomechanics may result in greater ACL loading (Oh et al., 2012) particularly if it is accompanied by increased transverse plane loading. Unanticipation has been found to increase knee adduction angle (Kim et al., 2014), however only one study has found that it lead to an increase in knee abductor moment (Besier et al., 2001). In contrast, Kim et al. (Kim et al., 2014) found a decrease in abductor moments while Cochrane et al. (Cochrane et al., 2010) did not find any effect. The contradictory results may be due to methodological differences as Besier et al., (Besier et al., 2001) analysed peak findings for the weight acceptance phase whereas Kim et al., (Kim et al., 2014) reported peak values during the entire stance phase. Non-contact ACL injuries occur during the weight acceptance phase of cutting (Krosshaug et al., 2007) may provide a clearer understanding of potential relationships with injury. Cochrane et al. (Cochrane et al., 2010) attributed their non-significant findings to potential learning effects or an increase in running speed.

In conclusion, while studies investigating the effects of unanticipation on biomechanical and neuromuscular risk factors for ACL injuries suggest that there is potential for increased sagittal and frontal plane loading of the ACL, they are limited in numbers. In addition, research is inconclusive on the effects in the transverse plane. Furthermore, no studies have investigated the effects of unanticipation on trunk kinematics with only one study having examined the effect on hip biomechanics. Also, an understanding of the combined effects of anticipation and fatigue on the biomechanical and neuromuscular risk factors for ACL injuries may provide additional information for the development of ACL IPPs.

Table 2.9 The effects of anticipation on biomechanical and neuromuscular for anterior cruciate ligament injuries during crossover cutting

Study and participants	Activity	Effect of Unanticipation	Magnitude of differences	Statistical value
Kim et al., (2014) 37 male, adolescent soccer players	Light indicating either a 45° side cutting or 45° crossover cutting manoeuvre was triggered at 90% stride distance before the centre of the force plate. Approach speed was 3.5 ms^{-1}	**Kinematic findings**		
		1. Increased peak knee flexion angles (°)	ANT 46.1 ± 10.9 *vs* UNA 52.6 ± 8.9	$P < 0.001$
		2. Increased knee adduction angles (°)	ANT 0.6 ± 7.7 *vs* UNA 2.6 ± 8.6	$P = 0.011$
		3. Increased peak knee external rotation in crossover cutting (°)	ANT -16.9 ± 11.0 *vs* UNA -17.9 ± 10.4	$P < 0.001$
		4. Increased peak hip flexion (°)	ANT 40.5 ± 8.2 *vs* UNA 46.3 ± 9.1	$P < 0.001$
		5. Increased peak ankle dorsiflexion (°)	ANT 17.7 ± 7.8 *vs* UNA 20.6 ± 6.9	$P = 0.027$
		6. Increased peak ankle inversion (°)	ANT 5.9 ± 3.7 *vs* UNA 6.4 ± 3.6	$P < 0.001$
		Kinetic findings		
		7. Greater peak knee extensor moment ($Nm.kg^{-1}$)	ANT 2.10 ± 1.33 *vs* UNA 2.42 ± 1.87	$P < 0.001$
		8. Smaller peak knee internal rotator moment ($Nm.kg^{-1}$)	ANT 0.61 ± 0.35 *vs* UNA 0.50 ± 0.36	$P < 0.001$
		9. Smaller peak hip extensor moment ($Nm.kg^{-1}$)	ANT 2.85 ± 3.22 *vs* UNA 1.31 ± 0.39	$P = 0.009$
		10. Greater peak hip abductor moment ($Nm.kg^{-1}$)	ANT 3.44 ± 0.78 *vs* UNA 4.45 ± 1.95	$P = 0.001$
		11. Smaller peak ankle plantar flexor moment ($Nm.kg^{-1}$)	ANT 3.17 ± 1.20 *vs* UNA 0.53 ± 0.21	$P < 0.001$
		12. Longer time to peak mediolateral GRF (s)	ANT 0.55 ± 0.10 *vs* UNA 0.61 ± 0.11	$P < 0.001$
		13. Longer time to peak vertical GRF (s)	ANT 0.53 ± 0.10 *vs* UNA 0.58 ± 0.09	$P < 0.001$
		14. Smaller peak mediolateral GRF (%BW)	ANT 0.74 ± 0.12 *vs* UNA 0.63 ± 0.14	$P < 0.001$
		15. Smaller peak vertical GRF(%BW)	ANT 2.62 ± 0.38 *vs* UNA 2.36 ± 0.33	$P < 0.001$
		EMG findings		
		16. Higher vastus lateralis EMG activity between 20-30% of stance	UNA - magnitude not provided	P <0.01 - = 0.02
		17. Lower vastus lateralis EMG activity between	UNA - magnitude not provided	P <0.01 - =

		50-60% of stance		0.02
		18. Higher lateral gastrocnemius EMG activity between 10-20% of stance	UNA - magnitude not provided	*P* = 0.03
Besier et al., (2001) 11 male subjects	Run at 3 ms^{-1} and 4 possibilities: straight run, crossover cut at 60°, side cut at 30° and side cut at 60°	**Kinematic findings**		
		1. Increased knee flexion during weight acceptance phase (°)	ANT 31.4 *vs* UNA 31.0	*P* = 0.931
		Kinetic findings		
		2. Knee extensor moment	Not significant	
		3. Greater knee abductor moment	100% greater during UNA in weight acceptance phase	*P* < 0.001
		4. Greater internal rotator moment	90% greater during UNA in weight acceptance phase	*P* < 0.001
Cochrane et al., (2010) 50 healthy, male subjects	Run at 4.0- 4.5 ms^{-1} and 4 possibilities: straight run, crossover cut at 60°, side cut at 30° and side cut at 60°	**Knee kinematic variables**	Not significant	
		Knee kinetic variables	Not significant	

ANT = Anticipated, UNA = Unanticipated, EMG = Electromyography, GRF = Ground reaction force

Table 2.10 A summary of the biomechanical effects of anticipation on crossover cutting manoeuvres

Variable	Sagittal	Frontal	Transverse
Hip Kinematics	↑ Flexion		
Hip Moments	↓ Extensor		
Knee Kinematics	↑ Flexion	↑ Adduction	
Knee Moments	↑ Extensor		
Ankle Kinematics	↑ Dorsiflexion	↑ Inversion	
Ankle Moments	↓ Plantarflexor		

Internal joint moments reported

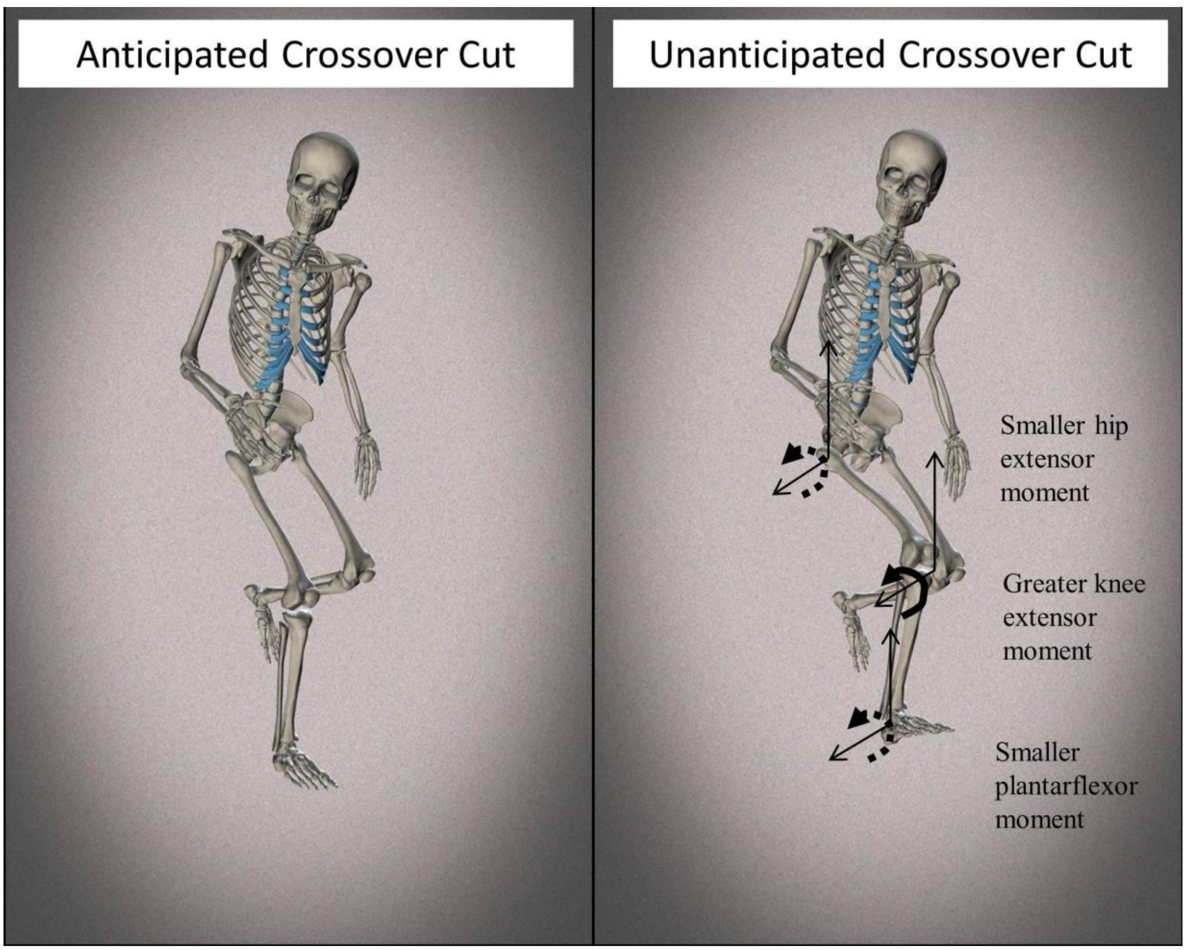

Figure 2.10 The biomechanical effects of anticipation on crossover cutting manoeuvres

2.7 The Effects of Anticipation and Fatigue Combined on Biomechanical and Neuromuscular Risk Factors for Anterior Cruciate Ligament Injuries during Cutting Activities

No studies have investigated the combined effects of anticipation and fatigue on biomechanical risk factors for ACL injuries during crossover cutting with only four studies investigating the effect on side cutting (Table 2.11). In relation to side cutting, it will be found that the combined effects of fatigue and anticipation are dependent on the nature of the fatiguing protocol.

Two studies that examined the effects of fatigue, using protocols to volitional exhaustion, found a greater combined effect of anticipation and fatigue compared with their individual effects (Borotikar et al., 2008; McLean and Samorezov, 2009). They found that the combination resulted in an average 2.1° (Borotikar et al., 2008) to 4° (McLean and Samorezov, 2009) increase in peak stance knee abduction angle in the post-fatigue, unanticipated condition compared with pre-fatigued, anticipated side cuts. Greater knee abduction angles have been shown to be strong predictors of ACL injury in the study by Hewett et al. (Hewett et al., 2005) although this has not been consistently found (Krosshaug et al., 2016; Leppanen et al., 2017). This suggests there is a potentially detrimental effect from a combination of fatigue and unanticipation during sporting tasks such as the side cut. They also found that the combination of fatigue and unanticipation resulted in a 6.6° (Borotikar et al., 2008) to 8.9° (McLean and Samorezov, 2009) increase in hip internal rotation angle at initial contact. Greater internal hip rotation has been proposed to lead to increased knee adduction moments (McLean et al., 2008) and ACL loading. Furthermore, McLean and Samorezov (McLean and Samorezov, 2009) demonstrated that the combination of fatigue and unanticipation resulted in greater knee adductor moment (21.5 Nm/54% increase) which has been shown to predict ACL injuries in 1 study (Hewett et al., 2005). Based upon these findings, Borotikar et al. (Borotikar et al., 2008) suggested that the combination of fatigue and unanticipation resulted in an elevated risk of ACL injuries during sports when that occurs. However, the fatigue protocol that was used by Borotikar et al. (Borotikar et al., 2008) and McLean and Samorezov (McLean and Samorezov, 2009) consisted of repeated squats until exhaustion and may not be representative of fatigue resulting from field sports such as soccer (Krustrup et al., 2010; Mohr et al., 2005).

Two further studies examined the combined effects of anticipation and fatigue using fatiguing protocols based upon the physiological demands of soccer (Collins et al., 2016; Khalid et al., 2015). They did not find any combined effects of anticipation and fatigue on knee joint biomechanics during run and side cutting tasks. Khalid et al. (Khalid et al., 2015) did report an anticipation by fatigue interaction effect for vertical GRF forces although this is difficult to interpret given that the average values for the four conditions (anticipated pre- and post-fatigue and unanticipated pre- and post-fatigue) were not reported. The absence of any combined effect of a running protocol and anticipation on knee biomechanics may be due to a number of reasons including the following. Firstly, there simply may not be any additional combined anticipation and fatigue effect following a running protocol that mimics the demands of soccer. Secondly, there may be limitations to the fatiguing protocols or the method of analysis. Finally the effects on the trunk and pelvis have not been investigated while the effects on the hip have only been investigated in two studies.

The fatiguing protocols used by Khalid et al. (Khalid et al., 2015) and Collins et al. (Collins et al., 2016) may result in physiological stress similar to that experienced at the end of soccer matches. However, they consisted of bouts of walking, running, jogging and sprinting (Collins et al., 2016) or a series of straight line runs with a change of direction every 20 metres (Khalid et al., 2015). This may not reflect the change of direction demands of soccer and may not challenge the muscles that contribute to frontal and transverse plane control sufficiently. This is supported by the fact that the running protocol used by Collins et al. (Collins et al., 2016) did not alter knee frontal and transverse plane biomechanics and the protocol used by Khalid et al. (Khalid et al., 2015) resulted in a reduction in knee loading in the transverse plane. Therefore, ecologically valid protocols that stress frontal and transverse plane muscular control should also be incorporated as part of the fatiguing protocol. As slightly more than half of ACL injuries in occur in the first half of soccer matches with 40% occurring during the first 15 minutes of the first and second halves (Walden et al., 2011), temporary fatigue, resulting from bursts of high intensity exercise may be a factor in sustaining ACL injuries. Therefore, studies should also use fatiguing protocols that induce temporary fatigue and simulate activities that require transverse and frontal plane control in activities such as cutting.

Despite the association between trunk biomechanics and noncontact ACL injuries (Zazulak et al., 2007), the combined effect of anticipation and fatigue on trunk and pelvic kinematics has not been investigated. Furthermore, only 2 studies have examined the effect on hip biomechanics notwithstanding the relationship between hip biomechanics and noncontact ACL injuries (Bedi et al., 2016; Ellera Gomes et al., 2014; Gomes et al., 2008; VandenBerg et al., 2017) and no studies have examined the effect on crossover cutting. In conclusion, there is a deficit in the understanding of the combined effect of anticipation and fatigue on side cutting and an absence in crossover cutting. Therefore, further research is required in this area. Future research using ecologically valid fatiguing protocols that induce temporary fatigue may increase our understanding of biomechanical risk factors for ACL injuries and facilitate the development of more efficient ACL IPPs.

Table 2.11. The effects of anticipation and fatigue on biomechanical and neuromuscular for anterior cruciate ligament injuries during side cutting

Study and participants	Activity	Fatiguing methodology	Significant Interaction Effect (Fatigue x Unanticipation)	Magnitude of differences	Statistical value
Borotikar et al., (2008) 25 female collegiate athletes	Single leg side jump/cutting manoeuvre indicated by light 350ms prior to ground contact	5 two legged squats and repeated trials until unable to complete 3 two legged squats unassisted	**Kinematic findings** 1. Increased peak stance knee abduction angle (°) 2. Increased hip internal rotation at initial contact (°) 3. Decreased hip flexion at initial contact (°)	 Pre fatigue ANT 8.2 ± 2.1 *vs* post fatigue UNA 14.8 ± 2.7 Pre fatigue ANT 3.5 ± 3.2 *vs* post fatigue UNA 7.5 ± 3.8 Pre fatigue ANT 31.0 ± 3.0 *vs* post fatigue UNA 21.9 ± 2.6	 $P < 0.05$ $P < 0.05$ $P < 0.05$
McLean and Samorezov (2009) 20 female collegiate athletes	Single leg side jump/cutting manoeuvre indicated by light 350ms prior to ground contact. Biomechanical variables examined over 100% of stance phase.	Single leg squats and single leg side jumps until unable to perform squats	**Kinematic findings** 1. Increased peak stance knee abduction angle (°) 2. Increased peak stance hip internal rotation angle (°) **Kinetic findings** 3. Greater peak stance knee adductor moment (Nm)	 Pre fatigue ANT 5.1 ± 3.6 *vs* post fatigue UNA 7.2 ± 3.5 Pre fatigue ANT 8.8 ± 4.7 *vs* post fatigue UNA 17.7 ± 4.8 Pre fatigue ANT 40.1 ± 7.5 *vs* post fatigue UNA 61.5 ± 12.2	 $P < 0.001$ $P < 0.001$ $P < 0.001$
Khalid et al., (2015) 12 collegiate soccer players (6 females, 6 males)	Cutting manoeuvres at 45° during a run of 4.5 – 5.5 ms^{-1} with an average reaction time of 290 ms for the UNA condition.	Yo-yo shuttle test. Discontinued when unable to complete two consecutive in the correct time	**Kinematic findings** **Kinetic findings** Fatigue results in greater vertical GRF, unanticipation results in smaller vertical GRF (BW)	Not significant Pre fatigue 2.13 ± 0.58 *vs* post fatigue 2.82 ± 0.60; ANT 2.35 ± 0.66 *vs* UNA 2.10 ± 0.50	 $P = 0.019$
Collins et al., (2016) 13 female, soccer players	15 m at 4.5–5.0 ms^{-1} Approximately600ms notification for UNA side cut	60 minute running protocol with varying levels of intensity	**Knee kinematic findings** **Knee kinetic findings**	No significant effect No significant effect	

GRF = Ground reaction force, UNA = Unanticipated, ANT = anticipated

2.8 The Efficacy of Anterior Cruciate Ligament Injury Prevention Programmes

ACL IPPs are generally found to be successful at reducing noncontact ACL injuries especially if they include general hip and trunk strengthening exercises. However, the effect of ACL IPPs on biomechanical risk factors for ACL injuries during cutting activities is unclear.

Two meta-analyses on female athletes found that ACL IPPs significantly reduce the incidence of ACL injuries (Myer et al., 2013; Taylor et al., 2015), whereas 1 meta-analysis that included male athletes did not find a significant effect (Grimm et al., 2015). Taylor et al. (Taylor et al., 2015) and Myer et al. (Myer et al., 2013) found a significant reduction in ACL injuries in female athletes following ACL IPPs (odds ratio: 0.35, 95% CI 0.23 to 0.54; odds ratio: 0.54, 95% CI 0.35 to 0.83, respectively). Myer et al. (Myer et al., 2013) also found a significantly greater effect in those who were 17 years of age or younger (odds ratio: 0.28, 95% CI 0.18 to 0.42) compared with those who were 18 years of age or older (odds ratio: 0.84, 95% CI 0.56 to 1.26). On the other hand, a meta-analysis conducted that included male and female participants found a non-significant protective effect of ACL IPPs (risk ratio = 0.66; 95% CI, 0.33 – 1.32, P = 0.24) (Grimm et al., 2015). These findings suggest that ACL IPPs are effective for female athletes, in particular younger females and may not be effective for males. Also, it is not clear by what mechanism ACL IPPs are effective and how they affect the biomechanics of high risk activities such as cutting.

2.8.1 The Efficacy of Components of Anterior Cruciate Ligament Injury Prevention Programmes

ACL IPPs generally consist of neuromuscular training programmes aimed at addressing modifiable biomechanical risk factors for ACL injuries. They are typically comprised of strength, agility, balance, plyometric training and technique correction (Sugimoto et al., 2016; Taylor et al., 2015). The absence of consensus on the ideal composition of the ACL IPPs in terms of exercise selection, duration, intensity and progression may be a contributing factor to the reduced protective effect of ACL IPPs in certain populations

(Myer et al., 2013; Taylor et al., 2015). To illustrate this lack of clarity, meta-analytical studies have found balance training to be associated with an increase (Taylor et al., 2015) and decrease (Myer et al., 2013) in noncontact ACL injuries. As a result of such contradictory findings, attempts have been made to identify the optimal exercises from successful ACL IPPs. Sugimoto et al. (Sugimoto et al., 2015) conducted a meta-analysis on 14 prospective, case controlled, studies investigating the effect of ACL IPPs on the incidence of ACL injuries. They divided the types of exercises into 4 categories namely (1) balance/postural exercises, (2) plyometric exercises, (3) general strengthening exercises and (4) hip and trunk control exercises. They found that the 10 studies that included general strengthening exercises in the IPPs had a significant reduction in noncontact ACL injuries (odds ratio = 0.32, 95% CI: 0.23 to 0.46, p = 0.001). On the other hand, 4 studies that did not include general strengthening exercises had no significant reduction in noncontact ACL injuries (odds ratio = 1.02, 95% CI: 0.63 to 1.64, p = 0.953). Similarly, they found that the 9 studies which included hip and trunk control exercises had a significant reduction in ACL injuries (odds ratio = 0.33, 95% CI 0.23 to 0.47, p = 0.001) whereas 5 that did not include hip and trunk exercises did not significantly affect ACL injury incidence rates (odds ratio = 0.95, 95% CI 0.60 to 1.50, p = 0.824). Furthermore, studies that included balance/postural exercises had no greater effect in the reducing the number of noncontact ACL injuries, compared with those that did not include, balance/postural exercises (odds ratio = 0.59, 95% CI 0.42 to 0.83, p = 0.003 *vs* odds ratio 0.34, 95% CI 0.20 to 0.56, p = 0.001, respectively). A similar finding was observed for the inclusion of plyometric exercises (odds ratio = 0.39, 95% CI 0.26 to 0.57, p = 0.001 *vs* odds ratio = 0.59, 95% CI 0.39 to 0.89, p = 0.012, respectively). These findings demonstrate the importance of including general strengthening exercises and targeted hip and trunk strengthening exercises in ACL IPPs. However, the effect of targeted hip and trunk strengthening exercises on biomechanical risk factors for noncontact ACL injury during cutting activities is poorly understood (Norcross et al., 2016; Pappas et al., 2015). A greater understanding of the effect of such exercises may facilitate improved efficacy of ACL IPPs.

2.8.2 The Effects of Exercise Interventions on Biomechanical Risk Factors for ACL Injuries during Cutting Activities.

There is a dearth of research investigating the effects of ACL IPPs on biomechanical risk factors for ACL injuries during cutting activities (Pappas et al., 2015). Despite the fact that ACL IPPs have been found to increase the EMG activity of the hamstrings, they have not been found to influence knee and hip joint biomechanics. Furthermore, there has been a paucity of research on the effect of ACL IPPs on unanticipated side cutting with no research investigating the effects on the biomechanics of crossover cutting.

2.8.2.1 The effect of injury prevention programmes on biomechanical factors associated with anterior cruciate ligament injury

Ten studies (Table 2.12) have investigated the effects of ACL IPPs on biomechanical factors associated with ACL injury, namely decreased knee flexion angle, increased knee abduction angle, greater knee adductor moment and greater vertical GRF. 8 studies largely did not find any effect of different ACL IPPs on these factors associated with ACL injury, 2 found negative affects while 1 found a beneficial effect. Dempsey et al (Dempsey et al., 2009) found that a 6 week technique modification programme on nine male footballers resulted in decreased peak knee adductor moments during anticipated and unanticipated side cutting, although it did not affect other knee joint biomechanics. They also found that the programme resulted in less trunk flexion away from the direction of cut and foot placement closer to the midline of the body, the latter of which correlated to the change in peak adductor moment (r = -0.468, P = 0.025). These findings indicate the beneficial effects of individualised, cutting technique training. However, given that this study investigated the effects on a small number of athletes who continued their normal training and did not include a control group, it is not clear if the improvements can be solely attributable to the intervention. Furthermore, it is important to note that the effect of the IPP on injury rates has not been investigated.

In contrast to these findings, 2 studies did not find any effect of an 18 week balance and technique training programme (Donnelly et al., 2012) and 8 weeks of the F-MARC 11^{+} (Thompson et al., 2017) on knee adductor moments. In fact, both studies found a main effect for time indicating that the intervention and control groups demonstrated an increase in peak knee adductor moments at post testing compared with pre testing. Although these findings do not suggest that the interventions used by Donnelly et al. (Donnelly et al., 2012) and Thompson et al (Thompson et al., 2017) lead to an increase in knee adductor loading, they do suggest that the interventions do not adequately address deficits in unanticipated side cutting that may contribute to ACL injury. Therefore our understanding of the effects of anticipation on biomechanical risk factors for ACL injury during side cutting and methods to ameliorate these factors, needs to be investigated further. Similarly, this needs to be investigated during crossover cutting as there has been no research in this area.

Four randomised controlled studies that have used IPPs which have been found to reduce the incidence of lower limb (Thompson et al., 2017; Thompson-Kolesar et al., 2017) and acute knee ligament injuries (Zebis et al., 2008; Zebis et al., 2016) to investigate the effect on biomechanical loading associated with ACL injuries during side cutting. They found that the IPPs did not alter peak knee flexion and abduction angles, peak hip flexion and adduction angles or peak knee adductor moments during anticipated and unanticipated side cutting. Similarly, Pappas et al. (Pappas et al., 2015) did not find any meta-analysed effect of seven ACL IPPs on knee adductor moment (p = 0.88). These findings demonstrate that successful IPPs do not change knee joint biomechanics associated with ACL injuries during anticipated and unanticipated side cutting. This suggests that successful IPPs have beneficial effects other than decreased knee loading. Given the association between trunk position and knee joint loading during cutting (Jamison et al., 2012; Mornieux et al., 2014) and the relationship between deficits in trunk control and ACL injury (Zazulak et al., 2007), it is surprising that no studies have investigated the effect of successful IPPs on trunk kinematics. Furthermore, there is limited analysis of the effect on hip biomechanics, despite the relationship between suboptimal hip range of motion and strength and ACL injury

(Bedi et al., 2016; Ellera Gomes et al., 2014; Gomes et al., 2008; Khayambashi et al., 2016; VandenBerg et al., 2017).

2.8.2.2 The effect of injury prevention programmes on technique factors associated with anterior cruciate ligament injury and loading

As lower EMG activity of the hamstrings relative to the quadriceps during cutting activities has been found to have a strong association with ACL injury (Zebis et al., 2009), 5 studies have investigated the effects of ACL IPPs on the EMG activity of the hamstrings and quadriceps with 4 studies demonstrating an increase in hamstring EMG activity following the intervention. ACL IPPs result in greater EMG activity of the semitendinosus prior to, and immediately after, initial contact (Thompson-Kolesar et al., 2017; Wilderman et al., 2009; Zebis et al., 2009; Zebis et al., 2016). This increase in hamstring activity, combined with no change (Thompson-Kolesar et al., 2017; Wilderman et al., 2009; Zebis et al., 2008) or a decrease (Zebis et al., 2016) in quadriceps activity, acts to reduce the quadriceps to hamstring activity ratio (Zebis et al., 2016) and subsequently increase tibiofemoral stability. On the other hand, one study investigating the effects of an IPP consisting of core stability and strengthening exercises did not find any effect on hamstring and quadriceps EMG activity (Bencke et al., 2000). This may be due to differences in the methodology concerning the side cutting manoeuvre. In contrast to the other four studies that required an approach run prior to cutting, participants in the study by Bencke et al. (Bencke et al., 2000) were only required to step forward one step prior to cutting.

In general, research has demonstrated that IPPs increase the EMG activity of the medial hamstrings relative to quadriceps. This may act to limit anterior tibial translation, reducing ACL loading and subsequent injury risk during cutting activities (Thompson-Kolesar et al., 2017). Furthermore, increased medial hamstring EMG activity may provide greater medial knee stability and reduced ACL loading when the knee is abducted and experiencing greater adductor moments (Thompson-Kolesar et al., 2017; Wilderman et al., 2009), both of which can predict ACL injury (Hewett et al., 2005). As the IPPs did not change knee joint kinematics and kinetics, this indicates that the IPPs altered the pattern of neuromuscular activity in a way that may reduce ACL loading (Zebis et al., 2016) and, at least partially, explain the success of the IPP in reducing acute knee ligamentous injuries. However, the same effect does not seem to occur during unanticipated side cutting and it has not been investigated during crossover cutting. A greater understanding of the effect of interventions on unanticipated cutting and crossover cutting in particular will assist in the development of ACL IPPs.

The efficacy of ACL IPPs is increased when trunk and hip strengthening and control exercises are included (Sugimoto et al., 2015). This may be due to the fact that trunk and hip strengthening and control exercises improve trunk control and strength, deficits of which predict ACL injury (Khayambashi et al., 2016; Zazulak et al., 2007). Furthermore, excessive trunk motion during cutting activities increases ACL loading and injury risk (Jamison et al., 2012; Mornieux et al., 2014). Despite the fact that 6 week core stability training protocols increase trunk strength and stability (Jamison et al., 2012, Lee & McGill, 2015, Lee & McGill, 2017) and sporting performance (Lee & McGill, 2017, Clark et al., 2017; Manchado et al., 2017), there is a paucity of research investigating their effects on biomechanical risk factors for ACL injury (Jamison et al., 2012). Only 3 studies have investigated the effect of an IPP targeting improved trunk control during side cutting with 1 study finding a beneficial effect, whereas the other 2 did not find any effect. Dempsey et al. (Dempsey et al., 2009) completed a cohort study on 9 male footballers investigating the effect of a 6 week technique correction programme on the biomechanics of anticipated and unanticipated side cutting manoeuvres. The programme emphasised the importance of placing the stance foot close to the midline during cutting, maintaining the trunk in an upright position (i.e. minimise trunk side flexion) and rotating the trunk in the direction of cut. Visual and video feedback was provided to improve technique. They demonstrated a decrease in trunk side flexion away from the direction of cut for the unanticipated side cut (pre 12.2° ± 4.9 *vs* post intervention 11.6° ± 3.5) and in particular the anticipated side cut (pre 7.4° ± 3.2 *vs* post intervention 3.9° ± 3.2). The decrease in trunk side flexion away from the direction of cut did not correlate with the decreased peak knee adductor moment observed following the intervention (r = -0.377, P = 0.135). Despite the limitations of the study in terms of study design, for example there was no control group in the study, and the small number of participants, it demonstrated that trunk kinematics during anticipated and unanticipated side cutting can be modified.

Table 2.12. The effect of ACL IPP exercises on biomechanical risk factors for acl injuries during cutting activities

Study and participants	Cutting activity	Intervention	Findings	Magnitude of differences	Statistical value
Zebis et al. (2016) 25 adolescent female football and handball players (15 INT group, 10 CON group)	ANT side cutting manoeuvre as fast and forcefully as possible	1. Neuromuscular training (NMT) programme 2. 12 weeks, 3 times a week with each session lasting 15 minutes	**Kinematic findings**	No effect	
			Kinetic findings	No effect	
			EMG findings		
			1. Decreased vastus lateralis to semitendinosus preactivity, greater effect in INT v CON group (% of MVIC)	43% (95% CI 19-55%)	*P* = 0.002
			2. Decreased vastus lateralis preactivity in INT group v CON group	23% (95% CI 10-36%)	*P* = 0.008
			3. Increased semitendinosus preactivity in INT group v CON group	18% (95% CI -21- -36%)	*P* < 0.001
			4. Increased biceps femoris preactivity in INT group v CON group	8% (95% CI -14- -2%)	*P* = 0.01
Zebis et al. (2008) 20 elite, adult female athletes (12 soccer, 8 handball). Previous season used as the study control	ANT side cutting manoeuvre as fast and forcefully as possible	1. Neuromuscular training (NMT) programme designed to improve motor control and body posture 2. 18 weeks, 2 times a week with each session lasting 20 minutes	**Kinematic findings**	No effect	
			Kinetic findings	Not investigated	
			EMG findings		
			1. Greater semitendinosus preactivity (% of peak EMG activity)	Pre 41 ± 12% *vs* post 52 ±16%	*P* < 0.01
			2. Greater semitendinosus activity in initial landing phase (% of peak EMG activity)	Pre 29 ± 12% *vs* post 39 ±20%	*P* < 0.05
			3. No change in quadriceps activity (% of peak EMG activity)	Pre 42 ± 14% *vs* post 32 ±11%	*P* < 0.01
			4. Decrease in biceps femoris activity in initial landing phase (% of peak activity)	Pre 68 ± 13% *vs* post 58 ±12%	*P* < 0.05
			5. Decrease in gluteus medius preactivity (% of peak EMG activity)	Pre 29 ± 12% *vs* post 39 ±20%	*P* < 0.05
			6. Decrease in gluteus medius activity in landing phase (% of peak EMG activity)	Pre 62 ± 15% *vs* post 51 ±14%	*P* < 0.05
94 female soccer players. INT group	ANT and UNA side cutting.	1. F-MARC 11+ injury prevention warm-up.	**Kinematic findings**	No effect	
			Kinetic findings	No effect	

-28 PREAD, 22 AD. CON group- 23 PREAD and 21 AD	Approach speed of 3.8 ± 0.5 ms^{-1}	2. 15 sessions (2 per week for 7-8 weeks) of 25 minutes	**EMG findings** 1. Increase in precontact knee flexor:extensor EMG activity. Greater in ANT cutting for PREAD vs AD following intervention	ANT: PREAD INT 0.3 ±0.02 *vs* AD INT -0.3±0.27	*P* = 0.004 *P* = 0.002
Thompson et al., (2017) 51 female soccer players.INT group - 28 PREAD, CON group - 23 PREAD	UNA and ANT side cutting. Approach speed of 3.8 ± 0.5 ms^{-1}	1. F-MARC 11+ injury prevention warm-up. 2. 15 sessions (2 per week for 7-8 weeks) of 25 minutes	**Kinematic findings** 1. Peak knee flexion angle (°) 2. Decreased peak knee abduction angle in UNA side cutting in control group (°) 3. Peak hip adduction angle (°) 4. Decreased peak ankle eversion angle in ANT side cutting (°) **Kinetic findings** 1. Peak knee extensor moment (%BW. Ht) 2. Greater peak knee adductor moment changes (%BW. Ht) for INT and CON groups 3. Peak hip abductor moment (%BW. Ht) 4. Smaller peak ankle invertor moment in ANT and UNA(%BW. Ht)	No change UNA:CON 3.1± 0.8 *vs* INT 0.8±0.5 No change ANT:CON 3.2 ±1.0 *vs* INT -0.2 ±1.2 No change UNA: CON 0.44 ±0.50 *vs* INT 0.99 ±0.47 No change ANT: CON - 0.57 ±0.24 *vs* INT 0.48 ±0.32 UNA: CON - 0.59 ±0.30 *vs* INT 0.69 ±0.29	*P* = 0.018 *P* = 0.034 *P* = 0.042 *P* = 0.015 *P* = 0.004
Weltin et al. (2017) 24 active female participants (12 in INT, 12 in CON)	UNA side cutting with 4.0 ±0.2 ms^{-1} approach speed and direction of cut indicated 650 ms prior to initial contact	1. Plyometric programme (CON) 2. Perturbation enhanced plyometric programme (INT) 3. 3 sessions per week for 4 weeks	**Kinematic findings** 1. Lateral trunk flexion 2. Reduction in trunk rotation (°) away from direction of cut for INT and CON groups 3. Increased pelvic rotation in direction of cut	No effect CON: pre 14.4 ± 8.9 *vs* post 10.5 ±8.1, INT; pre 13.6 ± 8.0 *vs* post 6.4 ±8.8 CON: pre 7.8 ± 5.8 *vs* post 9.2 ±5.7,	*P* = 0.008 η^2 = 0.277 *P* = 0.049

			  Kinetic findings	INT; pre 10.8 ± 5.5 *vs* post 13.3 ±5.6 No effect	$\eta^2 = 0.165$
Jamison et al. (2012) 21 male American footballers (10 INT, 11 CON)	UNA side cutting. 3 step jog approach to side cut.	1. INT –trunk stability exercises, CON – general strength training programme 2. 3, 1 hour sessions for 6 weeks	**Kinematic findings** **Kinetic findings** 1. Knee adductor moment 2. Knee external rotator moment	Not investigated No effect No effect	
Donnelly et al. (2012) 34 male Australian rules footballers. 20 INT group, 14 CON group	ANT and UNA side cutting with an approach speed of 4.5 – 5.5 ms^{-1}	1. INT – balance and side cut technique training 2. CON – straight line acceleration training 3. 20 min sessions, 3 times per week for 18 weeks	**Kinetic findings** 1. Smaller peak knee external rotator moment for INT and CON groups ($Nm.kg^{-1}.m^{-1}$) 2. Greater knee adductor moment for INT and CON groups ($Nm.kg^{-1}.m^{-1}$) 3. No specific effect for INT group	Pre ANT (INT and CON): 0.33 ± 0.36 *vs* post ANT 0.18 ±0.09 Pre UNA (INT and CON): 0.48 ± 0.27 *vs* post ANT 0.63 ±0.40	$P = 0.025$ $P = 0.022$
Wilderman et al. (2009) 30 female basketball players (15 to INT group, 15 to CON group)	ANT side cut with an approach speed of 3.3 – 4.3 ms^{-1}	1. CON – regular training 2. INT – agility training programme. 15 minutes in length, 4 sessions per week for 6 weeks	**Kinematic findings** **Kinetic findings** **EMG findings** 1. Greater semitendinosus EMG activity in initial stance phase in INT group post intervention compared with CON group (%MVIC)	No effect No effect INT: pre 60.9 ± 22.9, post 103.0 58.6 *vs* CON: pre 88.5 ± 27.9, post 71.2 ± 27.5	$P < 0.002$
Bencke et al. (2000) 17 male handball players (10 to INT group, 7 to CON group)	ANT side cutting following one step forwards	1. INT - single leg squats, single leg jumps, hamstring pulls, hip abductions. 2. Two sessions per week for 12 weeks	**Kinetic findings** **EMG findings** 1. EMG activity levels of semitendinosus, biceps femoris and quadriceps 2. Shorter onset of semitendinosus EMG preactivity in INT group (ms)	 No effect INT pre 132 ± 18 *vs* post 121 ±17	$P < 0.05$
Di Stefano et al. (2011) 65 adolescent	ANT side cutting	1. CON = Traditional ACL IPP programme 2. INT - The traditional	**Kinematic findings** 1. Decreased knee external rotation (°) at	INT 7.73 ±10.71 *vs* CON -0.35 ± 7.76	$P = 0.03$

soccer players (27 female, 38 male)		ACL IPP modified for adolescents	initial contact in INT v CON group **Kinetic findings**	No effect	
Dempsey et al. (2009) 9 male, team sport athletes	ANT and UNA side cutting with an approach speed of 5.2 ms^{-1}. For UNA side cutting, direction of cut was indicated approximately 400ms prior to foot contact	1. 2, 15 minute sessions per week for 6 weeks 2. Participants given oral and visual technique feedback	**Kinematic findings**		
			1. Knee flexion angle (°)	No effect	
			2. Decreased difference (cm) of foot from pelvis for ANT and UNA	ANT:pre 36.9±4.0 *vs* post 34.6 ± 4.4, UNA: pre 36.6± 1.7 *vs* post 34.4 ± 5.1	P = 0.039
			3. Trunk lateral flexion (°)	ANT: pre 7.4 ± 3.2 *vs* post 3.9 ± 3.2, UNA:pre 12.2 ± 4.9 *vs* post 11.6±3.5	P = 0.005
			Kinetic findings 1. Smaller peak adductor moment (Nm.kg^{-1}.m^{-1}) for ANT and UNA side cutting	ANT:pre 0.38±0.26 *vs* post 0.24 ± 0.22, UNA:pre 0.40±0.23 *vs* post 0.26 ± 0.11	P = 0.034

INT = Intervention, CON = Control, ANT = anticipated, UNA = unanticipated, ACL = Anterior cruciate ligament, MCL = Medial collateral ligament, PCL = Posterior cruciate ligament, AD = Adolescent. PREAD = Preadolescent, EMG = Electromyography

Only 1 study has examined the effects of a trunk control and strengthening programme on the biomechanics of unanticipated side cutting. Jamison et al. (Jamison et al., 2012) conducted a randomised controlled trial on 21 male American footballers, 10 of whom were assigned to the trunk control and strengthening programme and 11 to the control or general strengthening programme. The trunk control and strengthening programme consisted of static endurance and strengthening exercises and resulted in a significant increase ($p < 0.05$) in frontal plane trunk endurance as measured by the side plank test (pre: 81.8s ± 19.0 *vs* post: 88.6s ± 14.0) compared with the control group (pre: 78.8s ± 27.0 *vs* post: 71.7s ± 19.3) and lateral trunk core strength as measured by a resisted cable pull (pre 467.6N ± 92.4 *vs* post: 512.1N ± 105.2). However, the intervention did not affect peak knee joint adductor and external rotator moment during the weight acceptance phase of unanticipated side cutting. As hip and trunk biomechanics during side cutting were not investigated, it is not clear if a static trunk control and strengthening programme can change trunk kinematics during unanticipated side cutting. Also, this study does not provide insight into the effect of trunk control and strengthening exercises on trunk kinematics and knee loading during crossover cutting and anticipated side cutting. The absence of an effect in the study by Jamison et al. (Jamison et al., 2012) may be due to the absence of exercises targeting the dynamic control of the centre of mass. Future studies investigating the effect of trunk control and strengthening should incorporate dynamic exercises (Jamison et al., 2012) and investigate the effects on trunk and lower limb biomechanics during anticipated and unanticipated side and crossover cutting manoeuvres.

Finally, Weltin et al. (Weltin et al., 2016) investigated the effect of perturbation training on the biomechanics of unanticipated side cutting. They randomly assigned 24 female athletes (soccer, handball and basketball) to a 4 week, "perturbation-enhanced plyometric training" (intervention) or a 4 week, traditional plyometric training programme (control). The intervention consisted of lateral rebound jumps landing on to a motorised plate which applied the perturbation. The intervention did not affect trunk kinematics or knee joint loading during the weight acceptance phase of unanticipated side cutting when compared with the control group. Therefore it can be deduced that this method of "perturbation-enhanced plyometric training" does not improve trunk kinematics and knee biomechanics during anticipated side cutting. This is potentially due to the fact that perturbations did not specifically target centre of mass control proximally.

2.8.2.3 Conclusion

ACL IPPs do not affect knee biomechanics associated with ACL injury during anticipated and unanticipated side cutting. However, they have been found to alter hamstring relative to quadriceps EMG activity which may improve knee joint stability, particularly medial joint stability, during anticipated side cutting activities. Furthermore, the effect of ACL IPPs on the biomechanics of unanticipated side cutting and both anticipated and unanticipated crossover cutting is not understood. Finally, further research is required to understand the effect of exercises that target dynamic trunk and hip control on trunk kinematics during anticipated and unanticipated cutting activities.

2.9 Methods of Biomechanical Analyses

The biomechanics of high-risk activities in terms of ACL injury, such as cutting and landing, have traditionally been investigated using discrete point analysis (DPA). Although this method has provided a useful platform, it has a number of limitations that restrict our understanding of complex movements. Firstly, DPA extracts discrete or singular measures of a particular variable at a preselected time point or time span, for example peak knee abduction angle at initial contact or in the weight acceptance phase. This can over simplify the original datasets (Pataky et al., 2013) leading to the analysis of less than 5% of the original data (Richter et al., 2014). Furthermore, the pre-selection of time points introduces a potential source of bias (Pataky et al., 2015). Critically, as DPA does not take into account temporal characteristics of the selected variables, this can lead to comparison of unrelated features (Richter et al., 2014) and subsequent inaccurate conclusions. For example, using DPA to identify peak vertical GRF may result in the unintentional comparison of the passive peak and active peak in two subsequent side cuts (Figure 2.11). Therefore the shortcomings of DPA may limit our understanding of the biomechanics of complex movements such as cutting. Alternative methods, which analyse phases of biomechanical variables, are likely to add to the existing knowledge (Shultz et al., 2015) and may subsequently facilitate the development, and improved efficacy, of IPPs.

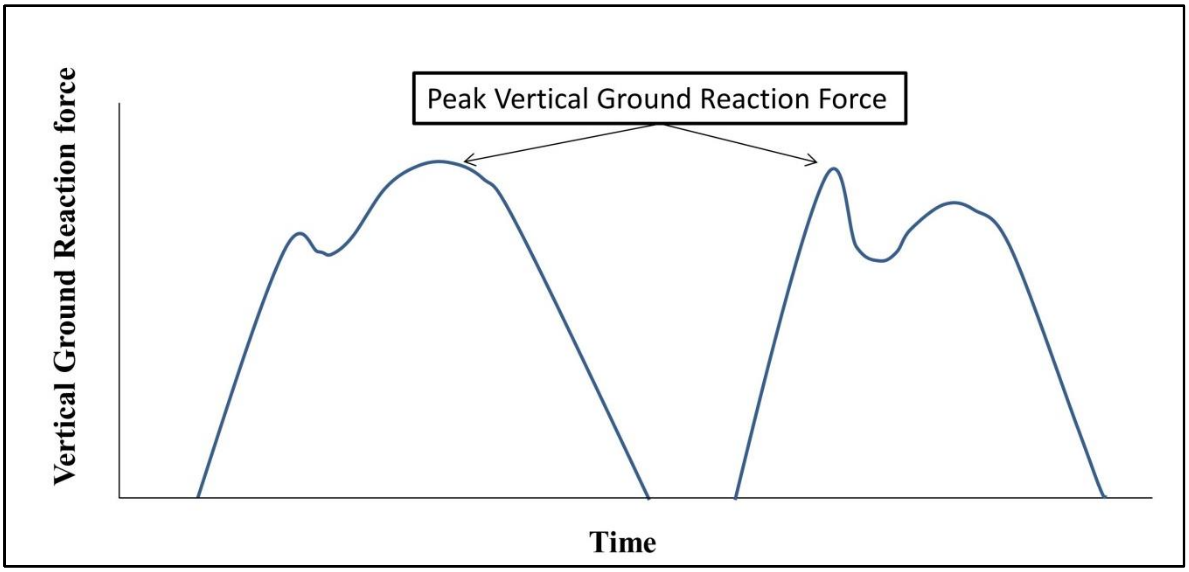

Figure 2.11 Demonstration of how discrete point analysis may lead to the comparison of functionally different aspects of the same variable

Continuous data analysis techniques, such as statistical parametric mapping (SPM), are more effective than DPA at identifying biomechanical characteristics of complex movements (Richter et al., 2014). They analyse the original biomechanical variables rather than predetermined discrete points, avoiding a potential source of bias. Prior to continuous data analysis, the original variables can also undergo dynamic time marking. This is a process by which the waveform of each variable is landmarked so that comparisons are made of the same biomechanical features (Richter et al., 2014). Furthermore, SPM provides a test statistic (for example an F value in ANOVA) for each data point of the variable. This can allow for phases of significant differences to be identified rather than discrete points. As this can lead to a large amount of data available for analysis, randomised field theory (Adler, 2007) can also be incorporated in SPM to determine a critical threshold which ensures that statistical differences found do not simply occur by chance. This will reduce the likelihood of a type 1 error (Pataky et al., 2013). For these reasons, SPM has been found to be more effective at identifying features related to task outcome (e.g. knee flexion angle and GRF) during activities such as the vertical drop jump (Pataky et al., 2013; Richter et al., 2014). Therefore, SPM may also be more sensitive at identifying biomechanical risk factors for ACL injuries during cutting activities (Shultz et al., 2015), which may ultimately facilitate the improvement of ACL IPPs.

2.10 The Effect of Fatigue on Dynamic Balance

Dynamic balance requires an individual to maintain postural control around a base of support during movements such as cutting and landing and it replicates the demands of sporting activity to a greater extent that static balance (Gribble et al., 2012). Dynamic balance is achieved by the coordination of the neuromuscular system's ability to control the centre of mass with respect to the base of support, and the sensorimotor system's capacity to process afferent information and instigate the appropriate response (Patla et al., 1999). It is one way of quantifying neuromuscular control (Gribble and Hertel, 2004a; Gribble and Hertel, 2004b) and can be measured in numerous ways including the Star excursion balance test (SEBT) (Gribble and Hertel, 2004a; Gribble et al., 2007; Gribble et al., 2009; Steib et al., 2013; Zech et al., 2012), the Y balance tests (a modification of the SEBT) (Johnston et al., 2018), time to stabilisation following a jump (Shaw et al., 2008; Wikstrom et al., 2004), centre of pressure sway (Steib et al., 2013; Zech et al., 2012) and the biodex stability system (Reimer and Wikstrom, 2010; Salavati et al., 2007; Wright et al., 2013; Wright et al., 2013). The SEBT only requires a tape measure, is easy to use and a reliable test with intra class correlation values of 0.84 – 0.92 (Kinzey and Armstrong, 1998; Munro and Herrington, 2010). In a prospective study, Plisky et al. (Plisky et al., 2006) found that participants with SEBT score deficits were 2.5 – 6.5 more likely to sustain lower extremity injuries. It has the ability to detect detriments in postural control associated with pathologies of the ankle, ACL and patellofemoral joint (Gribble et al., 2012). Therefore, the SEBT is a practical, reliable and sensitive test of dynamic balance.

Fatigue alters neuromuscular and sensorimotor control due to changes in the efficacy of muscle contraction and the afferent information from muscle spindles and may subsequently negatively affect dynamic postural control (Gribble et al., 2012). As deficits in dynamic postural control are associated with increased injury risk (Plisky et al., 2006), it has been suggested that fatigue-induced detriments in dynamic postural control may contribute to an increased lower limb injury risk (Zech et al., 2012). Despite a number of studies demonstrating a significantly adverse effect of fatigue on static and dynamic postural control (Gribble and Hertel, 2004b; Salavati et al., 2007; Steib et al., 2013; Wright et al., 2013; Zech et al., 2012) there has been little investigation on the influence of high intensity, intermittent exercise on dynamic postural control.

A number of studies have assessed the influence of fatigue on dynamic postural control (Table 2.13) with the majority showing that fatigue has a negative influence on dynamic postural control (Gribble et al., 2004; Gribble et al., 2007; Gribble et al., 2009; Johnston et al., 2018; Miller and Bird, 1976; Reimer and Wikstrom, 2010; Salavati et al., 2007; Steib et al., 2013; Wright et al., 2013; Zech et al., 2012). However, methodological variations in terms of study populations, assessments of dynamic balance, fatiguing protocols used and methods to determine fatigue, have led to some inconsistencies in results. This may explain why some studies have not found fatiguing protocols to detrimentally effect dynamic postural control (Wright et al., 2013; Zech et al., 2012). For example, Zech et al. (Zech et al., 2012) found in a group of 19 male handball players that a treadmill running or step up resistance exercise programme had no effect on dynamic postural control as measured by 4 directions and the mean value of the SEBT. In a study of 16 active adults, Wright et al. (2013) demonstrated that a cycling fatiguing protocol was not found to have an effect on dynamic postural control as measured by the Biodex Balance system whereas a treadmill running protocol had a detrimental effect on the stability index. This supports the previously held view that the effects of fatigue are task dependent fatigue (Enoka and Duchateau, 2008) and highlights the importance of using ecologically valid fatiguing protocols. For example, the level of fatigue induced in participants in previous research using sport simulated activities has often been greater than those observed in sport competition (Knicker et al., 2011) which may reduce the applicability of these findings to particular sports.

In conclusion, fatigue seems to negatively affect dynamic balance, which is an important requirement during jumping, landing and cutting activities. However, there are methodological inconsistencies in the current research in terms of methods to induce and determine fatigue. The effects of fatigue, particularly temporary fatigue following high intensity, intermittent exercise, on dynamic balance should be determined using ecologically valid fatiguing protocols.

Table 2.13 The effects of fatiguing protocols on dynamic balance

Study and participants	Assessment of postural control	Method used to induce fatigue	Method to determine end of protocol	Findings
Johnston et al., (2018) 20 healthy female and male participants	1. Y balance test assessed immediately post, 10 minutes post and 20 minutes post fatigue protocol	Modified Wingate protocol. Participants had to cycle on a stationary bike at maximum speed for 60 seconds.	Completion of task	Y balance test scores negatively affected by fatigue. Effects of fatigue lasted for 20 minutes following cessation of fatigue protocol
Gribble et al., (2004) 16 healthy participants (8 male, 8 female) and 14 with CAI (7 male, 7 female)	2. SEBT (3 out of 8 directions – anterior, medial and posterior)	Isokinetic exercise protocol 1. Ankle plantarflexors and dorsiflexors 2. Knee flexors and extensors 3. Hip flexion and extension 4. Lunging exercise 5. Control	1-3 Torque production below 50% of peak torque 4 – incorrect lunge technique or unable to complete at required pace	Fatigue and CAI adversely effected dynamic postural control ($p<0.05$)
Gribble et al., (2007) 30 physically active students. 14 with CAI and 16 without	1. SEBT (3 out of 8 directions – anterior, medial and posterior) 2. Sagittal plane kinematic analysis	1. Open chain isokinetic fatigue of plantar flexors and dorsiflexors 2. Lunging fatigue protocol with weighted vest of 10% body weight	Isokinetic torque dropped below 50% of peak torque for both muscle groups Unable to keep to correct rate or correct technique	CAI contributes to the decline in SEBT scores (medial and lateral) after lunging protocol only. Ankle exercise protocol had group effect (CAI weak predictor) in posterior direction
Gribble et al., (2009) 8 physically active males and 8 physically active females	1. SEBT (Anterior, medial and posterior directions only)	Isokinetic exercise protocol 1. Ankle plantarflexors and dorsiflexors 2. Knee flexors and extensors 3. Hip flexion and extension 4. Lunging exercise	1-3 Torque production below 50% of peak torque 4 – incorrect lunge technique or unable to complete at required	Protocols result in reduced normalised reach distances in SEBT. Females effected to a lesser extent by fatigue protocols

		5. Control	pace	
Johnston et al., (1998) 12 healthy males, 8 healthy females	1. KAT balance systems	Closed chain isokinetic dynamometer (similar to stairmaster) for hip, knee and ankle musculature	Torque production below 50% of peak torque	The exercise protocol negatively affected static balance (p<0.001) There was no effect on dynamic postural control ($p > 0.05$)
Miller and Bird, (1976) 100 males grouped into 4 INT and 1 CON groups	1. Dynabalometer	Localised muscular fatiguing protocols : 1. Dorsiflexors 2. Plantarflexor 3. Abdominals 4. hip and knee flexors and extensors	Subjective report	Localised protocol for hip and knee flexors and extensors negatively affected dynamic postural control (p<0.01)
Reimer and Wikstrom, (2010) 18 recreationally active university students	1. Biodex Stability System assessed the overall, medial/lateral, and anterior/posterior stability index (stability level 4)	1. single leg squat protocol – @65% RM 2. single leg calf raise @65% RM	Inability to maintain the standardised pace for three consecutive repetitions at a rate of 33 or 45 beats per minute respectively	Fatigue of both the ankle and hip musculature led to postural control impairments for the medial–lateral stability index ($p < 0.01$), and anterior–posterior stability index ($p < 0.01$). Only ankle fatigue resulted in deficits in the overall stability index ($p < 0.01$).
Salavati et al., (2007) 20 healthy males	1. Biodex Balance system (stability level 7) assessed overall, anterior-posterior, and medial-lateral stability indices	Isokinetic fatigue of 1. Ankle plantarflexors and dorsiflexors 2. Ankle invertors and evertors 3. Hip flexors and extensors 4. Hip abductors and adductors	Torque production below 50% of peak torque	All protocols associated with a significant increase in stability index. Fatigue of hip musculature had greater effect than fatigue of the ankle musculature
Shaw et al., (2008) 10 female volleyball players	1. Time to stabilisation (TTS)following jump landing using force platform	South Eastern Missouri agility drill with lunging and jumping exercise	Time to completion increase by 50% compared to baseline	Anterior posterior TTS increased post fatigue Medial to lateral TTS not effected by protocol

Steib et al., (2013) 14 (8 males, 6 females) with previous ankle sprain, 16 (11 male, 5 female) controls	1. TTS following jump landing 2. SEBT (average of 4 directions: anterior posterior, medial and lateral) 3. COP sway velocity	Treadmill running	17 on Borg RPE scale	Greater effect in participants with a history of ankle sprain compared to control group. 1. Increased Anterior to posterior TTS (p=0.05) 2. Decreased normalised SEBT distances (p=0.03)
Wikstrom et al., (2004) 8 healthy males and 12 healthy females	1. TTS following a jump and land	1. Isokinetic dynamometer of dorsiflexors and plantarflexors 2. Functional fatigue protocol	1. Force production below 50% of peak torque 2. Increase lap completion time by 50%	1. Fatigue results in greater vertical TTS and GRF 2. Significantly reduced TTS in medial lateral TTS 3. No difference between functional and isokinetic fatigue protocols. 4. No changes in medial lateral and anterior posterior TTS
Wright et al., (2013) 16 (11 male, 5 female)	1. Biodex Balance System via the Dynamic Balance Test post exercise and for up to 21 minutes stability index, anterior poster index medial lateral index	1. incremental cycle ergometer test using the ACSM cycle protocol 2. incremental treadmill test using the Bruce protocol	HR > 85% of age predicted max heart rate	1. Cycling fatiguing test did not significantly affect balance (p>0.01). 2. Treadmill protocol resulted in increased stability index immediately post protocol (p < 0.01)
Zech et al., (2012) 19 male handball players	1. SEBT 2. Centre of pressure sway (COP) (static)	1. Treadmill running 2. Step up with resistance	1. Subjective exhaustion (17 Borg RPE Scale) 2. Unable to complete a step up	1. No effect on SEBT 2. COP sway velocity increased post fatiguing protocols

HR = Heart rate, RPE = Rate of perceived exertion, CAI = Chronic ankle instability, SEBT = Star excursion balance test, INT = Intervention, CON = Control

2.11 The Effect of Fatigue on Neurocognitive Function

Deficits in baseline neurocognitive function are associated with noncontact ACL injuries (Swanik et al., 2007). Swanik et al. (Swanik et al., 2007) found that 8 American footballers who sustained ACL injuries had lower baseline neurocognitive performance scores when compared with 80 matched controls. Specifically, they found that injured athletes demonstrated significantly slower reaction time (p =.002), processing speed (p=.001) and performed worse on visual (p > 0.001) and verbal memory (p = 0.045). Sporting activities require adequate neurocognitive function to monitor the sporting environment, filter the appropriate information and to implement the appropriate motor programme (Swanik et al., 2007). Successful completion of these activities is achieved through a combination of feed-forward and feedback control which is ultimately regulated by the neurocognitive function of the cerebral cortex (Swanik et al., 2007). Provided there is sufficient time, feed-forward control allows for retrieval and implementation of the neuromuscular patterns from an internal programme based upon prior experience of the task (Kandel, 1999). Suboptimal neurocognitive function and regulation of these processes may result in impaired neuromuscular control predisposing an athlete to noncontact ACL injuries (Swanik et al., 2007). Additionally, exercise has been found to acutely affect neurocognitive function and therefore may affect an athlete's ability to safely complete sporting tasks such as cutting and landing.

A number of theories attempt to explain the relationship between cognitive performance and exercise intensity. The inverted u-theory proposes that moderate level activity will have the greatest beneficial acute effect on neurocognitive function while low or high intensity exercise will have a detrimental effect (Chang et al., 2012). In addition, the central drive theory proposes that higher intensity exercise will have the greatest beneficial effect on neurocognitive function (Chang 2012). On the other hand, the transient hypofrontality hypothesis proposed by Dietrich (Dietrich, 2006) proposes that exercise of sufficient intensity will result in competition with the frontal cortex for metabolic resources, resulting in an impairment of neurocognitive function. With these conflicting theories in mind, two meta-analytic reviews have generally found small, but beneficial, effects of exercise (Chang et al., 2012; Lambourne and Tomporowski, 2010). Both meta-analyses found a number of factors that influenced the findings including differences in exercise protocols (mode and intensity) and participant fitness levels (Chang et al., 2012; Lambourne and Tomporowski, 2010). This highlights the importance of ecologically valid exercise protocols on athletes in order to understand the effects of fatigue on neurocognitive function in athletes.

In a meta-analysis of 79 studies on all age groups, Chang et al. (Chang et al., 2012) found a small, but beneficial effect of acute exercise on neurocognitive function (Cohen's d = 0.097) during low to moderate intensity exercise. However, this improvement was not evident after high intensity exercise. In contrast with these findings, a meta-analysis on 40 studies of young adults by Lambourne and Tomporowski (Lambourne and Tomporowski, 2010) found that exercise resulted in a decline in neurocognitive function during exercise lasting 20 minutes or less. This duration of exercise may correspond to temporary fatigue, caused by bouts of high intensity exercise, which is inadequately researched (Knicker et al., 2011). Importantly, Lambourne and Tomporowski (Lambourne and Tomporowski, 2010) also discovered that the mode of exercise affected the overall outcomes. They found that fatigue protocols involving running lead to a decrease in neurocognitive function in contrast to studies that used cycling exercise. Critically, there is a paucity of research investigating the effects of high intensity, intermittent exercise, which results in temporary fatigue, on neurocognitive function in athletes.

In summary, neurocognitive function is required for adequate neuromuscular control during sporting activities such as cutting and landing and deficits in neurocognitive function are associated with ACL injuries. Exercise induced fatigue, particularly temporary fatigue, may be a risk factor for ACL injury and can also negatively affect neurocognitive function. Therefore in order to investigate potential relationships between fatigue and ACL injury, it is important to appreciate the connection between fatigue and neurocognitive function.

2.12 Conclusion of Literature Review

Noncontact ACL injuries lead to serious short and long-term consequences for an athlete making the development of prevention programmes critical. For an injury to occur, the loading experienced must be greater than the structural integrity of the ACL (Shultz et al., 2015). While it is difficult to change the structural integrity of the ACL, loading of the ACL may be altered by modifying the biomechanics of high risk activities such as cutting and landing. Cadaveric studies demonstrate greatest ACL loading when the tibiofemoral joint experiences a combination of extensor, adductor and external rotator moments and joint compression at small knee flexion angles (Shin et al., 2009). This pattern of loading is frequently observed at the time of ACL injury (Boden et al., 2000). In addition, greater compressive knee forces, knee valgus angle, knee adductor moment and decreased knee flexion angle, have been found to be predictors of ACL injuries, although the findings are equivocal. Furthermore, biomechanical and neuromuscular risk factors along the kinetic chain have also been associated with ACL injuries including limited hip range of motion and strength, suboptimal trunk control and reduced hamstring activity. Laboratory based studies have found that these risk factors can be reduced by altering the biomechanics of high risk activities. Therefore, the correction of these biomechanical risk factors has been the focus of ACL IPPs.

Fatigue has been demonstrated to negatively affect potential ACL loading during landing and cutting activities (Kernozek et al., 2008; McLean et al., 2007; Zebis et al., 2010). As ACL injuries occur more frequently in the first 15 minutes of each half and in the first half of a match compared with the second half (Walden et al., 2011), temporary fatigue may be a factor in sustaining ACL injuries. Furthermore, the risk is proposed to be greater when unanticipated high-risk activities are performed in a fatigued state (Borotikar et al., 2008). Although, laboratory studies have found that fatigue alters the biomechanics of the trunk, hip and knee during cutting activities, which may increase ACL loading and subsequent injury risk, they have frequently used fatiguing protocols that do not mimic the physiological demands of field sports. It is important to remember that fatigue is task dependent, i.e. the exercise that is performed dictates the mechanisms that cause fatigue (Enoka and Duchateau, 2008). For this reason, research investigating the effects of ecologically valid, high intensity, intermittent exercise protocols on cutting biomechanics is required to improve our understanding of biomechanical risk factors for ACL injuries.

ACL injuries frequently occur when an athlete performs unanticipated cutting activities in response to the sporting environment. Performing unanticipated cutting activities appear to result in greater potential ACL loading. Unanticipated side cutting leads to altered trunk and hip biomechanics and increased knee internal rotation angles and knee extensor and external moments, all of which are associated with increased ACL loading. Although there is significantly less research on crossover cutting, there is evidence that ACL loading may increase as a result of the increased knee extensor moment and altered knee adduction angles. However, there is limited research investigating the effects of anticipation on trunk and pelvic kinematics during side cutting with no research on this during crossover cutting. Furthermore, despite the proposal that ACL injury risk is heightened when unanticipated high risk activities are performed when fatigued, the effects of both anticipation and fatigue on crossover cutting or on trunk and pelvic kinematics during both side cutting have not been investigated. Finally, no studies that have analysed the effect of anticipation and fatigue have used high intensity, intermittent exercise protocols.

ACL IPPs have been found to reduce the incidence of ACL injuries, particularly when they include trunk and hip strengthening exercises. However, given the association between trunk and hip biomechanics and ACL injury (Bedi et al., 2016; Khayambashi et al., 2016; Zazulak et al., 2007), it is surprising that no studies have investigated the effect of targeted trunk and hip strengthening and control exercises on trunk kinematics during cutting. Furthermore, there has been a paucity of research on the effect of ACL IPPs on the biomechanics of unanticipated side cutting with no research investigating the effects on crossover cutting. A greater understanding of the effect of trunk and hip strengthening exercises may facilitate an improved efficacy of ACL IPPs.

Chapter 3 An investigation into the effect of a high intensity, intermittent exercise protocol on dynamic postural control in males and females.

Study 1

Study 1 "A high-intensity, intermittent exercise protocol and dynamic postural control in men and women."

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STATEMENT OF CONTRIBUTION: Kieran Moran was the research supervisor for this study. Aoife Burke and Elaine White assisted in the data collection and analysis.

3.1 Abstract

Context: Deficits in dynamic postural control predict lower limb injury. Differing fatiguing protocols negatively affect dynamic postural control. The effect of high intensity, intermittent exercise on dynamic postural control has not been investigated.

Objective: To investigate the effect of a high-intensity, intermittent exercise protocol (HIIP) on the dynamic postural control of men and women as measured by the Star Excursion Balance Test (SEBT).

Design: Descriptive Laboratory Study

Setting: University gymnasium

Patients or Participants:

Twenty male (20.83 ± 1.50 years, height = 179.24 ± 7.94 cm, mass = 77.67 ± 10.82 kg) and 20 female athletes (age 20.45 ± 1.34 years, height = 166.08 ± 5.83 cm, mass = 63.02 ± 6.67 kg) athletes.

Interventions: We recorded SEBT measurements at baseline, pre-HIIP, and post-HIIP. The HIIP consisted of 4 repetitions of 10-m forward sprinting with a 908 change of direction and then backward sprinting for 5 m, 2 repetitions of 2- legged jumping over 5 hurdles, 2 repetitions of high-knee side stepping over 5 hurdles, and 4 repetitions of lateral 5-m shuffles. Participants rested for 30 seconds before repeating the circuit until they reported a score of 18 on the Borg rating of perceived exertion scale.

Main Outcome Measures: A mixed between- and within– subjects analysis of variance was conducted to assess time (pre-HIIP, post-HIIP) 3 sex interaction effects. Subsequent investigations assessed the main effect of time and sex on normalized maximal SEBT scores. We used intraclass correlation coefficients to determine the test-retest reliability of the SEBT and paired-samples t tests to assess the HIIP effect on circuit times.

Results: We found a time x sex effect ($F_{8,69} = 3.5$; *P* range, <.001–.04; η^2 range, 0.057–0.219), with women less negatively affected. We also noted a main effect for time, with worse normalized maximal SEBT scores post fatigue ($F_{8,69} = 22.39$; $P < .001$; η^2 range, 0.324–0.695), and for sex, as women scored better in 7 SEBT directions ($F_{8,69} = 0.84$; *P* range, ,.001–008; η^2 range, 0.088–0.381). The intraclass correlation coefficients demonstrated high (0.77–0.99) test-retest repeatability. Paired-samples t tests demonstrated increases in circuit time post-HIIP ($P < .001$).

Conclusion: The HIIP-induced fatigue negatively affected normalized maximal SEBT scores. Women had better scores than men and were affected less negatively by HIIP-induced fatigue.

3.2 Introduction

Deficits in dynamic postural control are risk factors for sustaining lower limb injuries (Plisky et al., 2006). Dynamic postural control requires that postural control be maintained around a base of support during movement, thereby mimicking sporting demands more than static postural control does (Gribble et al., 2012). It is achieved by coordinating neuromuscular and somatosensory systems to process sensory information and react accordingly (Patla et al., 1999). The Star Excursion Balance Test (SEBT) is a measure of dynamic postural control that can predict injury (Plisky et al., 2006). It is sensitive to dynamic postural control deficits after ankle (Gribble et al., 2007) and anterior cruciate ligament (ACL) injuries (Herrington et al., 2009). The SEBT can also detect improvements in dynamic postural control after interventions in patients with chronic ankle instability (Hale et al., 2007) and healthy participants (Leavey et al., 2010). However, research into the reliability of normalized SEBT scores is lacking (Gribble et al., 2013).

Lower limb injuries are common in sports with intermittent bouts of high intensity exercise and multiple changes of direction (e.g. soccer, basketball and rugby) and toward the end of a period or game (Ekstrand et al., 2011; Lundblad et al., 2013; Woods et al., 2004). This has been noted particularly for knee ligament (Lundblad et al., 2013) and thigh injuries (Ekstrand et al., 2011), suggesting that fatigue is a risk factor. In laboratory-based studies, researchers also have demonstrated that fatigue results in unwanted changes in movement technique, such as increased knee abduction (Chappell et al., 2005), which is a significant component of the ACL injury mechanism (Hewett et al., 2005; Koga et al., 2010). Fatigue may contribute to detrimental changes in movement technique due to decreased efficiency of muscle spindle afferent information (Gribble et al., 2012), delayed muscle contraction (Hassanlouei et al., 2012), decreased muscle-torque generation (Gribble et al., 2012; Hassanlouei et al., 2012) and central nervous system changes (Gandevia, 2001).

Given that injury rates increase when athletes are fatigued (Ekstrand et al., 2011; Lundblad et al., 2013; Woods et al., 2004) and deficits in postural control are risk factors for sustaining lower limb injuries (McGuine et al., 2000; Plisky et al., 2006; Wang et al., 2006), fatigue-induced postural deficits may be expected to contribute to the incidence of injury. However, this has not been examined prospectively and whereas several investigators have demonstrated that fatigue negatively affects dynamic postural control (Reimer and Wikstrom, 2010; Steib, Hentschke et al., 2013; Wright et al., 2013), others have not (Johnston et al., 1998; Wright et al., 2013; Zech et al., 2012). These contrasting findings may be due to methodological variations and sex-related difference. For example, researchers (Gribble et al., 2004; Gribble et al., 2007) have found that functional exercise protocols cause greater deficits than isokinetic protocols do. In addition, whereas fatigue affects dynamic postural control less in women (Gribble et al., 2009), results from males and females have been combined in several studies (Gribble et al., 2004; Johnston et al., 1998; Steib, Zech et al., 2013).

Despite investigations into the effects of fatigue on dynamic postural control using continuous whole body fatiguing exercise (Steib et al., 2013; Steib et al., 2013; Wikstrom et al., 2004; Wright et al., 2013; Zech et al., 2012) and localised muscle fatiguing exercise (Gribble and Hertel, 2004; Gribble et al., 2009; Reimer and Wikstrom, 2010; Wikstrom et al., 2004), no researchers have investigated the effects of effects of high intensity, intermittent exercise on dynamic postural control. This topic needs to be examined given the high injury incidence during sports that contain regular bouts of high intensity, intermittent exercise when participants are fatigued (Ekstrand et al., 2011; Lundblad et al., 2013; Woods et al., 2004).

Walden et al (Walden, Hagglund et al. 2011) reported a sex disparity in the incidence of certain lower limb injuries, with higher incidences of ACL injuries in women. Researchers (Hewett et al. 2005, McLean, Samorezov 2009) have proposed that it is due to neuromuscular and biomechanical differences. Despite the sex disparity in ACL injuries (Walden, Hagglund et al. 2011) and the relationship between lower SEBT scores and injury (Plisky et al. 2006), the only authors (Gribble et al. 2009) who investigated the effect of fatigue and sex on dynamic postural control found that women had better scores of dynamic postural control and were less negatively affected than men. It is unclear if similar findings would be reproduced when male and female athletes are fatigued by high intensity, intermittent exercise.

Therefore, the primary purpose of our study was to investigate the reliability of the SEBT. Our main purpose was to compare the effect of a high-intensity, intermittent exercise protocol (HIIP) on SEBT scores between men and women. Our secondary aims were to investigate the effect of the HIIP on SEBT scores across both sexes and the effect of sex on SEBT scores. We hypothesized that SEBT test retest reliability would be high and that women would be less negatively affected by the HIIP than men. We also hypothesized that the HIIP would negatively affect SEBT scores in both men and women and that women would have higher SEBT scores

3.3 Methods

3.3.1 Participants

A cohort of 40 male and female university athletes volunteered to participate in the study (Table 3.1). Inclusion criteria required participants to be free from all lower extremity or head injuries within the past 6 months before the study, to not be involved in a balance training programme, and to be generally healthy. Participants were excluded if they had ankle or knee joint instability, had any neurologic or central nervous system deficits, or were taking any medication that may have affected their balance. All participants provided written informed consent, and the study was approved by the Research Ethics Committee of Dublin City University.

Table 3.1 Participant characteristics

	Males (n=20)	**Females (n=20)**
Age (years)	20.83 ±1.5	20.45 ±1.34
Height (cm)	179.24 ± 7.94	166.08 ± 5.83
Leg length (cm)	94.58 ± 6.05	87.95 ± 3.91
Mass (kg)	77.67 ±10.82	63.02 ± 6.67
Primary sport		
Gaelic Football	12	10
Hurling/Camoige	5	6
Soccer	3	4

3.3.2 Procedure

Participants were required to attend 3 separate sessions. They underwent a familiarization session, and after a 2-day interval, they underwent a baseline measurement session that was used to determine the repeatability of the SEBT protocol. Participants returned 5 to 7 days later for assessment of the SEBT pre-HIIP and post-HIIP.

3.3.2.1 Star excursion balance test

The SEBT was performed according to Gribble et al. (2009). Each participant stood with hands on hips and their dominant leg in the centre of a grid, placed on the floor, consisting of 8 lines extending at 45° increments from the centre. The dominant leg was defined as the leg with which they would kick a ball (Gribble et al., 2007). The navicular tuberosity was marked while each participant was weight bearing and positioned over the centre of the

SEBT. The hallux was positioned over the anterior progression line. Participants were instructed to maintain a single-legged stance while reaching with the free limb as far as possible along a given direction, to lightly touch the farthest point possible, and to return to bilateral stance while maintaining their equilibrium.

Reach directions were completed in a random order 3 times in each direction and recorded manually by the tester (E. W., A. B., or E. W.). Trials were discarded and repeated if the participant used the reaching limb for a substantial amount of support on touching the tape, removed the foot from the designated center of the grid, lifted the heel of the stance limb, took hands off hips, or was unable to maintain balance. As Plisky et al (Plisky et al., 2006) described, participants had a minimum of 6 practice trials in each direction using both limbs and then sat quietly for 5 minutes before data collection on each testing day. Measures of maximal SEBT reach distances were normalized for limb length using the following formula: normalized maximal reach distance = reach distance (cm)/leg length (cm) x 100 (Gribble et al., 2009). The overall average SEBT score was calculated by averaging the normalized maximal SEBT in each direction. Lower limb length was measured as the distance from the anterior-superior iliac spine to the most distal point of the medial malleolus (Gribble et al., 2009).

3.3.2.2 High intensity, intermittent exercise protocol

After a warm-up (Whyte et al., 2010) participants completed the HIIP with maximal effort (Figure 3.1). The HIIP was designed to mimic periods of high-intensity, intermittent activity leading to temporary fatigue (Krustrup et al., 2006; Mohr et al., 2003). Di Mascio and Bradley (Di Mascio and Bradley, 2013) reported that during such periods, soccer players ran for an average of 6.7 6 1.8 m and recovered for an average of 30 seconds between repeated periods. During the HIIP, participants sprinted forward 5 m, cut at a 90° angle, sprinted forward another 5 m, and backpedaled 5 m. This activity was repeated 4 times before participants completed a series of hurdle activities. First, they performed 2-legged jumps over 5 hurdles that were 30 cm high, turned, and repeated the jumps. Second, they performed side-stepping exercises over the 5 hurdles. Third, they completed four 5-m lateral shuffles. Circuit time was recorded using infrared timing gates (model TC; Brower Timing Systems, Draper, UT). After completing a circuit, participants rested for 30 seconds before repeating the circuit. The HIIP was discontinued when participants reported a score of 18 on the Borg rating of perceived exertion (RPE) scale, which ranges from 6 to 20 (Borg, 1970). Heart rate was monitored throughout the HIIP using a heart-rate monitor (model FT1; Polar Electro Inc, Lake Success, NY). Dynamic postural-control assessments using the SEBT commenced within 15 seconds and were completed within 3 minutes. To minimize the effects of recovery, practice trials were conducted pre-fatigue protocol only (Gribble et al., 2009; Steib et al., 2013; Zech et al., 2012).

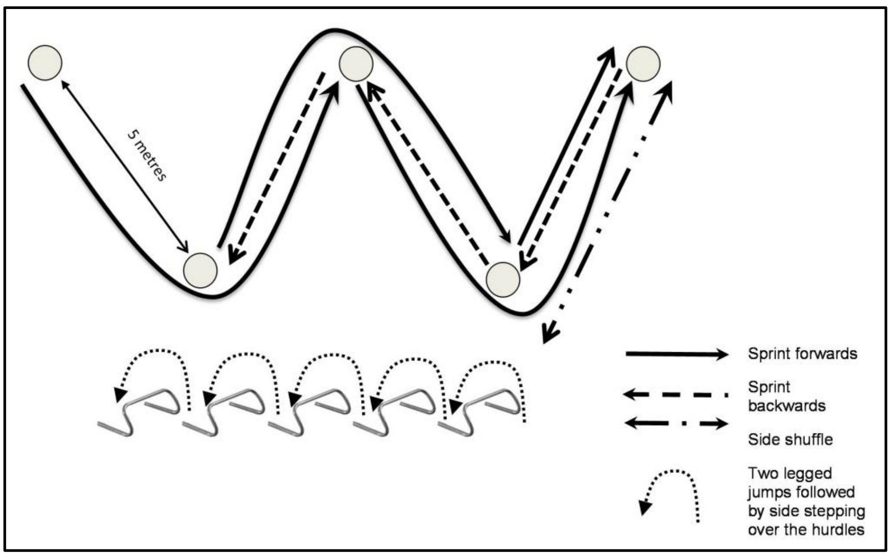

Figure 3.1 The high intensity, intermittent exercise protocol

3.3.2.3 Statistical analysis

Test-Retest Reliability. Intraclass correlation coefficients (ICCs) determined the repeatability of normalized maximal SEBT scores taken at baseline and pre-HIIP. Standard error of measurement (SEM) was used to assess intersession variability (ie, the degree of variation between repeated assessments in the same group). The SEM was calculated multiplying the standard deviation of the test scores (S_T) by the square root of one less the reliability coefficient (r_{xx}) ($SEM_x = S_T \sqrt{[1 - r_{xx}]}$).

Physiologic Effects of the HIIP. Paired-samples *t* tests compared the first and final HIIP circuit-completion times and resting heart rate with post-HIIP heart rate.

Effect of HIIP and Sex on Dynamic Postural Control. The main aim of the study was investigated using a mixed between- and within-subjects analysis of variance to assess time (pre-HIIP, post-HIPP) x sex interaction effect (ie, to determine if the difference in SEBT scores between men and women varied post-HIIP). We subsequently analyzed the effect of time and sex on normalized maximal SEBT scores. The dependent variables analyzed were the normalized maximal SEBT scores in 8 directions and the overall average normalized SEBT score. Post hoc Bonferroni analyses were conducted to correct for multiple comparisons. To determine the magnitude of any effect, effect sizes (η^2) were calculated and ranked using the Cohen d classification (0.01 = small effect, 0.06 = medium effect, 0.14 = large effect) (Pallant, 2010). We used SPSS software (version 17.0; SPSS Inc, Chicago, IL) for all analyses and set the α level at < .05.

3.4 Results

3.4.1 Test-retest Reliability

The range of ICC values (0.77–0.99) for the 8 normalized maximal SEBT scores and the overall average SEBT score demonstrated strong test-retest reliability per the Cohen d classification. The SEM ranged from 0.62 to 2.60 for the different SEBT measurements (Table 3.2).

3.4.2 Time by Gender Interaction Effect

We found a time 3 sex interaction effect, with small to large effect sizes for each direction of the 8 normalized maximal SEBT scores and the overall average SEBT score (Wilks Λ =

0.71; $F_{8,69} = 3.5$; P range, <.001–.04; η^2 range, 0.057–0.219), indicating that the negative effects of the HIIP on dynamic postural control were less in women (**Table 3.3**).

3.4.3 Main Effects for Time and Gender

A main effect of time indicated that the HIIP had a detrimental effect on the normalized maximal SEBT scores in men and women for the 8 directions and the overall average SEBT score, with large effect sizes (Wilks $\Lambda = 0.28$; $F_{8,69} = 22.39$; $P < .001$; η^2 range, 0.324–0.695; Table 3.3). We found a main effect for sex with small to large effect sizes (Wilks $\Lambda = 0.91$; $F_{8,69} = 0.84$; P range <.001–.008; η^2 range, 0.088–0.381), with women demonstrating better normalized maximal SEBT scores than men in all directions except anterolateral (Table 3.4). The relationship between the SEM calculated for the SEBT and the fatigue-induced decline in SEBT scores is displayed in Table 3.4.

Table 3.2 Intraclass correlations between baseline and pre-exercise normalised star excursion balance test maximal reach distances (n=40)

SEBT Direction	Average SEBT†	Anterior	Antero-medial	Medial	Postero-medial	Posterior	Postero-lateral	Lateral	Antero-lateral
Intra Class Correlation	0.98	0.90	0.77	0.77	0.99	0.99	0.97	0.99	0.82
(95% confidence interval)	(0.96-0.99)	(0.81-0.95)	(0.60-0.88)	(0.56-0.88)	(0.99-1.0)	(0.98-0.99)	(0.95-0.99)	(0.98-0.99)	(0.62-0.91)
Standard Error of Measurement	0.62	1.65	2.36	2.60	0.62	0.65	1.14	0.81	2.13

Table 3.3 Pre-fatigue and post-fatigue values and pre-post differences for dependent measures, mean ±SD, and effect sizes of main and interaction effects

	Males			Females			Exercise Main Effect [a]		Gender Main Effect [b]		Interaction Effect [c]	
	Pre HIIP	Post HIIP	Percentage change	Pre HIIP	Post HIIP	Percentage change	P	ES	P	ES	P	ES
SEBT[d]	83.73±3.75	78.29±5.36	6.49	87.54±4.04	84.67±4.32	3.26	<0.001	0.695	<0.001	0.281	0.007	0.179
Anterior	84.84±4.83	80.33±6.62	5.31	86.96±5.09	84.73±6.22	2.56	<0.001	0.554	0.008	0.088	<0.001	0.152
Antero-medial	84.71±4.10	80.92±5.69	4.48	87.35±4.93	85.82±4.75	1.75	<0.001	0.337	<0.001	0.152	0.008	0.090
Medial	84.12±4.12	78.89±5.85	6.23	88.87±5.50	87.07±5.59	2.03	<0.001	0.461	<0.001	0.311	<0.001	0.219
Postero-medial	86.69±5.30	80.74±5.49	6.87	92.70±5.43	89.50±4.90	3.45	<0.001	0.540	<0.001	0.381	0.003	0.110
Posterior	88.00±6.33	81.25±7.23	7.67	94.52±5.00	89.55±5.47	5.26	<0.001	0.648	<0.001	0.305	0.001	0.133
Postero-lateral	84.63±5.65	77.81±6.13	8.05	90.19±6.44	86.20±5.89	4.43	<0.001	0.625	<0.001	0.274	0.003	0.111
Lateral	77.71±7.37	70.94±7.72	8.71	82.24±8.86	78.03±9.09	5.11	<0.001	0.493	0.001	0.143	0.035	0.057
Antero-lateral	79.12±4.41	75.49±5.83	4.59	77.49±5.67	76.47±5.31	1.32	<0.001	0.324	0.503	0.006	0.002	0.118

a. Indicates pre-HIIP post-HIIP difference
b Indicates gender differences
c Indicates group differences for prefatigue–postfatigue changes. Positive effect sizes indicate greater changes in the male group.
d Mean of all 8 directions normalized to leg length (reach distance in cm * 100/leg length in cm).
ES – Effect size (eta-squared values)

Table 3.4 Comparison of Standard Error of Measurement and Fatigue-Induced Decline in Star Excursion Balance Test Scores in Men and Women

Star Excursion Balance Test	Standard Error of Measurement	Star Excursion Balance Test Score Decline Induced by High intensity, Intermittent Exercise Protocol	
		Men	Women
Average score	0.62	5.44	2.87
Direction			
Anterior	1.65	4.51	2.23
Anteromedial	2.36	3.79	1.53
Medial	2.60	5.23	1.80
Posteromedial	0.62	5.95	3.20
Posterior	0.65	6.75	4.97
Posterolateral	1.14	6.82	3.99
Lateral	0.81	6.77	4.21
Anterolateral	2.13	3.63	1.02

3.4.4 Effect of the High Intensity, Intermittent Exercise Protocol on Lap Completion Times and Heart Rate

On average, participants completed 6.58 (± 0.81) circuits of the HIIP before reporting a score of 18 on the Borg RPE scale, with an average heart rate at completion of 189.95 ± 5.58 beats/min versus 56.20 ± 9.40 beats/min at rest (Table 3.5). Circuit-completion times increased from the initial (54.86 ± 4.48 seconds) to the final (57.42 ± 4.56 seconds) circuit ($P < .001$; Table 3.5).

Table 3.5 Markers of exertion

	Males (n=20)	Females (n=20)	Overall (n=40)
Resting hear rate, beats/min			
Maximal heart rate, beats/min	190.33±5.54	189.59±6.31	189.95±5.58
% HR max at final lap	95.56±2.78	95.01±3.16	95.28±2.95
Number of circuits	6.35±0.67	6.8±0.89	6.58±0.81
Time of circuit 1 (s)	56.59± 3.15	53.13± 5.02	54.86± 4.48
Time of final circuit (s)	59.04±3.94*	55.8±4.65*	57.42±4.56*
	($p < 0.001$)	($p < 0.001$)	($p < 0.001$)

* = significant difference between time of time of circuit 1 and final circuit (P <0.01)

Table 3.6 Comparison of standard error of measurement and fatigue induced decline in star excursion balance test scores in males and females

	Standard Error of Measurement	HIIP Induced SEBT Score Decline	
		Males	Females
Average SEBT Score	0.62	5.44	2.87
Anterior	1.65	4.51	2.23
Antero-medial	2.36	3.79	1.53
Medial	2.6	5.23	1.8
Postero-medial	0.62	5.95	3.2
Posterior	0.65	6.75	4.97
Postero-lateral	1.14	6.82	3.99
Lateral	0.81	6.77	4.21
Antero-lateral	2.13	3.63	1.02

3.5 Discussion

The results of our study demonstrate HIIP-induced decrements in SEBT scores in both men and women, with women less negatively affected than men. Women also had higher SEBT scores pre-HIIP and post-HIIP. Confidence can be placed in these findings for 2 reasons. First, strong test-retest reliability, similar to that previously reported (Gribble and Hertel, 2003), was shown between baseline and pre-HIIP SEBT scores. Second, all SEBT scores in men and 6 of 9 SEBT scores in women showed an HIIP-induced decline greater than the respective SEM (Table 3.4), indicating that the HIIP resulted in reductions in SEBT scores that were greater than intersession variability in most measurements.

The HIIP physiologically stressed participants, with high RPEs and heart rates similar to the 187 ± 9 beats/min reported in soccer players during match play (Table 3.5) (Mohr et al., 2003). It required participants to run for an average distance of 460.6 m over 8 minutes, 47.8 seconds, on average. This approximates to 261.7 m in 5 minutes, similar to the typical peak distance reported for soccer players over 5 minutes (219 ± 8 m) (Mohr et al., 2003). The increased circuit-completion times indicated that the HIIP induced fatigue, which may be due to a number of potential central and peripheral mechanisms. Centrally, the HIIP can lead to altered cardiovascular, respiratory, thermoregulatory, and muscular afferent input that, when combined with high levels of perceived exertion, can negatively affect cortical motor drive and result in the inhibition of the lower motor neuron at the spinal level (Gandevia, 2001; Taylor and Gandevia, 2008), ultimately decreasing muscle-force output and contributing to the observed increased circuit time. Peripherally, the HIIP may result in several changes, such as decreased neuromuscular transmission (Reid et al., 1999;Wu and Betz, 1998) and elevated inorganic phosphate levels (Allen et al., 2008), that can interfere with cross-bridge formation. It can also lead to hyperkalaemia (Allen et al., 2008), acidosis (Boyas and Guevel, 2011; Fitts, 2008) and the presence of reactive oxygen species (Allen et al., 2008; Knicker et al., 2011); these changes can impair contractile protein activity and stimulate muscle afferents, reducing central motor drive. The combined central and peripheral effects can result in reduced muscle-force output that, when combined with fatigue-induced decreases in sensorimotor afferent information (Gribble et al., 2009) and delayed muscle contraction (Hassanlouei et al., 2012) may lead to technique changes (Chappell et al., 2005;McLean et al., 2007) and dynamic postural-control deficits (Gribble et al., 2009; Steib et al., 2013; Wikstrom et al., 2004; Wright et al., 2013) in exercising athletes. This could increase the susceptibility to lower limb injuries and explain, at least in part, the observed relationship between fatigue and injury in sports with intermittent bouts of high-intensity exercise (Ekstrand et al., 2011; Lundblad et al., 2013; Woods et al., 2004).

Our results support previous research in which investigators demonstrated that a fatiguing protocol led to worse dynamic postural control (Gribble et al., 2009; Steib et al., 2013; Wikstrom et al., 2004; Wright et al., 2013). Given that the exercise performed dictates the resultant fatigue (Reimer and Wikstrom, 2010;Wright et al., 2013), it is not surprising that our results conflict with other studies (Johnston et al., 1998; Wright et al., 2013; Zech et al., 2012) in which researchers used different protocols including closed kinetic chain dynamometry (Johnston et al., 1998), cycling (Wright et al., 2013), continuous treadmill running (Steib et al., 2013;Wright et al., 2013;Zech et al., 2012), and step up with resistance (Zech et al., 2012). These protocols may have fatiguing effects that are very different from the HIIP. Indeed, the percentage changes in normalized maximal SEBT scores were generally greater in our study than in previous studies (Gribble and Hertel, 2004; Gribble et al., 2009), suggesting that the HIIP-induced fatigue resulted in lower normalized maximal SEBT scores than did fatigue induced by continuous treadmill running (Steib et al., 2013; Zech et al., 2012) and isokinetic and lunging (Gribble and Hertel, 2004;Gribble et al., 2009) and step-up-with-resistance (Zech et al., 2012) protocols. Even studies in which researchers have investigated the effect of functional whole-body fatiguing protocols on dynamic postural control have demonstrated conflicting results, with several showing detrimental effects (Shaw et al., 2008;Steib et al., 2013;Wikstrom et al., 2004;Wright et al., 2013) or no effects (Wright et al., 2013;Zech et al., 2012). Steib et al (Steib et al., 2013) found that a treadmill fatiguing protocol involving an RPE similar to the one we used to terminate the protocol (17 and 18, respectively) led to a decrease in normalized maximal SEBT scores for healthy male athletes in the anterior (0.8), medial (1.44), lateral (1.65), and posterior (0.55) directions. These scores are considerably lower than the HIIP-induced decreases we observed (4.51, 5.23, 6.77, and 6.75, respectively), supporting the concept that fatigue and its effects depend on the exercise (Reimer and Wikstrom, 2010; Wright et al., 2013). In general, protocols involving continuous running with multiple changes of direction (Shaw et al., 2008;Wikstrom et al., 2004) have resulted in decrements in dynamic postural control; however, protocols involving continuous treadmill running only (Steib et al., 2013; Wright et al., 2013; Zech et al., 2012) have produced mixed results, and protocols involving cycling have produced no effects (Wright et al., 2013; Zech et al., 2012). A potential limitation in studies using running protocols with multiple changes of direction (Shaw et al., 2008; Wikstrom et al., 2004) is that the fatigued state was determined when the lap-completion time increased by 50%. Given that Krustrup et al (Krustrup et al., 2010) reported sprint times increased by only 4% after a soccer match, the severity of this induced fatigue was greater than that observed in some sports. The HIIP that we used, however, increased circuit-completion time by 4.7% on average.

Dynamic postural control requires the coordination of the neuromuscular and somatosensory systems to process sensory information and react accordingly (Patla et al., 1999). Fatigue has been reported to negatively affect joint proprioception (Forestier et al., 2002) because of decreased muscle-spindle activity (Gandevia, 2001) and increased joint laxity (Wojtys et al., 1996) which may disturb the somatosensory input of ligament mechanoreceptors. In addition, Hassanlouei et al., (Hassanlouei et al., 2012) observed that fatigue delays muscle-contraction onset and decreases activation. These results may have the combined effect of reducing the efficiency of neuromuscular and somatosensory coordination and impairing postural control. In a prospective study, Plisky et al (Plisky et al., 2006) demonstrated that deficits in postural control predict lower limb injury. Their results, in conjunction with ours, suggest that HIIP-induced fatigue may increase susceptibility to lower limb injury. Prospective investigations should be conducted to determine if a link exists between injury and fatigue-induced decrements in postural control.

Our results support the findings of Gribble et al (Gribble et al., 2009) that fatigue negatively affects women less than men. This finding may partially result from the observed sex differences in muscle fatigability, purportedly due to several interrelated processes (Hunter, 2009). Researchers have demonstrated that men have a lower rate of oxidative muscle metabolism than women (Hunter et al., 2006) and a strength-dependent reduction in muscle perfusion (Hunter, Schletty et al., 2006). These characteristics can lead to an accumulation of muscle metabolites and subsequent greater stimulation of inhibitory afferents in men (Hunter, 2009), resulting in a decreased motor response and evidence of neuromuscular fatigue (Gandevia, 2001). Given that neuromuscular control is an essential element of dynamic postural control (Patla et al., 1999), a reduction therein may explain, in part, the observation that fatiguing exercise has a less negative effect on SEBT scores in women. In conjunction with the relationship between lower SEBT scores and a higher incidence of lower limb injury (Plisky et al., 2006) this suggests that women are at lower risk of sustaining lower limb injuries post- HIIP. However, to our knowledge, no one has investigated the relationship between fatigue-induced decrements in SEBT scores and injury incidence.

Our findings are also consistent with those of Gribble et al (Gribble et al., 2009) in that women had better SEBT scores than men in fatigued and unfatigued conditions. We observed that in all directions except the anterolateral direction, women had prefatigue SEBT scores ranging from 2.12 to 6.52 higher than men. In the fatigued condition, women had SEBT scores ranging from 0.98 to 8.76 points higher than men in all directions. Women demonstrated greater knee and hip flexion during the SEBT than their male counterparts (Gribble et al., 2009), allowing women to lower their centers of gravity and achieve better SEBT scores (Gribble et al., 2009; Robinson and Gribble, 2008). Given that lower postural-control scores (Plisky et al., 2006) and lower SEBT scores in particular (Plisky et al., 2006) predict a higher incidence of lower limb injury, this implies that the women in our study were at a lower risk of sustaining lower limb injuries than the men and seems to contradict epidemiologic data that demonstrate women have a higher incidence of certain lower limb injuries, such as ACL injuries (Walden et al., 2011). However, investigators (Gribble et al., 2009; Robinson and Gribble, 2008) studying differences in SEBT technique have examined kinematics only in the sagittal plane. The increased knee flexion and hip abduction during the SEBT possibly were combined with hip and knee transverse- and frontal-plane motions that may predispose women to injury. This may be especially relevant in fatigued conditions, because authors (Chappell et al., 2005; McLean et al., 2007) of laboratory-based studies have observed that fatigue results in altered movement patterns in the frontal, transverse, and sagittal planes, which may increase the loading of the ACL in female compared with male athletes. Therefore, the observed higher injury incidence of certain lower limb injuries in women may be due to biomechanical differences in technique (eg, landing (McLean et al., 2007)) in fatigued and unfatigued conditions rather than changes in dynamic postural control as measured by the SEBT. Kinematic studies of sex differences in SEBT techniques in 3 planes of motion in both the fatigued and unfatigued states would be valuable additions to our understanding of sex differences in the SEBT and their potential relationship with injury.

3.5.1 Limitations

Our study had limitations. No criterion standard measurement of dynamic postural control exists (Zech et al., 2012) so it has been assessed in a number of tasks, including the SEBT (Gribble and Hertel, 2004; Gribble et al., 2009) time to stabilization (Shaw et al., 2008;Wikstrom et al., 2004), center-of-pressure sway (Steib et al., 2013; Zech et al., 2012) and Biodex Balance System (Reimer and Wikstrom, 2010;Wright et al., 2013). Given the differences in these tasks, our findings should be related to sporting movements aligned with the SEBT, such as kicking and cutting.

The HIIP developed for our study to mimic the high-intensity, intermittent activities common in field sports has not been investigated, making comparisons with previous research difficult. However, in a recent review of the manifestations of fatigue in sport, Knicker et al (Knicker et al., 2011) specifically recommended examining temporary fatigue resulting from high-intensity, intermittent activity.

The determinant of the fatigued condition (18 on the Borg RPE scale) is a subjective measurement of exertion and may not be consistent across participants. However, it has been used in previous studies of the effect of fatigue on postural control (Steib et al., 2013; Zech et al., 2012) and resulted in similarly elevated heart rates and distances covered during bouts of high-intensity, intermittent activity in soccer (Mohr et al., 2003).

Finally, the HIIP-induced decline in SEBT scores for women in the anteromedial, medial, and anterolateral directions was lower than the SEM for these directions (Table 3.4). This indicates that the intersession variability, rather than the effect of HIIP-induced fatigue, may account for the decline in scores. The relatively lower ICC values in these directions may partially explain the higher SEM values. The reason these values were relatively lower is unclear and may be that, in the anterior- and medial-reach directions, maintaining a level pelvis is especially challenging (Norris and Trudelle-Jackson, 2011).

As there is no gold standard measurement of dynamic postural control (Zech et al., 2012), it has been assessed in a number of tasks including the SEBT (Gribble and Hertel, 2004; Gribble et al., 2009), time to stabilisation (Shaw et al., 2008; Wikstrom et al., 2004), centre of pressure sway (Steib et al., 2013; Zech et al., 2012) and the biodex stability system (Reimer and Wikstrom, 2010; Wright et al., 2013). Given the differences in these tasks, the findings of the present study should be related to sporting movements aligned with the SEBT such as kicking and cutting.

The HIIP developed for the current study to mimic the high intensity, intermittent activities common in field sports has not been researched previously making comparisons with previous research difficult. However, in a recent review on the manifestations of fatigue in sport, Knicker et al. (Knicker et al., 2011) specifically recommended investigations into temporary fatigue resulting from high intensity, intermittent activity. Also, the HIIP used in the current study mimics the high intensity, intermittent activities during certain sports

Also, the determinant of the fatigued condition (18 on Borg’s RPE scale) is a subjective measurement of exertion and may not be consistent across participants. However, it has been used in previous studies investigating the effect of fatigue on postural control (Steib et al., 2013; Zech et al., 2012), and resulted in similar elevated heart rates and distances covered seen during bouts of high intensity, intermittent activity in soccer (Mohr et al., 2003).

3.6 Conclusion

The HIIP negatively affected dynamic postural control as assessed by the SEBT in athletes. Women were affected less negatively by the HIIP and displayed better levels of dynamic postural control than men. Given that many field sports consist of high-intensity, intermittent exercise, our results suggest that athletes involved in these sports should perform postural-control programs after such exercise and aim to increase their abilities to reduce the extent and effect of fatigue.

3.7 Link to Chapter 4

The results of study 1 have a number of important implications in terms of biomechanical risk factors for ACL injuries. Firstly, the HIIP leads to physiological responses similar to those observed in soccer and has a detrimental effect on dynamic postural control as measured by the SEBT. The HIIP had a fatiguing effect as it resulted in a decrease in performance as assessed by lap completion times. Another important finding was that males have lower measures of dynamic balance pre-HIIP and are affected to a greater extent by the HIIP. This finding, in conjunction with the fact that deficits in dynamic balance predict lower limb injuries, suggests that fatigue induced by high intensity exercise may be a factor in sustaining lower limb injuries, especially in males. In order to progress this further and relate it to ACL injuries specifically, the effect of HIIP-induced fatigue on the biomechanics of tasks associated with ACL injuries, such as the vertical drop jump, should be investigated.

The vertical drop jump is frequently used as a screening tool to identify athletes at risk of ACL injury. It is used because it is a highly reliable test that mimics a jump landing, which is a common mechanism of ACL injuries. Critically, biomechanics of the vertical drop jump have been found to be predictive of ACL injuries in some studies. As the HIIP has been shown to negatively affect dynamic balance and deficits in dynamic balance predict lower limb injuries, study 2 (chapter 4) will investigate the effect of the HIIP on biomechanical risk factors for ACL injuries during the vertical drop jump.

Chapter 4 The Effects of Limb Dominance and a Short Term, High Intensity Exercise Protocol on Both Landings of the Vertical Drop Jump: Implications for the Vertical Drop Jump as a Screening Tool

Study 2

Study 2 “The effects of limb dominance and a short term, high intensity exercise protocol on both landings of the Vertical Drop Jump: implications for the Vertical Drop Jump as a screening tool.”

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4.1 Abstract

The effectiveness of vertical drop jumps (VDJs) to screen for noncontact ACL injuries is unclear. This may be contributed to by discrete point analysis, which does not evaluate patterns of movement. Also, limited research exists on the second landing of VDJs, potential lower limb performance asymmetries and the effect of fatigue. Statistical parametric mapping investigated the main effects of landing, limb dominance and a high intensity, intermittent exercise protocol (HIIP) on VDJ biomechanics. Twenty-two male athletes (21.9 ± 1.1 years, 180.5 ± 5.5 cm, 79.4 ± 7.8 kg) performed VDJs pre- and post-HIIP. Repeated-measures ANOVA identified pattern differences during the eccentric phases of the first and second landings bilaterally. The first landing displayed greater knee flexor ($\eta^2 = 0.165$), external rotator ($\eta^2=0.113$) and valgus ($\eta^2 = 0.126$) moments and greater hip ($\eta^2 = 0.062$) and knee ($\eta^2 = 0.080$) flexion. The dominant limb generated greater knee flexor ($\eta^2 = 0.062$), external rotator ($\eta^2 = 0.110$), and valgus ($\eta^2 = 0.065$) moments. The HIIP only had one effect, increased thoracic flexion relative to the pelvis ($\eta^2 = 0.088$). Finally, the dominant limb demonstrated greater knee extensor moments during the second landing ($\eta^2 = 0.100$). ACL injury risk factors were present in both landings of VDJs with

the dominant limb at potentially greater injury risk. Therefore, VDJ screenings should analyse both landings bilaterally

4.2 Introduction

Anterior cruciate ligament (ACL) injuries are generally non-contact in nature and occur during the deceleration phase of high-risk sporting manoeuvres, such as landing from a jump (Koga et al., 2010; Krosshaug et al., 2007; Walden et al., 2015). Given that they lead to considerable detrimental consequences, such as high medical costs (Wojtys and Brower, 2010) and early osteoarthritis (Oiestad et al., 2009), there has been a significant emphasis on developing ACL injury prevention programmes. Prevention requires the identification, understanding and modification of risk factors (Finch, 2006) with much research focussing on investigating modifiable biomechanical risk factors. Cadaveric studies demonstrate that combined (internal) knee extensor (Weinhandl et al., 2013), adductor and especially, external rotator moments (Markolf et al., 1995; Oh et al., 2012; Shin et al., 2011) increase ACL strain. Proposed biomechanical risk factors for non-contact ACL injury include reduced trunk kinematic control (Hewett et al., 2009; Hewett et al., 2010; Zazulak et al., 2007), less hip and knee flexion angles (Leppanen et al., 2017), larger ground reaction forces (GRFs) (Leppanen et al., 2017), greater (internal) knee adductor moment and valgus angle (Hewett et al., 2005), greater hip adduction and internal rotation (Alentorn-Geli et al., 2009) and asymmetrical lower limb biomechanics (Hewett et al., 2010; Paterno et al., 2010). Also, fatiguing protocols have been found to negatively affect the biomechanics of the vertical drop jump (VDJ) and increase an athlete's predisposition to ACL injury (McLean et al., 2007; Smith et al., 2009).

The vertical drop jump (VDJ) is a commonly used screening and risk factor identification tool (Bates et al., 2013b; Moran and Marshall, 2006). It consists of a drop from a height, a first landing followed by a maximal vertical jump and a second landing. It mimics a common mechanism of ACL injury (Bates et al., 2013b; Hewett et al., 2005) and has the benefit of high levels of reliability (Malfait et al., 2014). Nevertheless, the ability of the

VDJ to prospectively predict ACL injuries is inconsistent (Hewett et al., 2005; Krosshaug et al., 2016; Leppanen et al., 2017) leading some to conclude that it is a poor screening tool for ACL injuries (Krosshaug et al., 2016). This inconsistency may be contributed to by the traditional discrete point analysis (DPA). DPA, used in the prospective studies above, selects predetermined, discrete points to represent the whole data signal. This may be insufficient to represent the whole kinematic/kinetic signal (Shultz et al., 2015) as less than 5% of the movement is captured and important information may be discarded (Federolf et al., 2013; Richter et al., 2014b). In addition, and most importantly, analysing discrete points such as peak vertical ground reaction force (vGRF) can result in the comparison of discrete points during different functional elements of the task (Richter et al., 2014a). An alternative method of analysis is statistical parametric mapping (SPM). It is more effective at identifying differences in patterns of movement (Shultz et al., 2015), a key aspect of the ACL injury mechanism (Krosshaug et al., 2016). SPM analyses the original vectors rather than predetermined discrete points. It takes into account the influence of time on vectors and the influence different components of a vector have on other components, for example the effect knee rotation and knee flexion angles have on knee abduction angles (Pataky et al., 2013). For these reasons, SPM is more effective at identifying task features (e.g. knee flexion angle and GRF) during activities such as the VDJ (Pataky et al., 2013; Richter et al., 2014b). Therefore, SPM may add to the existing body of knowledge from DPA studies enhancing our understanding of the biomechanics of the VDJ and its usefulness as a screening tool.

Despite the potential for VDJs to identify risk factors for noncontact ACL injuries, there has been a lack of research on the second landing of the VDJ, potential performance asymmetries in the first and second landings and the effect of fatigue thereon. Prospective screening studies analyse the first landing of the VDJ only (Hewett et al., 2005; Krosshaug et al., 2016; Padua et al., 2015) despite video analysis demonstrating that the second landing is a common mechanism of non-contact ACL injuries in males (Cochrane et al., 2007; Walden et al., 2015). As the first and second landing of the VDJ has only been examined in one cohort of (female) participants using DPA (Bates et al., 2013a; Bates et al., 2013b; Bates et al., 2013c) our understanding of the ability of the VDJ to identify potential risk factors for ACL injury in the first and second landing is limited. Furthermore, it is common practice to analyse one limb during the VDJ screening process and infer injury risk to both limbs (Padua et al., 2015). This is despite the fact that leg to leg asymmetries have been found to predict ACL re-injury (Paterno et al., 2010), are theorised to predispose an athlete to first time ACL injury (Hewett et al., 2010) and 74% of non-contact ACL injuries occur on the dominant leg in males (Brophy et al., 2010). Finally, fatigue, which occurs temporarily after intense activity, has also been proposed to be a risk factor for ACL injury (McLean and Samorezov, 2009; Shultz et al., 2015). Although studies have examined the effects of different fatiguing protocols on the VDJ, such as general running (Moran and Marshall, 2006) and local muscular fatiguing protocols (Haddas et al., 2016; Weinhandl et al., 2011), to the authors' knowledge no studies have investigated the effects of a high intensity, intermittent exercise protocol (HIIP) on the biomechanics of the VDJ. Critically, the effects of these three factors on the biomechanics of the VDJ have not been examined using SPM which may limit our understanding of the VDJ as a screening tool for ACL injuries.

The aims of this study were to investigate the effects of: (1) landing (first versus second landing), (2) limb dominance, and (3) a HIIP (pre-HIIP versus post-HIIP) on the pattern of trunk and lower limb biomechanics during the VDJ. It was hypothesised that there would be significant main effects for landing, limb dominance and the HIIP. Specifically, it was hypothesised that detrimental biomechanics in terms of ACL injury risk would be 1. greater in the second landing of the VDJ, 2. greater in the dominant limb and 3. greater post-HIIP. We also hypothesised that interaction effects would demonstrate that the HIIP would have a greater effect on the dominant limb during the first landing .

4.3 Methods

4.3.1 Experimental Approach

The biomechanics of the eccentric phase during the first and second landings of the VDJ were collected bilaterally, pre- and post-HIIP, using a Vicon 3-D motion analysis system and two force plates. Participants were required to attend a familiarisation and data collection session in a university biomechanics laboratory.

4.3.2 Participants

A power analysis using previous findings (Bates et al., 2013b) revealed a minimum requirement of 14 single-sex participants to achieve a 95% statistical power (alpha level > 0.05). To allow for potential dropout, twenty two male, varsity athletes participated in this study (21.9 ± 1.1 years, 180.5 ± 5.5 cm, 79.4 ± 7.8 kg). Inclusion criteria were that participants must be free from lower extremity injury within the last six months, have no history of lower limb ligamentous reconstructive surgery and participate in varsity field sports at least three times per week. All participants provided written informed consent, the study was approved by the Dublin City University Research Ethics Committee and the rights of the participants were protected.

During the familiarisation session, participants completed a physical activity readiness questionnaire (PAR-Q) and a general health questionnaire. Participant measurements (height, mass, leg dominance, leg length, knee and ankle widths) were recorded with leg dominance being defined as the limb with which the participant would prefer to kick a ball (Whyte et al., 2015). Knee and ankle width were measured using a digital caliper (Absolute Digimatic Caliper, Mitutoyo, Kawasakai, Japan) from the medial to lateral femoral epicondyle and medial and lateral malleoli, respectively. Participants were also familiarised with the VDJ (Moran and Marshall, 2006) and the HIIP (see descriptions below). Participants completed a minimum of 5 laps of the HIIP and 10 VDJs until they were comfortable with the correct execution of the tasks.

The data collection session began by recording participants' baseline heart rate after ten minutes of quiet sitting using a Polar heart rate monitor (model FT1; Polar Electro Inc., Lake Success, NY, USA). Following this, participants completed a warm-up of a 5 minute light jog, dynamic stretching of the lower limb and two laps of the HIIP at a self-selected jogging pace. Participants performed 5 sub maximal practices of the VDJ prior to data collection. Three maximal VDJs were then performed pre- and immediately post-HIIP with the average values analysed. For the VDJ, participants were instructed to place their hands on their hips, step off a 30cm box, land with feet in a toe first landing pattern on separate force plates and jump vertically upwards as high and quickly as possible before landing again on the force plates. Trials were excluded if participants did not land on separate force plates for the first and second landings of the VDJ. Briefly, the HIIP (Whyte et al., 2015) began with a 5 m forward sprint, a 90° change of direction and another forwards sprint of 5 m. Participants then backpedalled for 5 m before repeating the forwards and backwards sprints 4 times. Participants then performed 10 two legged jumps over 30 cm hurdles and 10 sidesteps over the same hurdles. The circuit was completed with 4 side shuffles over 5 m. HIIP circuit time was recorded using infrared timing gates (model TC; Brower Timing Systems, Draper, UT, USA). Participants were asked to complete the circuit at maximal effort and had a 30 second break between each circuit. Heart rate and rate of perceived exertion using the Borg 6–20 scale (Borg, 1970) were recorded immediately following completion of each HIIP circuit. The HIIP was discontinued when participants reported a score of 18 on the Borg Scale (Borg, 1970).

4.3.3 Data Collection and Processing

The Vicon plug-in-gait marker set, which consisted of 16 lower limb (Kim et al., 2014) and 4 trunk markers (Gutierrez et al., 2003) were applied to each participant. Additional tracking markers were placed midway between the anterior and posterior superior iliac spines bilaterally to assist with pelvic marker identification. Three dimensional trunk and lower extremity movements were recorded using a 12 camera Vicon motion analysis system (Oxford metrics Ltd., Oxford, UK). Two AMTI force plates (BP-600900; Advanced Mechanical Technology Inc., Watertown, Massachusetts, USA) recorded GRF data at 2000Hz. Force and marker trajectory data were filtered using a zero-lag, fourth-order, Butterworth technique (15 Hz cut off frequency) (Kristianslund et al., 2012). Nexus

VICON software (version 1.8.5; Vicon, Oxford, UK) synchronised the motion and force data at 250Hz and used inverse dynamics (Winter, 2009) to generate internal hip, knee and ankle joint moments which were projected onto the joint axes according to the anatomical coordinate system of the distal segment. The eccentric phases of the first and second landings were defined as the time from initial contact (when the unfiltered vGRF exceeded 10N) until the first occurrence of concentric centre of mass power. Kinematic and kinetic data [normalised to body mass (Bates et al., 2013b)] were exported from Nexus to self-written MATLAB software (R2012a, Math- Works Inc.,USA) for statistical analyses.

4.3.4 Statistical Analyses

SPM (Pataky et al., 2013) was used to statistically analyse trunk, pelvic, and thoracic on pelvic kinematics and hip, knee and ankle kinematics in the sagittal, frontal and transverse planes. Similarly, internal hip, knee and ankle moments in the sagittal, frontal and transverse planes and vertical, mediolateral and anterior-posterior GRFs, were statistically analysed. SPM analysed each point of the variable and subjected it to a 3 way repeated measures ANOVA to determine the effect of landing (first landing *versus* second landing), limb dominance (dominant *versus* non-dominant), HIIP (pre-HIIP *versus* post-HIPP) and any interaction effects. SPM provides a test statistic field (F value) and an evaluation of the significance (*p* value) in a similar way to univariate analysis. However, SPM also calculates these values over a range of points along the variable. This allows the identification of phases of significant differences rather than discrete, predetermined points. SPM incorporates a randomised field theory correction to ensure that any significant findings were not due to chance (Pataky et al., 2013). Data processing and statistical analyses were performed in MATLAB (R2012a, Math- Works Inc., USA). A Tukey-Kramer post hoc analysis corrected for multiple comparisons and an alpha level of $p < 0.05$ was used. Partial eta squared effect sizes (η^2) were classified as small 0.01- 0.06; medium 0.06 – 0.14; and large >0.14 (Pallant, 2010). For the purposes of concise reporting, it was decided that all statistically significant differences with medium to large effect sizes were considered clinically important and treated as such in the results and discussion. The physiological effects of the HIIP, reported in Table 4.1Table 4.1, were analysed by using paired sample t-tests to compare the first and final HIIP circuit completion times as well as heart rates at rest and post-HIIP.

Table 4.1 Indicators of exertion

Indicators	**Mean ± SD**
Number of circuits	6.2 ± 2.7
Time of circuit 1 (seconds)	39.5 ± 2.0
Time of final circuit (seconds)	43.4 ± 3.0*
Resting heart rate (bpm)	68.3 ± 7.5
Heart rate at final lap (bpm)	189.0 ± 9.4**
% Maximal heart rate at final lap †	95.3 ± 4.7

* = significant difference between first and final circuit time ($p = 0.039$)
** = significant difference between resting and maximal heart rates ($p < 0.001$)
† = $\% \text{ Maximal heart rate} = \left(\frac{\text{Heart rate}}{220 - \text{participiants age}}\right) \text{x } 100$

4.4 Results

There were a number of medium to large effects for landing on the biomechanics of the VDJ (Table 4.2). Greater trunk flexion, smaller hip and knee flexion angles, and smaller internal hip, knee and ankle moments were observed in the second landing compared with the first. Additionally, an initial greater vGRF followed by a smaller vGRF and a smaller posterior GRF were noted in the second landing.

There were also a number of medium to large effects for limb dominance (Table 4.3). Participants performed the VDJ with greater thoracic on pelvis side flexion away from the dominant limb and pelvic rotation towards the dominant limb. Participants also demonstrated greater internal hip, knee and ankle moments on the dominant leg. There was only one medium effect for the HIIP with increased trunk on pelvis flexion post-HIIP (Table 4.3).

Finally, there were a number of interactions with medium to strong effects for landing (first *vs* second) and limb dominance (dominant vs non-dominant) (Table 4.4). Participants demonstrated greater internal hip moments in dominant limb in the first landing while displaying greater internal knee moments in second landing for dominant leg.

4.5 Discussion and Implications

We hypothesised that potentially detrimental biomechanics in terms of ACL injury risk would be greater post-HIIP, in the dominant limb and during the second landing of the VDJ. We also hypothesised that interaction effects would demonstrate that the HIIP would have a greater effect on the first landing and the dominant limb. The results of our study only partially supported our hypotheses as demonstrated by significant main effects for landing and limb dominance (Table 4.2 -Table 4.4). Contrary to our hypothesis, the HIIP had only one medium main effect and there were no medium or strong interaction effects involving the HIIP.

Results for the main effect of landing (Table 4.2) demonstrated patterns of greater hip, knee and ankle moments in the first landing compared with the second. In particular, participants displayed greater (internal) knee extensor, valgus and external rotator moments. This pattern of loading can increase ACL strain (Oh et al., 2012) and may lead to a higher risk of injury in the first landing of the VDJ. However, there were also specific effects in the second landing that may increase the susceptibility to injury. Similar to Bates et al. (Bates et al., 2013b), who observed less peak knee and hip flexion angles in the second landing, we found less hip and knee flexion throughout the entire eccentric phase of the second landing compared with the first. A more extended limb posture throughout the eccentric phase may increase anterior tibial translation and ACL strain (Markolf et al., 2014) and, at least partially, explain the relationship between this posture during VDJ screening and ACL injury occurrence (Leppanen et al., 2017). This posture, coupled with the increased vGRF at the beginning of the second landing (19-45%), may increase the

Table 4.2 Main effects for landing on the biomechanics of the vertical drop jump

	Main Effect for Landing			
Variable	**% of phase**	***p***	**η^2**	**Effect**
Thoracic on pelvis flexion angles	1-52	0.028	0.068	Greater flexion in second landing
Hip flexion angles	1-100	0.001	0.062	Less hip flexion in second landing
Hip rotator moment	1-5	0.006	0.063	Smaller external rotator in second landing
Hip flexor moment	86-100	<0.001	0.085	Smaller extensor moment to greater flexor moment in second landing
Knee flexion angles	1-100	0.001	0.080	Less knee flexion in second landing
Knee rotator moment	52-100	<0.001	0.133	Smaller knee external rotator moment in second landing
Knee flexor moment	66-100	<0.001	0.165	Smaller extensor moment in second landing
Knee abductor/adductor moment	60-100	<0.001	0.126	Smaller abductor moment in second landing
Ankle rotator moment	53-100	<0.001	0.214	Smaller internal rotator moment in second landing
Ankle flexor moment	58-100	<0.001	0.163	Smaller plantarflexor moment in second landing
Ankle flexor moment	22-46	<0.001	0.069	Smaller plantarflexor moment in second landing
Vertical GRF	19-45	<0.001	0.087	Greater vGRF in second landing
Vertical GRF	60-100	<0.001	0.215	Smaller vGRF in second landing
Posterior GRF	61-100	<0.001	0.098	Smaller posterior GRF in second landing

GRF – Ground reaction force

Table 4.3 Main effects for Limb Dominance and a High Intensity, Intermittent Exercise Protocol on the Biomechanics of the Vertical Drop Jump

	Main effect for Limb Dominance			
Variable	**% of phase**	***p***	**η^2**	**Effect**
Thoracic on pelvis abduction angles	71-100	0.038	0.132	Greater side flexion to non-dominant side
Pelvic rotation angles	1-7	0.049	0.076	Greater rotation to dominant side
Hip rotator moment	58-61	0.012	0.063	Greater external rotator moment in dominant limb
Hip rotator moment	63-77	<0.001	0.070	Greater external rotator moment in dominant limb
Knee rotator moment	1-36	<0.001	0.110	Greater external rotator moment (v internal rotation) in dominant limb
Knee flexor moment	1-9	0.001	0.062	Greater flexor moment versus extensor moment in dominant limb
Knee abductor/adductor moment	39-85	<0.001	0.065	Greater abductor moment in dominant limb
Ankle rotator moment	1-30	<0.001	0.149	Greater internal rotator moment in dominant limb
Ankle flexor moment	1-11	0.014	0.096	Greater plantarflexor moment in dominant limb
Ankle abductor/adductor moment	1-100	<0.001	0.183	Greater evertor moment in dominant limb
Posterior GRF	9-34	<0.001	0.206	Smaller posterior GRF in dominant limb
Posterior GRF	66-100	<0.001	0.241	Smaller posterior GRF in dominant limb
	Main Effect for HIIP			
Variable	**% of phase**	***p***	**η^2**	**Effect**
Thoracic on pelvis flexion angles	1-100	<0.001	0.088	Increased flexion post-HIIP

HIIP – High intensity, intermittent exercise protocol

Table 4.4 Landing Phase by Limb Dominance Interaction Effects on the Biomechanics of the Vertical Drop Jump

	Landing Phase by Limb Dominance Interaction Effects			
Variable	**% of phase**	***p***	**η^2**	**Effect**
Hip rotator moment	1-8	<0.001	0.079	Greater external rotator moment for dominant leg in first landing
Hip flexor moment	1-7	<0.001	0.066	Greater extensor moment for dominant limb in first landing
Knee flexor moment	1-8	0.002	0.100	Greater extensor moment for dominant limb in second landing

susceptibility to injury in the early phase of the second landing. This is because an increased vGRF is associated with ACL injuries (Leppanen et al., 2017) and ACL injuries occur during the early stages of landing (Krosshaug et al., 2007). In contrast with this finding, Bates et al., (2013a) did not find any difference in peak vGRF between the first and second landings. As gender differences in normalised vGRF during drop jumps have been identified (Pappas et al., 2007; Quatman et al., 2006), the conflicting results may be due to fact that Bates et al., (2013a) investigated females whereas we analysed males. It may also be that SPM analysis provides a greater understanding of complex biomechanics compared with DPA (Richter et al., 2014). Any increase in susceptibility to injury towards the end of the eccentric phase of the second landing, due to the extended limb posture, is likely to be offset by the pattern of smaller knee extensor, valgus and external rotator moments in this phase as previously reported (Bates et al., 2013b). The findings of the current study demonstrate that there are different kinematic and kinetic patterns in the first and second landings of the VDJ which may pose different risks for ACL injuries and should be considered when screening athletes.

The differences between the two landings observed in the current study can be explained by the two distinctive landings of the VDJ. The first landing requires participants to counteract the impact GRFs and generate the sufficient power for the vertical jump while the second landing mimics a drop landing as it only requires the counteraction of the GRFs (Bates et al., 2013c). Therefore the first landing is similar to the commonly reported VDJ whereas the second is similar to a drop landing. In a study comparing drop landings to the first landing of a VDJ, Cruz et al., 2013 demonstrated that participants employ smaller hip and knee flexion angles during drop landings compared with the VDJ, which may explain our kinematic findings. However, Cruz et al. (2013) did not find any differences in joint kinetics between drop landings and VDJs which contrasts with the findings of the current study. This may be due to the different analyses used: Cruz et al. (2013) measured joint kinetics at peak anterior tibial strain whereas we analysed joint kinetics throughout the eccentric phase of the first and second landings.

The results of the current study demonstrate that biomechanical patterns associated with ACL injury risk are present in both landings of the VDJ. Traditionally the first landing only is used for screening (Hewett et al., 2005; Krosshaug et al., 2016; Padua et al., 2015) which results in potential risk factors in the second landing being overlooked, despite the epidemiological relationship with the second landing and ACL injury (Walden et al., 2015) Therefore, both landings should be included if the VDJ is being used for ACL injury risk factor screening. Also, incorporation of improved landing technique, for example encouraging greater knee and hip flexion during the second landing of VDJ, into ACL prevention and rehabilitation programmes may augment positive results. Future studies should investigate the effects of risk factors such as reaction to sporting stimuli during and after the second landing to increase the ecological validity of the task.

Asymmetrical lower limb biomechanics may predispose an athlete to ACL injury (Hewett et al., 2010; Pappas and Carpes, 2012). A study by Paterno et al. (Paterno et al., 2010) found that asymmetrical lower limb biomechanics during the VDJ predict noncontact ACL re-injury in male and female athletes. Although Paterno et al. (Paterno et al., 2010) did not report the mechanisms of ACL re-injury, the athletes investigated were involved in sports with jumping, cutting and lateral movements during which ACL injuries tend to occur (Koga et al., 2010; Krosshaug et al., 2007; Walden et al., 2015). Therefore, it would suggest that the findings of the VDJ are transferrable to more dynamic environments. The findings of the current study identified lower limb asymmetries with a number of moderate to strong main effects for limb dominance. The dominant limb demonstrated patterns of greater hip, knee and ankle moments during both landings of the VDJ. Greater knee external rotator moments increase ACL strain (Oh et al., 2012). Therefore, the greater knee external rotator moments in the first 36% of the landing phases in particular, suggest that the dominant leg is at an increased risk of ACL injury. This risk may be partially offset by the greater knee flexor moment observed at the beginning of landing as hamstring activation decreases ACL strain (Baratta et al., 1988). The pattern of increased external knee rotator moment may be contributed to by trunk and pelvic kinematics as they contain a significant proportion of body mass. Specifically, there was greater pelvic rotation towards the dominant side and greater thoracic side flexion away from the dominant side, relative to the pelvis. This may be important given that reduced trunk kinematic control is associated with ACL injury (Hewett et al., 2009; Shimokochi et al., 2013; Zazulak et al., 2007). Also, reduced hip external rotator strength predicts ACL injury (Khayambashi et al., 2016) and re-injury (Paterno et al., 2010). The asymmetry of hip external rotator moments observed in the current study suggests a greater demand for external hip rotator strength in the dominant limb. If this demand is not met, the athlete may be placed at a greater risk of ACL injury. This may at least partially explain the higher rate of ACL injuries in the dominant limb of males (Brophy et al., 2010). In summary, the asymmetrical loading evident in healthy varsity athletes involved in field sports in the current study may be a cause for concern. From a clinical perspective, the practice of assessing the mechanics of one limb and assuming the same findings apply to the unexamined limb (Padua et al., 2015) may lead to erroneous findings and should be reconsidered.

The landing by dominance interaction effects (Table 4.4) demonstrated patterns of greater dominant hip rotator and flexor moments in the first landing and greater dominant knee extensor moment during the second landing. This shows that the dominant limb utilizes different strategies during the beginning (1-8% of the landing) of the eccentric phases of both landings of the VDJ. The greater knee extensor moment in the second may increase the strain of the ACL at this point, particularly as the knee is in a relatively extended position (Markolf et al., 2014). This reinforces the importance of including the second landing of the VDJ in ACL injury screening, prevention and rehabilitation protocols. The greater hip extensor and rotator moments evident at the beginning of the first landing demonstrate the importance of hip control of the dominant limb during this phase.

Contrary to our hypotheses, the HIIP only had one medium effect (Table 4.3) which displayed a pattern of increased thoracic flexion relative to the pelvis. Increased trunk flexion has been found to lead to greater hip extensor moments and a decrease in knee extensor moment (Shimokochi et al., 2013). However, there were no moderate or strong main or interaction effects involving the HIIP on lower limb kinetics or kinematics in the current study despite participants reporting high levels of exertion and demonstrating increased circuit completion times, which would indicate fatigue (Table 4.1). As the HIIP, which simulates field sports (Whyte et al., 2015), is not sufficient to lead to moderate or strong biomechanical alterations, it suggests that participants utilised a neuromuscular reserve for short duration, maximal activities such as the VDJ. This contrasts with previous studies which demonstrated that fatigue detrimentally affected the biomechanics of the VDJ (Haddas et al., 2016; Weinhandl et al., 2011). This contrast may be explained by the fact that previous studies used local muscle fatiguing protocols i.e. repetitive VDJs (Weinhandl et al., 2011) and squats (Haddas et al., 2016), whereas the current study and that by Moran and Marshall (2006) used running protocols which did not lead to kinematic differences. Future studies should consider employing methods to minimize the focus of an athlete on the post-fatigue testing, such as by continuously testing throughout the fatiguing protocol. The findings of the current study suggest that there is limited benefit to completing screening VDJs post-HIIPs in order to identify kinetic and kinematic risk factors for ACL injuries in male, field sport athletes.

There are a number of limitations to the current study. Firstly, although the VDJ has high levels of reliability (Malfait et al., 2014) and mimics the rebound jump landing in basketball (Bates et al., 2013a), it does not replicate the sporting situation fully. This could be improved in future studies by including a sports specific activity immediately after the second landing such as running or cutting. Incorporation of a reactive element to this would further improve ecological validity as the majority of non-contact ACL injuries occur when an athlete is reacting to the sporting environment (Walden et al., 2015). Secondly, the HIIP is a short protocol that does not replicate the cumulative fatigue that occurs during field-sports. Thirdly, although the analysis of kinetic and kinematic patterns is more informative than discrete points, it makes comparison with previous research difficult. Finally, the results of the current study are applicable to males only as there are gender specific differences in jumping and landing activities (McLean et al., 2007).

4.6 Conclusion

In conclusion, statistical parametric mapping of the VDJ adds to the existing understanding of VDJ biomechanics and the effects of three factors thereon, namely landing (first or second), limb dominance and fatigue. The first and second landings of the VDJ demonstrate detrimental biomechanics in terms of risk factors for ACL injuries and should be included in ACL screening, prevention and rehabilitation programmes. The dominant limb in males generates greater hip, knee and ankle moments compared with the non-dominant limb, which may place the dominant limb at a higher risk of injury. However, the general fatiguing protocol, the HIIP, did not result in any medium or strong main or interaction effects on the biomechanical patterns of the VDJ. Practitioners should analyse both limbs and landings of the VDJ when using the VDJ as a screening or assessment tool.

4.7 Link to Chapter 5

The results of study 2 demonstrated that the HIIP induced fatigue in participants as demonstrated by the increase in HIIP lap completion times. However, the HIIP-induced fatigue did not affect the biomechanics of the VDJ. Therefore, the results of this study 2 suggest that HIIP-induced fatigue does not increase the risk of ACL injury during double leg landing activities, such as the VDJ. Therefore, performance of the VDJ following the HIIP may not sufficiently stress the neuromuscular control of athletes. As ACL injuries often occur during single leg, change of direction tasks, such as cutting, particularly when the athlete is responding to unanticipated sporting situations, biomechanical risk factors for ACL injuries may be more apparent. The ability of the athlete to safely complete tasks, particularly in response to unanticipated situations, necessitates a high level of neuromuscular control using feed-forward and feedback mechanisms. This system of control requires adequate neurocognitive function. Given that athletes are proposed to be an elevated risk of ACL injury when completing unanticipated high risk activities in a fatigued state, it is important to understand the relationship between fatigue and neurocognitive function. Therefore, study 3 (chapter 5) investigates the effect of a HIIP on neurocognitive function.

Chapter 5 The Effect of a High Intensity, Intermittent Exercise Protocol on Neurocognitive Function in Healthy Adults; Implications for Return to Play Management Following Concussion

Study 3

Study 3 “The effect of a high intensity, intermittent exercise protocol on neurocognitive function in healthy adults; implications for return to play management following concussion”

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STATEMENT OF CONTRIBUTION: Kieran Moran was the research supervisor for this study. Grainne Kerr and Nicola Gibbons assisted in the data collection and analysis.

Author’s note: In line with the overall thesis, this study aimed to investigate the effects of a high intensity, intermittent exercise protocol on neurocognitive function. However, in attaining publication, feedback from the reviewers led to an emphasis on the effects of the HIIP on neurocognitive function in relation to concussion in the introduction and discussion sections. Section 5.7 (Link to chapters 6 and 7) will outline the relevance of the findings of study 3 to the overall aims of the thesis.

5.1 Abstract

Context: Determination of return-to-play (RTP) following sports-related concussion (SRC) is critical given the potential consequences of premature RTP. Current RTP guidelines may not identify persistent exercise-induced neurocognitive deficits in asymptomatic athletes following SRC. Therefore, post-exercise neurocognitive testing has been recommended to further inform the RTP determination. In order to implement this recommendation, the effect of exercise on neurocognitive function in healthy athletes should be understood.
Objective: To examine the acute effects of a high intensity, intermittent exercise protocol (HIIP) on neurocognitive function assessed by the SDMT and Stroop Interference tests.
Design: A cohort study.
Setting: University laboratory.

Participants: 40 healthy male athletes (age 21.25 ± 1.29 years, years of education 16.95±1.37).

Intervention: Each participant completed the SDMT and Stroop Interference tests at baseline and following random allocation to a condition (HIIP versus control). A mixed between-within subjects ANOVA assessed time (pre- versus post-condition) by condition interaction effects.

Main Outcome Measures: SDMT and Stroop Interference test scores.

Results: There was a significant time by condition interaction effect ($p < 0.001$, $\eta^2 = 0.364$) for the Stroop Interference test scores indicating that the HIIP group scored significantly lower (56.05 ± 9.34) post-condition compared with the control group (66.39 ± 19.6). There was no significant time by condition effect ($p = 0.997$, $\eta^2 < 0.001$) for the SDMT indicating that there was no difference between SDMT scores for the HIIP and control groups (59.95 ± 10.7 *vs*. 58.56 ± 14.02).

Conclusions: In healthy athletes, the HIIP results in a reduction in neurocognitive function as assessed by the Stroop Interference test, with no effect on function as assessed by the SDMT. Testing should also be considered following high intensity exercise in determining RTP decisions for athletes following SRC in conjunction with the existing recommended RTP protocol. These results may provide an initial reference point for future research investigating the effects of a HIIP on the neurocognitive function of athletes recovering from SRC.

5.2 Introduction

Determination of return-to-play (RTP) following sports-related concussion (SRC) is critical given the potential consequences of premature RTP, including lower neurocognitive function and an increased risk of further concussive/severe brain injury (McCrory et al., 2013). Current best RTP practice involves both comparing preseason with post-concussive computerised neurocognitive scores at rest, and employing a monitored, graded RTP protocol (McCrory et al., 2013) during which an athlete subjectively reports symptom status. This approach faces a number of challenges. Firstly, reliance on patient-reported symptoms may not be appropriate as neurocognitive deficits persist in otherwise asymptomatic athletes post SRC, with testing such as the Stroop Interference test and Symbol Digits Modality Test (SDMT) sensitive to these deficits (McCrea et al., 2005). Secondly, exercise induces neurocognitive deficits in asymptomatic athletes who were previously concussed but subsequently cleared to RTP following completion of the recommended protocol (McGrath et al., 2013). This suggests that a period of exercise-induced cerebral dysfunction and vulnerability persists beyond the symptomatic period, making such athletes more vulnerable to injury should they return to sport at this point (McCrea et al., 2005). Therefore, as neurocognitive tests are sensitive to subtle neurocognitive deficits post concussive symptoms resolution (McCrea et al., 2005), it has been proposed that neurocognitive tests should be conducted immediately after sports-specific exercise (McGrath et al., 2013) that mimic the high intensity exercise of field sports where concussion is common. The computer dependence and time requirement of neurocognitive testing, may limit its applicability in assessing the acute effects of exercise (Eckner et al., 2014)**.** On the other hand, the Stroop Interference test (Barwick et al., 2012; Register-Mihalik et al., 2012) and SDMT (Iverson et al., 2005; McCrea et al., 2003) are: commonly used in concussion research and assessment, reliable (Register-Mihalik et al., 2012), inexpensive, quick to administer and suitable for serial neurocognitive testing (Register-Mihalik et al., 2012); as such they may be suitable to assess the acute effect of exercise on neurocognitive function following SRC.

As exercise induces findings similar to SRC, such as decline in computerised neurocognitive testing (McGrath et al., 2013) and the manifestation of concussive-like symptoms (Alla et al., 2010) in healthy athletes, the acute effect of exercise on neurocognitive function in healthy subjects should be understood in order to facilitate appropriate RTP decisions following SRC. This study aims to investigate the effect of a high intensity, intermittent exercise protocol (HIIP) on neurocognitive function as assessed by the SDMT and Stroop Interference Tests. Whereas moderate intensity exercise positively affects neurocognitive function (Lambourne and Tomporowski, 2010; McMorris and Hale, 2012), high intensity exercise negatively affects neurocognitive function (Del

Giorno et al., 2010; McMorris et al., 2009; Wang et al., 2013). Therefore, it was hypothesised that the HIIP would have a negative effect on test scores.

5.3 Methods

5.3.1 Participants

40 male universtiy athletes participated in the study (age 21.25 ± 1.29 years, years of education 16.95±1.37). Participants were excluded if they had previously taken the SDMT or Stroop Interference tests, were colour blind, injured, previously concussed, or were taking medication which may affect their neurocognitive ability. Participants refrained from exercise, alcohol and caffeine consumption for 24 hours prior to testing. Approval was granted by the relevant research ethics committee.

Participants were required to attend on two separate occasions. Initially they underwent a familiarization session with demonstration of the HIIP protocol. Following a 2-5 day interval, participants returned for SDMT and Stroop Interference testing. 20 participants were randomly assigned to the control (age 21.24 ± 1.25 years, years of education 17 ± 1.41) and HIIP intervention (age 21.05 ± 1.33 years, years of education 16.89 ± 1.33) groups. Participants were directly matched across groups by their field sport which included Gaelic football, hurling and rugby.

5.3.2 Procedures

The SDMT and Stroop Interference tests were administered at baseline. Participants in the intervention group only, completed a five minute dynamic warm-up followed by the HIIP. The HIIP was discontinued when participants reported 18 on Borg's rating of perceived exertion scale. Within 15 seconds of completion, the SDMT or Stroop Interference test was re-administered (in a random order). After this, participants completed further HIIP circuits until they again reported 18 on Borg's scale and then completed the remaining test. Control participants were administered the tests at the same time as the intervention participants. Tests were administered in a random order.

The HIIP (Figure 3.1) consisted of 5 metre (m) forward sprinting, a 90° angle change of direction, another 5m forward sprint and then a 5m backpedal. This was repeated four times before completing 5 two-legged jumps over hurdles (30cm height), turned and repeated another 5 times. Then they performed side-stepping exercises over the hurdles before 4 lateral shuffles back and forth of 5m each. Participants were instructed to complete the above as quickly as possible. This protocol would relate to the 3rd rehabilitation stage (Sport Specific Exercise) on the RTP guidelines following SRC (McCrory et al., 2013). Circuit time was recorded using infrared timing gates (Brower, USA). Following circuit completion, participants rested for 30s before repeating the circuit. Heart rate was monitored using a Polar heart rate monitor.

The Stroop Interference test assesses frontal lobe executive function because of the inhibitory control it requires (Lezak, 2012). It consists of a list of words of colours which are different from the colour in which they are printed. Participants were given 45s to state aloud the ink colour rather than the printed word. The score was calculated as the number of correct responses in 45s.The SDMT assesses the parietal and frontal lobe functions of attention and information processing speed (Iverson et al., 2005). This pen and paper task involved the pairing of numbers to symbols in a series of boxes in lines of fifteen. The score was calculated as the number of correct responses in 90s. In both tests, higher scores represent improved neurocognitive function.

5.3.3 Statistical Analysis

The main aim of the study was investigated using a mixed between-within analysis of variance to assess any time (pre- versus post-condition) by condition (HIIP versus control) interaction effect. The dependent variables analyzed were Stroop Interference and SDMT test scores. Effect sizes (partial-eta squared) were calculated and ranked using the Cohen classification (0.01=small, 0.06=moderate, 0.14=large effects). Paired sample t-tests compared the first and final HIIP circuit completion time, and resting and post HIIP heart rates. SPSS statistical analysis software (SPSS version 17.0) was used for all analyses with the level of statistical significance set at $p<0.05$.

5.4 Results

Regarding the primary aim of the study, there was a significant time by condition interaction with a large effect size for the Stroop Interference scores (Lambda = 0.64, $F_{(1.38)} = 21.77$, $p < 0.001$, $\eta^2 = 0.364$), indicating that the HIIP resulted in significantly lower Stroop Interference test scores (Table 5.1). There was a significant main effect for time for the Stroop Interference Test Scores (Lambda = .61, $F_{(1.38)} = 23.9$, $p = 0.001$, $\eta^2 = 0.386$) indicating that there was a significant increase post condition. There was no significant effect for condition for the Stroop Interference test scores ($F_{(1.38)} = 1.96$, $p = 0.169$, $\eta^2 = 0.049$) indicating that there was no significant difference between conditions.

There was a significant main effect for time for the SDMT (Lambda = .829, $F_{(1.38)} = 7.84$, $p = 0.008$, $\eta^2 = 0.171$) indicating that there was a significant increase post condition (Table 5.1). However, There was no significant time by condition interaction effect (Lambda = 1.0, $F_{(1.38)} = .000$, $p = 0.997$, $\eta^2 = 0.000$) indicating that the HIIP did not have any different effect than rest. There was no significant effect for condition for the Stroop Interference test scores ($F_{(1.38)} = 0.645$, $p = 0.427$, $\eta^2 = 0.017$) indicating that there was no significant difference between conditions.

Participants completed on average 6.58 (± 0.81) circuits of the HIIP with an average heart rate of 188.03±7.03 beats per minute (bpm) or 94.6 ± 3.5% maximum predicted heart rate at completion, versus 63.12 ± 6.91bpm at rest (Table 5.2). Circuit completion times increased significantly (p<0.001) from the initial to final circuit (51.09 ± 2.61s *vs.* 54.23 ± 2.12s, respectively).

Table 5.1 Precondition and postcondition Stroop Interference Test and Symbol Digit Modalities Test scores, mean ±SD, and effect sizes of main and interaction effects

	Control Group		Exercise Intervention Group		Condition Main Effect [a]		Time Main Effect [b]		Time by Exercise Interaction [c]	
	Pre-condition	Post-condition	Pre-condition	Post-condition	*P*	ES	*P*	ES	*P*	ES
Stroop Interference Test Score	54.89±15.36	66.39±19.6	55.76±10.48	56.05±9.34	0.28	.031	0.000	0.386	0.000	0.364
Symbol Digit Modalities Test Score	53.06±11.5	58.56±14.02	55.86±7.51	59.95±10.7	0.526	.011	0.008	0.171	0.997	0.000

a. Indicates difference between HIIP and control conditions, b. Indicates difference between pre and post condition , c. Indicates group differences for pre and post condition. ES = Effect size (partial eta-squared value)

Table 5.2 Markers of Exertion for the HIIP Condition Group

	Males (n=20)
Resting Heart Rate (bpm)	63.12 ± 6.91
Maximal Heart Rate (bpm)	188.03±7.03
% HR max at final lap	94.6±3.5
Number of circuits	6.9±1.68
Time of circuit 1 (s)	51.09± 2.61
Time of final circuit (s)	54.23±2.12*

* = significant difference between time of circuit 1 and time of final circuit ($p < 0.05$)

5.5 Discussion

There is a need to understand the effect of exercise on neurocognitive capacity in healthy individuals to facilitate future research examining the effect of exercise on neurocognitive function in athletes following SRC. The hypothesis of the current study that a HIIP would have a negative effect on neurocognitive function in healthy adults was partially supported with reduced neurocognitive function evident as assessed by the Stroop Interference Test but not the SDMT. This partially supports previous studies that found that high intensity exercise negatively effects neurocognitive function (McMorris, Davranche et al. 2009, Wang, Chu et al. 2013, Del Giorno, Hall et al. 2010). The HIIP resulted in fatigue as defined as an ‘exercise-induced decline of performance’ (Knicker, Renshaw et al. 2011) indicated by the significant increase in lap completion times. Therefore, the results from this study may only be applicable to high intensity exercise protocols that induce fatigue. It must also be noted that although all participants were varsity athletes, there may have been differing levels of fitness which may affect the participants’ response to the HIIP and affect the results.

The results of the current study support previous research that found the Stroop Interference test sensitive to the effects of fatigue (Barwick, Arnett et al. 2012). They also support previous studies (McGrath, Dinn et al. 2013) demonstrating that fatigue does not affect processing speed and reaction time elements of computerized, neurocognitive tests which correlate highly with SDMT scores (McGrath, Dinn et al. 2013). These results may be explained by the transient hypofrontality hypothesis (Dietrich 2006). This states that exercise of a sufficient intensity may require frontal lobe resources, impairing frontal executive function. Previous studies have demonstrated that exercise intensities resulting in greater than 60% (Del Giorno et al. 2010) and 80% maximum heart rate (Max HR) (Del Giorno et al. 2010, Wang et al. 2013) lead to impaired frontal lobe executive function. This suggests that the observed 94% Max HR in the current study may impair frontal lobe function which is assessed by the Stroop Interference test (Lezak 2012), whereas the SDMT assesses both frontal and parietal lobe activity (Iverson, Lovell et al. 2005).

Currently, RTP decisions incorporating neurocognitive scores at rest, and a monitored, graded RTP protocol (McCrory, Meeuwisse et al. 2013) does not detect exercise-induced neurocognitive deficits in otherwise asymptomatic athletes following SRC (McGrath, Dinn et al. 2013). Therefore, it is recommended that determination of RTP following SRC is further informed by post- exercise neurocognitive testing (McGrath et al. 2013, Eckner, Kutcher et al. 2014). To accurately determine RTP using post-exercise neurocognitive testing, the effect of exercise on such tests in healthy athletes needs to be understood. The results of the current study demonstrate that in healthy athletes SDMT scores should improve when tested post-HIIP in comparison with pre-HIIP testing, while the Stroop Interference Test scores should remain unchanged. These results may provide an initial reference point for future studies investigating the effects of a HIIP on the neurocognitive function of athletes recovering from SRC.

5.6 Conclusion

High intensity exercise negatively affects aspects of neurocognitive function, specifically the frontal executive function as measured by the Stroop Interference Test. In addition to the currently recommended RTP protocol following SRC, the proposed neurocognitive testing immediately following sport-specific exercise (McGrath, Dinn et al. 2013) should be conducted after high intensity exercise. Future research should also examine the acute effects of exercise on such neurocognitive test scores in athletes recovering from SRC.

5.7 Link to Chapters 6 and 7

The results of study 3 demonstrate that HIIP-induced fatigue negatively affects aspects of neurocognitive function as measured by the Stroop Interference test. Adequate neurocognitive function is required during sporting activities to filter information from the sporting environment, implement and modify the appropriate motor programme (Swanik et al., 2007; Swanik, 2015). A deficit in neurocognitive function may affect these processes, resulting in the implementation of a suboptimal motor programme, altered biomechanics and an increase injury risk. Specifically in relation to ACL injuries, Swanik et al., (2007) demonstrated that American footballers who sustained an ACL injury had significantly lower neurocognitive function scores than a matched control group (Swanik et al., 2007). Given this association between reduced neurocognitive function and ACL injury and the fact that study 3 demonstrated that a HIIP-induced detrimental effect on neurocognitive function, a HIIP may place an athlete at an increased risk of ACL injury.

It is important to consider the potential increase risk of injury due to a HIIP-induced negative effect on neurocognitive function with the results of study 1. Study 1 found that a HIIP resulted in decreased dynamic postural control, which may increase the risk of lower limb injury. As adequate dynamic postural control and neurocognitive function are required to safely complete sporting activities. The findings of studies 1 and 3 indicate that athletes may be at a greater risk of injury when performing activities following high intensity, intermittent exercise. It is possible that the negative effects of a HIIP on neurocognitive function and dynamic postural control may lead to an alteration in the biomechanics of cutting activities. To follow on from this, study 4 (chapter 6) and study 5 (chapter 7) will investigate the effects of the HIIP on the biomechanics of crossover cutting and side cutting, respectively. Furthermore, Swanik (2015) proposed that performance of sporting activities in unanticipated situations places greater demands on neurocognitive function than during anticipated situations. Therefore, it is important to consider the individual and combined effects of unanticipation and fatigue induced by the HIIP. This is particularly the case given that performance of cutting activities in the combined unanticipated and fatigued state has a greater detrimental effect on biomechanical risk factors for ACL injuries compared with the individual fatigued or unanticipated condition (Borotikar et al., 2008; McLean and Samorezov, 2009). For these reasons, study 4 (chapter 6) and study 5 (chapter 7) will investigate the individual and combined effects of the HIIP and unanticipated condition on the biomechanics of crossover cutting and side cutting. The results of these studies will assist in our understanding of the risk factors for ACL injuries and may facilitate the development and refinement of ACL injury prevention programmes.

Chapter 6 The effect of High Intensity Exercise and Anticipation on Trunk and Lower Limb Biomechanics during a Crossover Cutting Manoeuvre

Study 4

Study 4: "The effect of high intensity exercise and anticipation on trunk and lower limb biomechanics during a crossover cutting manoeuvre."

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STATEMENT OF CONTRIBUTION: Kieran Moran was the research supervisor for this study. Siobhan O'Connor and Chris Richter assisted in the data collection and analysis.

6.1 Abstract

We investigated the effects of high intensity, intermittent exercise (HIIP) and anticipation on trunk, pelvic and lower limb biomechanics during a crossover cutting manoeuvre. Twenty-eight male, varsity athletes performed crossover cutting manoeuvres in anticipated and unanticipated conditions pre- and post-HIIP. Kinematic and kinetic variables were captured using a motion analysis system. Statistical parametric mapping (repeated-measures ANOVA) was used to identify differences in biomechanical patterns. Results demonstrated that both unanticipation and fatigue (HIIP) altered the biomechanics of the crossover cutting manoeuvre, whereas no interactions effects were observed. Unanticipation resulted in less trunk and pelvic side flexion in the direction of cut ($d = 0.70$ – 0.79). This led to increased hip abductor and external rotator moments and increased knee extensor and valgus moments with small effects ($d = 0.24$ - 0.42), potentially increasing ACL strain. The HIIP resulted in trivial to small effects only with a decrease in internal knee rotator and extensor moment and decreased knee power absorption ($d = 0.35$), reducing potential ACL strain. The effect of trunk and hip control exercises in unanticipated conditions on the crossover cutting manoeuvre should be investigated with a view to refining ACL injury prevention programmes.

6.2 Introduction

Anterior cruciate ligament (ACL) injuries lead to considerable short and long term consequences, including the early development of osteoarthritis (Oiestad et al., 2009), reduced sports participation (Maquirriain and Megey, 2006) and high medical costs (Silvers and Mandelbaum, 2007). Non-contact ACL injuries occur during high-risk sporting manoeuvres such as deceleration and cutting activities (Krosshaug et al., 2007; Walden et al., 2015) particularly when the athlete is responding to the sporting environment (Boden et al., 2009; Walden et al., 2015). This, coupled with the higher incidence of injuries towards the end of matchplay (Hawkins et al., 2001), has resulted in a significant amount of research investigating the effects of fatigue and unanticipation on the biomechanics of side cutting manoeuvres (Borotikar et al., 2008; Kim et al., 2014; McLean and Samorezov, 2009). However, the crossover cutting manoeuvre has received much less attention (Cortes et al., 2014) despite the fact that it is regularly used to change direction in field sports (Andrews et al., 1977; Potter et al., 2014). During the crossover cutting manoeuvre, the athlete plants the foot ipsilateral to the new running direction (i.e. the right foot for a right crossover cut) and then crosses the contralateral limb around in the new direction of travel. To achieve this, the stance limb must first decelerate the forward progression of the centre of mass and then accelerate it in the new direction of travel. This requires eccentric control of the joints of the lower limb in all three planes, allowing the pelvis and trunk to rotate around the stance limb and progress in the new direction (Andrews et al., 1977; Nyland et al., 1999). This change of direction technique increases loading of the knee and may place the ACL at an increased risk of injury (Besier, Lloyd, Cochrane et al., 2001). Despite this, crossover cutting manoeuvres have been proposed to be a safer change of direction technique (McGovern et al., 2015; Potter et al., 2014) as non-contact ACL injuries occur less frequently during crossover cutting (Cochrane et al., 2007). This proposal should be treated with caution as the lower observed incidence of ACL injuries during crossover cutting may be due to lower frequency of crossover cutting manoeuvers as a change of direction technique compared with side cutting (Potter et al., 2014). Furthermore, the specific effects of risk factors for ACL injuries such as fatigue and unanticipation on crossover cutting manoeuvres are not well understood (Cortes et al., 2014).

Only three studies have investigated the effects of unanticipation on discrete kinematic and kinetic measures of crossover cutting, with findings being contradictory (Besier, Lloyd, Ackland et al., 2001; Cochrane et al., 2010; Kim et al., 2014). Besier et al., (2001a) found unanticipation resulted in increased internal knee varus and external rotation moments, whereas Kim et al. (2014) reported an increase in internal knee extensor moments, and a decrease in varus and external rotation moments. Moreover, Cochrane et al., (2010) found no effect of unanticipation. These contradictory results may be due to methodological differences as Besier et al., (2001) analysed peak findings for the weight acceptance phase whereas Kim et al., (2014) reported peak values during the entire stance phase. As non-contact ACL injuries occur during the weight acceptance phase of cutting manoeuvres (Krosshaug et al., 2007), analysis during this phase may provide a clearer understanding of potential relationships with injury. Furthermore, given that reduced hip strength leads to decreased knee control (Claiborne et al., 2006; Willson et al., 2006) and is a predictor for non-contact ACL injuries (Khayambashi et al., 2015), it is surprising that only one study (Kim et al., 2014) analysed the effects on hip biomechanics. In addition, altered trunk position directly affects knee joint loading (Donnelly et al., 2012; Jamison et al., 2012; Shimokochi et al., 2013) and deficits in trunk control predict non-contact ACL injuries (Zazulak et al., 2007) both of which may explain the more extended and side flexed trunk position observed during non-contact ACL injuries (Hewett et al., 2009). However, the effect of anticipation on trunk, pelvic and hip biomechanics during the crossover cut has not been investigated.

Fatigue has been found to have a negative effect on knee biomechanics during a crossover cut. It leads to a decrease in hip (McGovern et al., 2015; Potter et al., 2014) and knee flexion (Cortes et al., 2014; McGovern et al., 2015; Potter et al., 2014), a move towards a knee valgus angle (Cortes et al., 2014) and decreased knee rotational control (Nyland et al., 1999). All of these fatigue-induced changes are considered to increase the risk of ACL injuries (Shultz et al., 2015). As ACL injuries frequently occur in sports with intermittent bouts of high-intensity exercise and multiple changes of direction (Krosshaug et al., 2007; Walden et al., 2015), it is important that fatiguing protocols closely resemble the physiological demands of that sport (Knicker et al., 2011). Knicker et al. (Knicker et al., 2011), recommend greater attention on temporary fatigue that occurs during sport following high intensity, intermittent exercise. The combination of fatigue and unanticipation has been proposed to be the ‘worst case scenario’ in terms of ACL injury risk (Borotikar et al., 2008; McLean and Samorezov, 2009). While some research has found an interaction effect between fatigue and unanticipation in side cutting (Borotikar et al., 2008; McLean and Samorezov, 2009) this has not been supported recently in studies using fatigue protocols which more closely mimic field-based sports (Collins et al., 2016; Khalid et al., 2015). The individual and combined effects of fatigue and unanticipation on the crossover cutting manoeuvre are unclear. To date, only one study has investigated the effect of fatigue on unanticipated crossover cutting manoeuvres (Cortes et al., 2014). That is, it compared unanticipated crossover cutting manoeuvres pre fatigue with unanticipated crossover cutting manoeuvres post fatigue. As this study did not investigate the effects of fatigue on both anticipated and unanticipated crossover cutting manoeuvres, it is not possible to discern any interaction effects between fatigue and unanticipation. Therefore, in order to improve our understanding of risk factors for ACL injuries, it is important to analyse the individual and combined effects of fatigue and unanticipation on the pattern of trunk, pelvic and lower limb biomechanics during crossover cutting manoeuvres.

The biomechanics of activities such as the crossover cutting manoeuvre have traditionally been investigated using discrete point analysis (DPA). DPA involves feature reduction and subsequent analysis of kinematic and kinetic waveforms, often capturing less than 5% of the data (Richter et al., 2014). DPA presents a number of limitations. Firstly it requires the preselection of measures or features (example peak knee flexion angle) to be analysed based on previous research. This may discard potentially important information (Richter et al., 2014) and over simplify the original highly multivariate datasets (Pataky et al., 2013). Secondly, DPA does not take into account temporal characteristics of the selected features and can therefore compare unrelated features against each other forming false conclusions

(Richter et al., 2014). Lastly, there is growing evidence that continuous analysis techniques can provide greater insight into biomechanical data (Pataky et al., 2013; Richter et al., 2014). A technique that has gain in popularity is statistical parametric mapping (SPM) because of it easy implementation and representation of findings. SPM analyses the original vectors rather than discrete points (Pataky et al., 2013). It also uses randomised field theory (Adler, 2007) to determine a critical threshold to ensure that vector differences found do not simply occur by chance (Pataky et al., 2013). This may explain why SPM is more effective at identifying biomechanical features that affect outcome and that can discriminate between conditions such as knee flexion angle and ground reaction force (Pataky et al., 2013; Richter et al., 2014). Therefore, our understanding of the individual and combined effect of different ACL injury risk factors is likely to be enhanced by using continuous analysis techniques (Shultz et al., 2015).

The aims of this study were to investigate the effects of: (1) a high intensity, intermittent exercise protocol (HIIP) (pre-HIIP versus post-HIIP), and (2) the state of anticipation (anticipated versus unanticipated condition), on the biomechanics of the trunk and stance limb during the weight acceptance phase of the crossover cutting manoeuvre. It was hypothesized that there would be significant main effects for the HIIP leading to decreased knee flexion angles and increased knee extensor moments. It was also hypothesised that state of anticipation would result in altered trunk kinematics and increased knee moments in the frontal and transverse planes.

Finally, it was hypothesised that interaction effects would demonstrate that biomechanical risk factors (such as increased knee valgus angle, internal varus, extensor, and external rotator moments, and altered trunk kinematics) proposed to increase ACL loading would be greater during the performance of unanticipated cutting manoeuvres post-HIIP.

6.3 Methods

6.3.1 Participants

A power analysis to achieve a 95% statistical power with an alpha level of greater than 0.05 revealed a required minimum of 22 participants using previous data from a study investigating the effects of perturbation training on trunk control during lateral reactive jumps (Weltin et al., 2016). To allow for potential dropout, twenty eight male, varsity Gaelic footballers participated in this study (age, 21.71 (SD 2.16) years; height 178.71 (SD 14.64) m; body mass 81.82 (SD 11.44) kg). Gaelic football is a field sport that requires running, catching, kicking and multiple changes of direction(O'Connor et al., 2016). Inclusion criteria were that participants were male, currently injury-free, and participating in Gaelic football training on at least three occasions per week. Participants were excluded if they had experienced any lower limb injury within the last six months, or had lower limb ligamentous reconstructive surgery. All participants provided written informed consent, and the study was approved by the local Research Ethics Committee. Participants were required to attend a familiarisation session and a data collection session.

6.3.2 Experimental Procedure

During the familiarisation session, participant measurements (height, weight, leg dominance, leg length, knee and ankle widths, and maximum horizontal jump distance) were recorded. Baseline heart rate was recorded after ten minutes of quiet sitting. Leg length was measured as the distance from the anterior superior iliac spine to the medial malleolus. Following this, the maximum standing horizontal jump distance was recorded. Participants were instructed to “begin by standing on two feet and then jump as far forwards as you can and land on your dominant foot”. The best of three attempts was recorded (maximum mean jump distance 2.3m (SD 0.30)). Crossover and side cutting manoeuvres, and stop jumps were explained and demonstrated to the participants with only data for the crossover cutting manoeuvre analysed in this study. Participants were given a minimum of ten minutes to practise the tasks. Following explanation of the HIIP, participants practised the HIIP until they were comfortable and accurate in its execution.

During the data collection session participants performed crossover cutting manoeuvres following a standardized warm-up consisting of a 10 minute light jog and 5 minutes of dynamic stretching. Participants performed a horizontal jump equalling 70% of their maximum jump distance before landing on their dominant leg and performing the crossover cutting manoeuvre at 45^0(Cortes et al., 2014) to their dominant side (Figure 0.1). The crossover cut was completed on the dominant leg as 74% of non-contact ACL injuries occur on the dominant leg in males (Brophy et al., 2010). Four sets of light gates (Smartspeed™, Fusion Sports, Australia) were used to ensure participants performed the activities at the correct angle. Light gate sensors and reflectors were positioned one meter from the ground. The first speed gate was positioned at 20% of the subject's maximum jump distance from the starting position. The force-plate was located a further 50% of the maximum jump distance beyond light gate 1. The other three light gates were positioned 2 meters from the centre of the force-plate at 45^0 to the left, straight ahead and 45^0 to the right. When the light beam of light gate 1 was broken, light gate 2, 3 or 4 was randomly activated, indicating the task to be performed. The order of the light gate activation was randomized using a random number generator (www.random.org). In the unanticipated condition, participants broke the light beam from gate 1 as they jumped through it. The time from breaking the light to the initial force-plate contact (pre-contact preparation time) was recorded. In contrast, in the anticipated condition, gate 1 was manually broken a minimum of 3 seconds before execution of the task, allowing the participant sufficient time to plan their task. Tasks altered between anticipated and unanticipated conditions. A custom made circuit (JTEC Ltd, Dundalk, Ireland) integrated the Smartspeed photorelay and the VICON motion system to trigger data capture with a delay of 0.012 seconds. For all pre-HIIP trials, a minimum of 1 minute resting period was given to minimize any potential effect of fatigue.

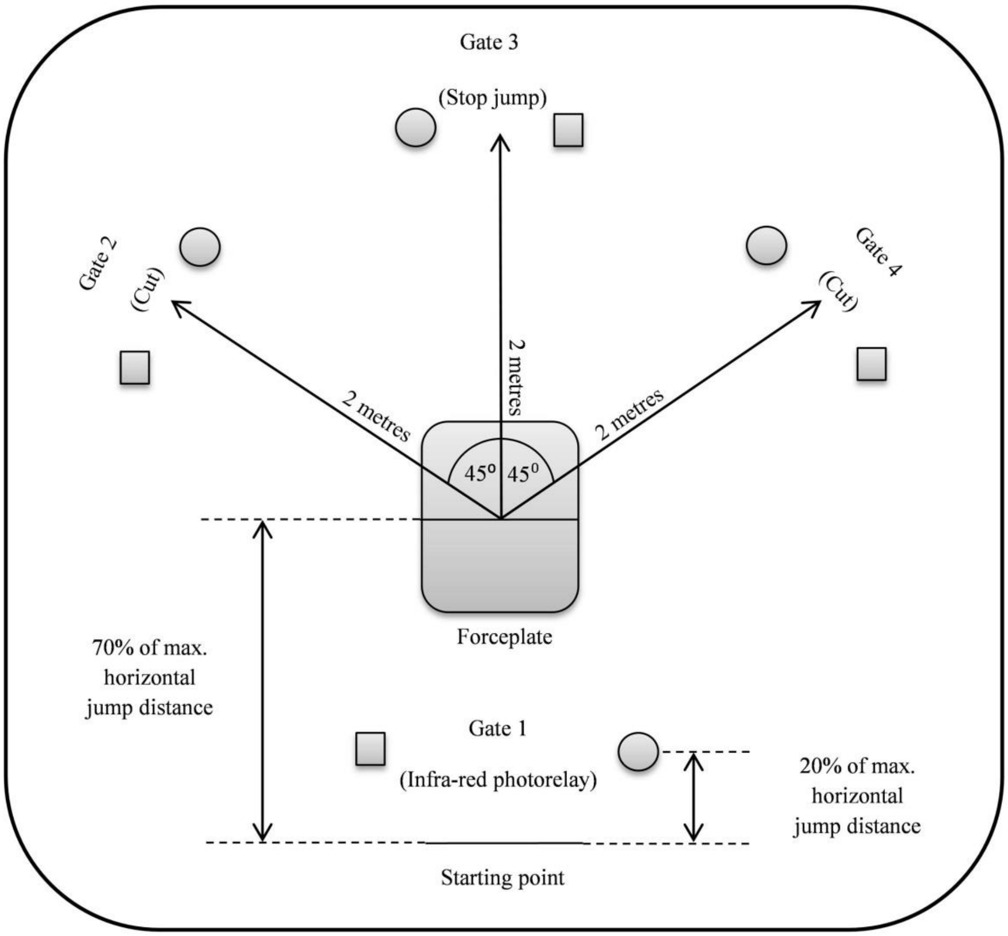

Figure 0.1 Experimental Set Up

The HIIP (Figure 3.1) was designed to mimic periods of high intensity, intermittent activity that occur in field-sports (Di Mascio and Bradley, 2013), leading to temporary fatigue (Krustrup et al., 2010). The protocol has previously been described and found to detrimentally affect dynamic balance in athletes (Whyte et al., 2015). Participants began the HIIP by sprinting forwards 5 m, cutting at a 90° angle and sprinting forwards another 5 m and then backpedalling 5 m. This was repeated 4 times with cutting direction alternating between left and right after which participants performed 10 two-legged jumps over 30 cm hurdles and 10 side-stepping exercises over the hurdles. Circuit time was recorded using infrared timing gates (model TC; Brower Timing Systems, Draper, UT, USA). Following completion of a circuit, the participant was given 30 seconds rest before repeating the circuit at maximum effort. The HIIP was discontinued when the participant reported a score of 18 on the Borg 6–20 rating of perceived exertion (RPE) scale (Borg, 1970). Heart rate was monitored throughout using a Polar heart rate monitor (model FT1; Polar Electro Inc., Lake Success, NY, USA). Participants performed the cutting tasks in the anticipated and unanticipated conditions within 30 seconds of completion of the HIIP. In order to maintain the effects from the HIIP, a circuit of the HIIP was repeated after every four cutting tasks.

6.3.3 Data Collection and Processing

Three dimensional trunk and lower extremity movements were recorded using a 12 camera Vicon motion analysis system (Oxford metrics Ltd., Oxford, United Kingdom) and the Vicon plug-in-gait marker set, which consisted of 16 lower limb (Kim et al., 2014) and 4 trunk markers (Gutierrez et al., 2003). To assist with pelvic marker identification, we placed an additional marker midway between the anterior and posterior superior iliac spines bilaterally. An AMTI force platform (BP-600900; Advanced Mechanical Technology Inc., Watertown, Massachusetts, USA) was use to record ground reaction force (GRF) data. Nexus VICON software (version 1.8.5; Vicon, Oxford, Great Britain) controlled simultaneous collection of motion and force data at 250Hz and used inverse dynamics to generate lower limb kinetic data. For this study, trunk, pelvic, hip, knee and ankle data were extracted during the weight acceptance phase, defined as the period from heel contact to first trough (minimum) in the vertical GRF (Besier et al., 2001; Cochrane et al., 2010; Dempsey et al., 2007). Kinetic measures (GRF and net hip, knee and ankle internal moments, and powers) were exported for analysis.

6.3.4 Statistical Analysis

Trunk, pelvic, and trunk on pelvic kinematics in the sagittal, frontal and transverse planes were exported for SPM analysis along with hip, knee and ankle kinematics. Likewise, hip, knee and ankle moments in the sagittal, frontal and transverse planes and resultant powers, as well as vertical, mediolateral and anterior-posterior GRFs, were also analysed. SPM (Pataky et al., 2013) was used to calculate the test statistic for each point of the examined vector. Each point of the vector was subject to a 2 way repeated measures ANOVA to determine the effect of HIIP (pre-HIIP *versus* post-HIPP), the state of anticipation (anticipated versus unanticipated) and any interaction effects. SPM provides a test statistic field (F value) and an evaluation of the significance of this field in a similar way to univariate analysis; the primary differences being that SPM considers vector covariance and field smoothness when calculating the test statistic (F value) and *P* value, respectively, over a range of points along the vector. This allows us to identify phases of significant differences rather than discrete, predetermined points. A randomised field theory correction was used to ensure that any significant findings were not down to chance (Adler, 2007; Pataky et al., 2013).Where interactions were found, a planned comparison between the anticipated crossover cut pre-HIIP and the unanticipated crossover cut post-HIIP was conducted. Effect sizes (Cohen's d) were calculated in a point-by-point matter and classified as: small 0.2 – 0.49, medium 0.5 - 0.79, and large > 0.8 (Pallant, 2010).

In order to assess the physiological effects of the HIIP, paired sample t-tests compared the first and final HIIP circuit completion times as well as heart rates at rest and post-HIIP. Paired sample t-tests were used to determine if the HIIP had an effect on approach velocities. Approach velocity was calculated by dividing the distance from gate 1 to the force plate (Figure 0.1) by the time from breaking the light to the initial force-plate contact (pre-contact preparation time). Data processing and statistical analyses were performed in MATLAB (R2012a, Math- Works Inc., USA). An alpha level was set a *priori* at 0.05 for all analysis.

6.4 Results

6.4.1 Physiological Analysis

The physiological effects of the HIIP are displayed in table 6.1. Participants completed on average 6.59±1.84 circuits before reporting a score of 18 on the RPE scale (Borg, 1970), with an average heart rate at completion of 184.4 (SD 5.6) (92.98 ± 2.66% of heart rate maximum) versus 64.8 (SD 5.6) (32.67 ± 2.86% of heart rate maximum) beats/min at rest ($p < 0.001$; Table 0.1). Circuit-completion times increased from the initial to the final circuit (mean 46.8 (SD 4.2) versus 49.9 (SD 4.3) seconds; $p < 0.001$; Table 0.1). There were no differences in approach velocity (mean 8.03 (SD 1.55) and 8.07 (SD 1.53) $m.s^{-1}$; $p = 0.87$) or pre-contact preparation time (mean 150 (SD 20) and 149 (SD 22) ms; $p = 0.76$) between pre- and post-HIIP.

Table 0.1 The effects of the high intensity, intermittent exercise protocol

	Pre HIIP	Post HIIP	***p* value**
Heart Rate (bpm)	64.77 ± 5.61	184.39 ± 6.29	<0.001
Time of circuit (s) (First vs final circuit)	46.82 ± 4.16	49.91 ± 4.26	<0.001
Pre-contact preparation time (ms)	150 ± 22	149 ± 20	0.76
Approach velocity ($m.s^{-1}$)	8.03 ± 1.55	8.07 ±1.53	0.87

6.4.2 Biomechanical Analysis

6.4.2.1 Interaction effect

No significant state of anticipation by HIIP interaction effects were observed.

6.4.2.2 Main effect for HIIP

Table 0.2 - Table 0.4 and Figure 0.2- Figure 0.6 display the main effects for HIIP on trunk and lower limb biomechanics during the crossover cutting manoeuvre. Performance of the crossover cutting manoeuvre post-HIIP resulted in less anterior pelvic tilt (P = 0.049, d = 0.33, 84-100%), hip flexion (P = 0.029, d = 0.26, 24-100%) and knee flexion (P < 0.001, d = 0.33, 75-100%) compared to the pre-HIIP cuts. Post-HIIP participants also exhibited smaller hip external rotator (P = 0.029, d = 0.27, 87-100%), and knee extensor (P < 0.001, d = 0.35, 26-100%) and internal rotator moments (P = 0.019, d = 0.35, 73-89%), and less power absorption at the knee (P < 0.001, d = 0.32, 43-77%) compared to pre-HIIP crossover cuts. No significant changes were observed at the ankle or for GRFs post-HIIP.

6.4.2.3 Main effect for anticipation

Table 0.3 and Table 0.4 and Figure 0.2- Figure 0.6 display the effects of anticipation on the biomechanics of the trunk and lower limb during the crossover cutting manoeuvre. Unanticipation resulted in less trunk side flexion ($p < 0.001$, d = 0.79, 1-100%), pelvic side flexion ($p < 0.001$, d = 0.70, 1-100%) and trunk on pelvic side flexion ($p = 0.034$, d = 0.26, 23-100%) in the direction of travel and a decrease in trunk flexion ($p = 0.002$, d = 0.30, 1-100%) and anterior pelvic tilt ($p < 0.001$, d = 0.33, 1-100%). Unanticipation also led to reduced hip flexion ($p = 0.049$, d = 0.18, 39-47%) and abduction ($p = 0.001$, d = 0.30, 1-100%), less knee varus ($p = 0.049$, d = 0.17, 25-28%), less ankle dorsiflexion ($p = 0.040$, d = 0.22, 62-81%), ankle eversion ($p = 0.005$, d = 0.20, 1-71%) and ankle external rotation ($p = 0.007$, d = 0.21, 1-66%). During unanticipated crossover cuts participants demonstrated greater hip abductor ($p < 0.001$, d = 0.33, 40-100%) and external rotator ($P = 0.013$, d = 0.42, 79-100%) moments, greater knee extensor ($p = 0.045$, d = 0.24, 95-100%) and valgus ($p < 0.001$, d = 0.29, 51-100%) moments compared with anticipated crossover cuts. The unanticipated condition gave rise to smaller ankle plantarflexor ($p = 0.035$, d = 0.26, 14-23%) and invertor ($p = 0.040$, d = 0.18, 51-59%) moments, less ankle power absorption ($p < 0.001$, d = 0.26, 72-100%) and a smaller lateral GRF ($p = 0.031$, d = 0.47, 1-8%; $p < 0.001$, d = 0.42, 28-65%; $p = 0.001$, d = 0.50, 76-100%).

Table 0.2 The effect of the HIIP on trunk and lower limb biomechanics during the weight acceptance phase of the crossover cutting manoeuvre

		Sagittal Plane				Frontal Plane				Transverse Plane			
		Effect (extent of effect)	**% of phase**	***p***	**d**	**Effect (extent of effect)**	**% of phase**	***p***	**d**	**Effect (extent of effect)**	**% of phase**	***p***	**d**
Trunk	Angles	NS				NS				NS			
Pelvis	Angles	Less anterior pelvic tilt (6.2%)	84-100%	0.049	0.33	NS				NS			
Trunk on pelvis	Angles	NS				NS				NS			
Hip	Angles	Less flexion (5.4%)	24-100%	0.029	0.26	NS				NS			
	Moments	NS				NS				Smaller external rotator (11.4%)	87-100%	0.029	0.27
Knee	Angles	Less flexion (6.8%)	75-100%	<0.001	0.33	NS				NS			
	Moments	Smaller extensor(28%)	26-100%	<0.001	0.35	NS				Smaller internal rotator(58%)	73-89%	0.019	0.35
Ankle	Angles	NS				NS				NS			
	Moment	NS				NS				NS			
GRF		NS				NS				NS			

NS = Non significant

Table 0.3 The effect of unanticipation on trunk and lower limb biomechanics during the weight acceptance phase of the crossover cutting manoeuvre

		Sagittal Plane				Frontal Plane				Transverse Plane			
		Effect (extent of effect)	**% of phase**	***p***	**d**	**Effect (extent of effect)**	**% of phase**	***p***	**d**	**Effect (extent of effect)**	**% of phase**	***p***	**d**
Trunk	Angles	Less anterior pelvic tilt (11.2%)	1-100%	0.009	0.30	Less side flexion in cut direction (224%)	1-100%	<0.001	0.79	NS			
Pelvis	Angles	Less anterior pelvic tilt (10.9%)	1-100%	<0.001	0.33	Less pelvic side flexion in cut direction (68%)	1-100%	<0.001	0.70	NS			
Trunk on pelvis	Angles	NS				Greater side flexion away from cut direction (62%)	23-100%	0.034	0.26	NS			
Hip	Angles	Less flexion (3.9%)	39-47%	0.049	0.18	Greater adduction(39.3%)	1-100%	0.001	0.30	NS			
	Moment	NS				Greater abductor (19%)	40-100%	<0.001	0.33	Greater external rotator (18.7%)	79-100%	0.013	0.42
Knee	Angles	NS				Less varus(11.9%)	25-28%	0.049	0.17	NS			
	Moment	Greater extensor(7.2%)	95-100%	0.045	0.24	Greater valgus (17%)	51-100%	<0.001	0.29	NS			
Ankle	Angles	Less dorsiflexion (37.4%)	62-81%	0.040	0.22	Less eversion (34.8%)	1-71%	0.005	0.20	Less external rotation (38%)	1-66%	0.007	0.21
	Moment	Smaller plantarflexor (42%)	14-23%	0.035	0.26	Smaller invertor (41.8%)	51-59%	0.040	0.18	NS			
						Smaller invertor (32%)	88-100%	0.031	0.47				
GRF		Smaller lateral GRF (42%)	1-8%	0.031	0.47	NS				NS			
		Smaller lateral GRF (37%)	28-65%	<0.001	0.42								
			76-100%	0.001	0.48								

NS = Non significant, GRF = ground reaction force

Table 0.4 The effect of the HIIP and anticipation on hip, knee and ankle resultant powers during the weight acceptance phase of the crossover cutting manoeuvre

Joint	Main effect for Anticipation				Main effect for Fatigue			
	Effect (extent of effect)	% of phase	*p*	d	Effect (extent of effect)	% of phase	*p*	d
Hip	NS				NS			
Knee	NS				Less power absorption (28.5%)	43-77%	<0.001	0.32
Ankle	Less absorption (14.6%)	79-97%	0.001	0.26	NS			

NS = Non significant

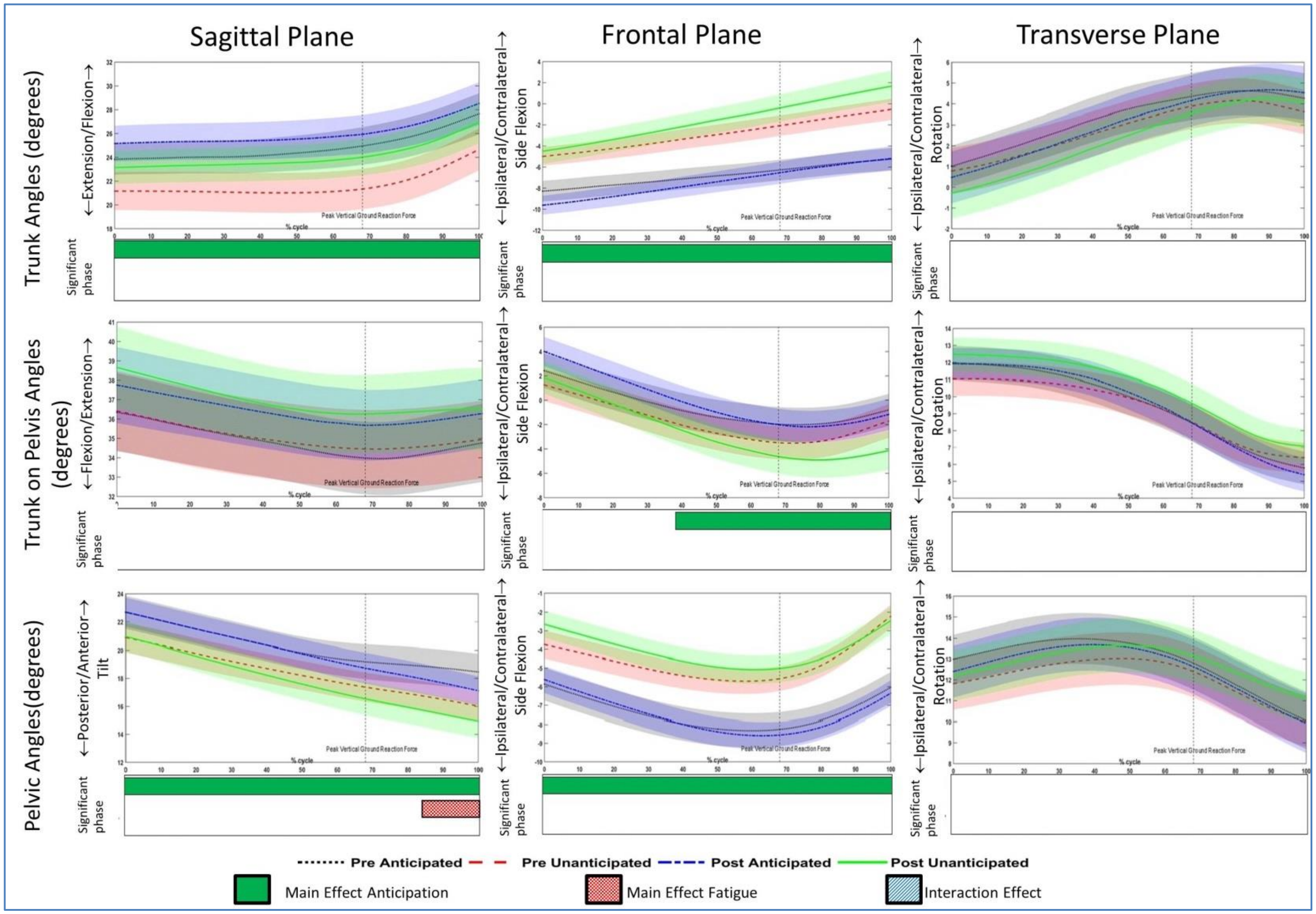

Figure 0.2 Trunk and pelvic kinematics during the weight acceptance phase of the crossover cutting manoeuvre

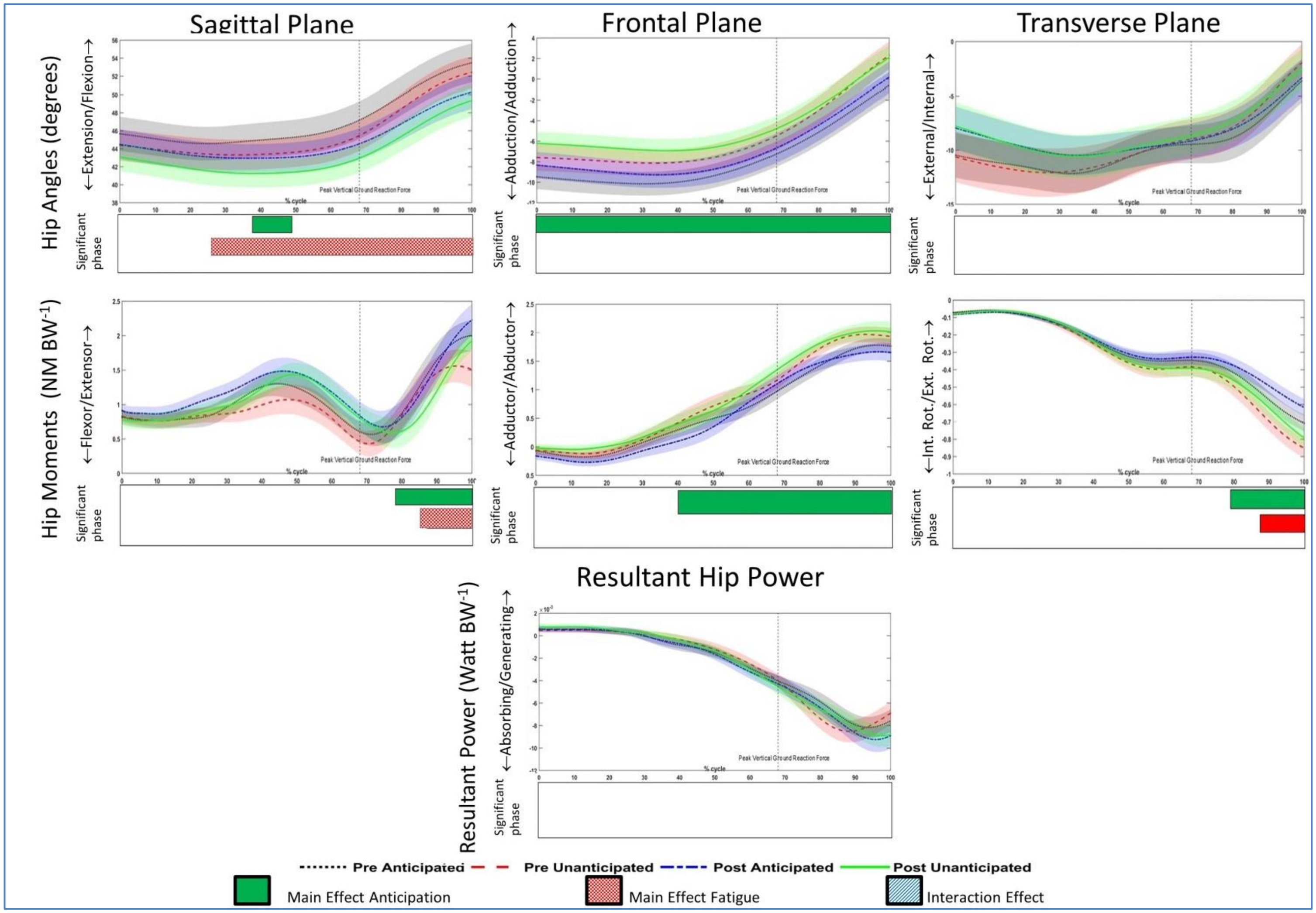

Figure 0.3 Hip biomechanics during the weight acceptance phase of the crossover cutting manoeuvre

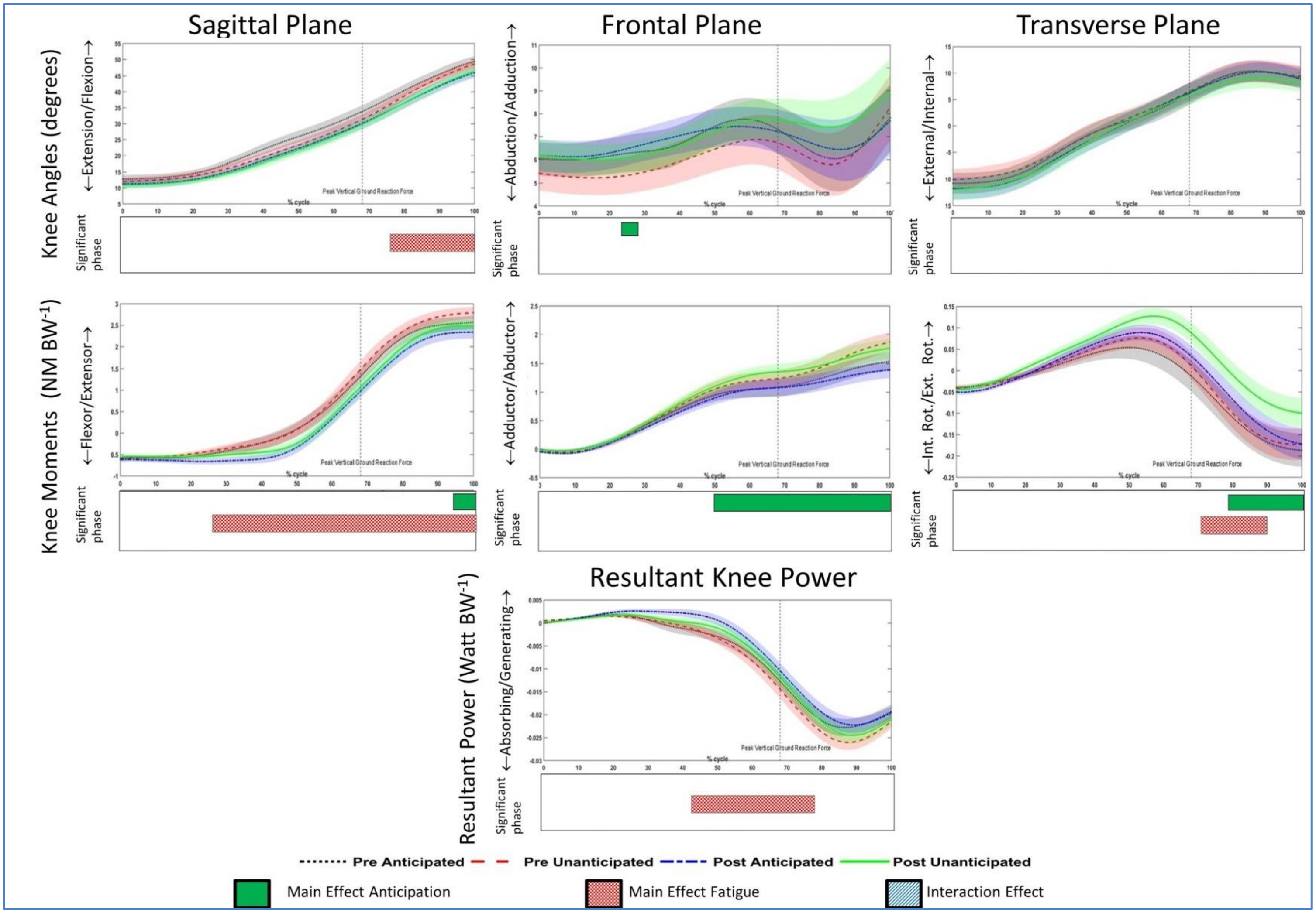

Figure 0.4 Knee biomechanics during the weight acceptance phase of the crossover cutting manoeuvre.

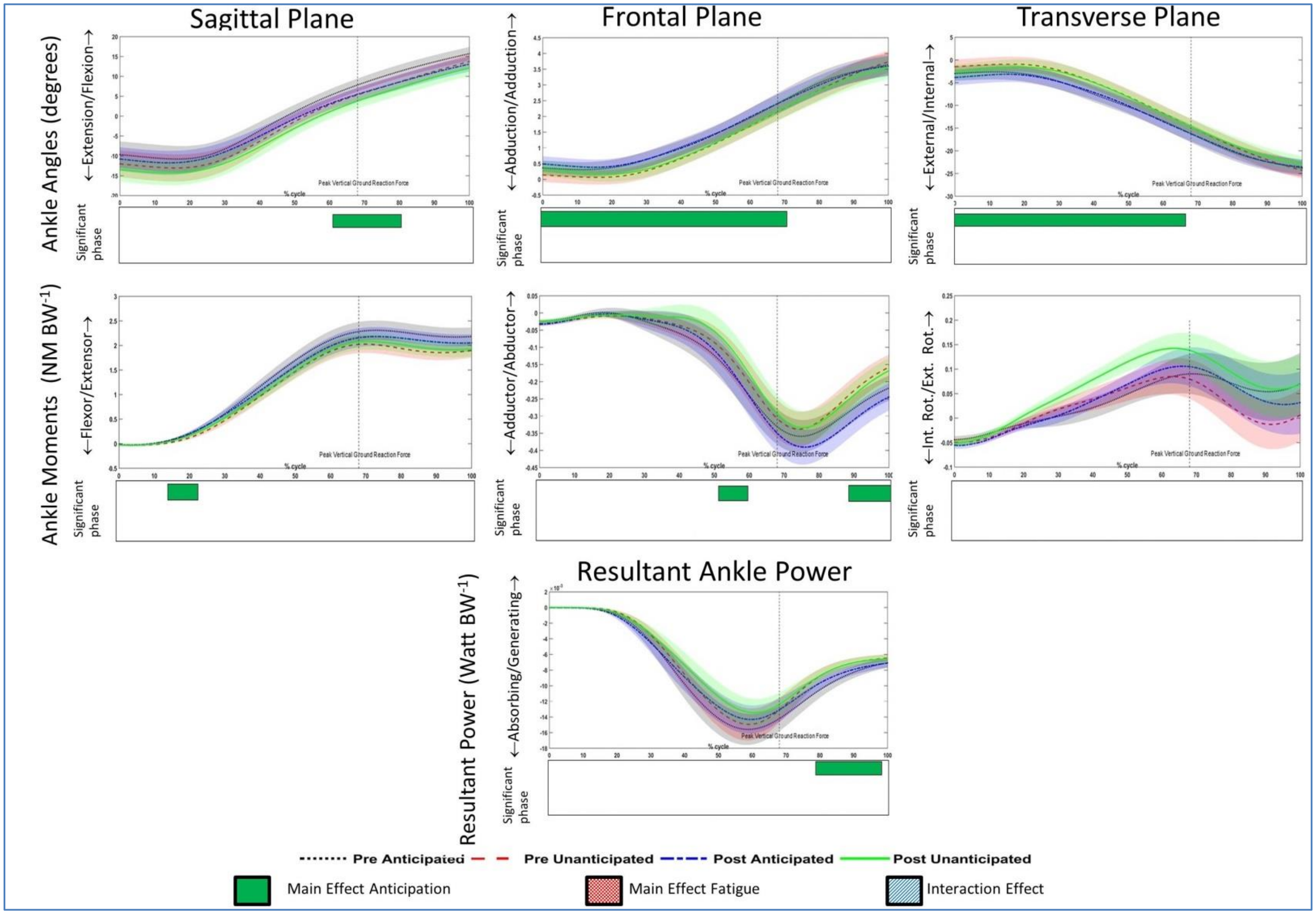

Figure 0.5 Ankle biomechanics during the weight acceptance phase of the crossover cutting manoeuvre

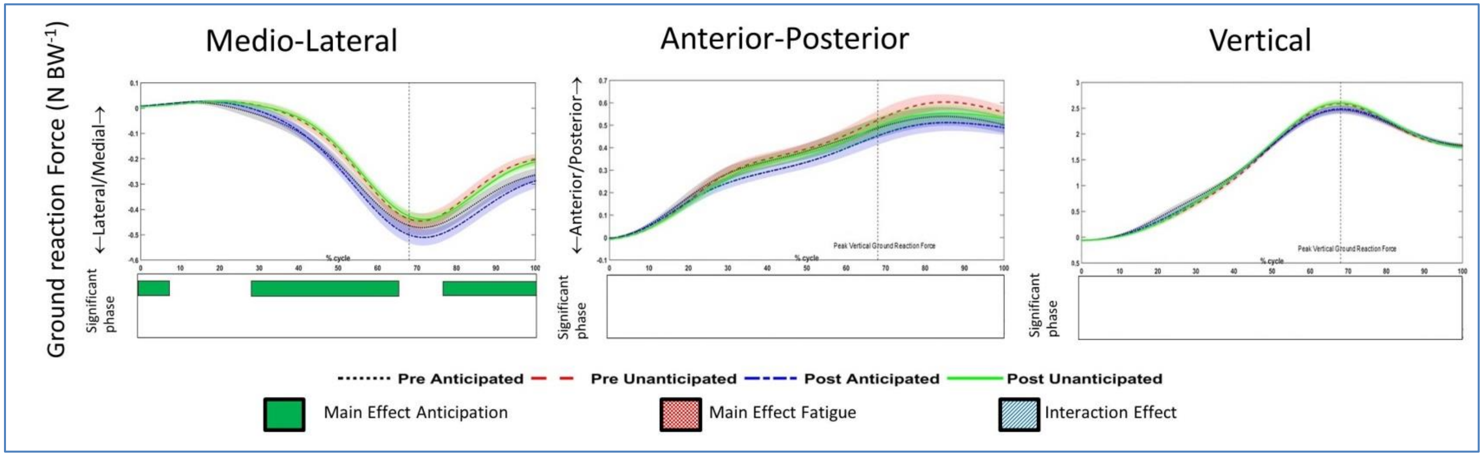

Figure 0.6 Ground reaction forces during the weight acceptance phase of the crossover cutting manoeuvre

6.5 Discussion

We hypothesized that there would be significant main effects for HIIP and state of anticipation, and interaction effects. The results of our study partially supported our hypotheses as demonstrated by significant main effects of anticipation (Table 0.3, Table 0.4, Figure 0.2- Figure 0.6). The strongest effects of unanticipation were observed at the trunk and pelvis in the frontal plane. Unanticipation resulted in potentially detrimental trunk and pelvic biomechanics, leading to frontal and transverse plane hip and sagittal plane knee modifications, with the potential to increase ACL strain. There were only small or trivial effects main effects for the HIIP (Table 0.3 and Table 0.4, Figure 0.2- Figure 0.6), predominantly in the sagittal plane, that may decrease ACL injury risk. The combination of fatigue and unanticipation did not increase the risk of ACL injury. No significant interactions between these conditions were found indicating that anticipation and HIIP did not combine to further increase alterations in the biomechanics of the crossover cutting manoeuvre and increase ACL injury risk. These findings enhance our understanding of the biomechanics of the crossover cutting manoeuvre and may contribute to research into ACL injury prevention programmes.

It appears that performance of unanticipated crossover cutting manoeuvres does not allow for the implementation of a pre-programmed neuromuscular strategy that would optimize technique and joint stability (Borotikar et al., 2008; Patla et al., 1999). Instead, participants must identify the stimulus, select and implement the correct neuromuscular programme (Schmidt, 2008) within the time constraints. Given the short pre-contact preparation time in the current study (Table 0.1), the integration of these processes will likely lead to potentially hazardous neuromuscular control patterns and biomechanics (Besier et al., 2001; Borotikar et al., 2008). This may explain the altered trunk and pelvic kinematics in the frontal plane observed in the current study throughout the entire weight acceptance phase, when the movement was unanticipated. Overcoming the centre of mass re-direction constraint is an essential component of crossover cutting manoeuvres (Patla et al., 1999). As the trunk and pelvis contains a large proportion of the body mass, the observed decrease in trunk and pelvic side flexion in the direction of the cut indicate that the challenge to re-orientate the centre of mass is greater in the unanticipated condition. The more extended trunk, pelvic and hip posture of participants may contribute to this challenge as it increases the height of the centre of mass. This posture may be adopted due to the nature of the unanticipated condition. Participants had to complete one of three potential tasks (crossover cut, side cut or stop jump) during the unanticipated condition. The short pre-contact preparation time (150 ms ± 20) was insufficient to allow implementation of a pre-programmed strategy for the task. As each task was randomly chosen, participants could not predict the unanticipated task. Therefore, participants may have adopted an upright posture to allow successful completion of any of the tasks, despite the resultant increased centre of mass re-direction challenge observed. The restricted ability to implement a pre-programmed strategy during the unanticipated condition and the ensuing extended posture is likely to cause a delay in the re-orientation of the centre of mass. This results in a longer time to peak lateral GRF during unanticipated crossover cutting manoeuvres (Kim et al., 2014) and may explain the decrease in lateral GRF during the weight acceptance phase observed in the current study.

Failure to overcome the centre of mass re-direction constraint has been well investigated (Jamison et al., 2012; Lee et al., 2013) in side cutting, but not in crossover cutting. During side cutting, the challenge to re-direct the centre of mass leads to altered trunk kinematics which directly affects the biomechanics of the lower limb (Donnelly et al., 2012; Jamison et al., 2012; Shimokochi et al., 2013) by displacing the centre of mass lateral to the supporting knee joint centre, and increasing the knee abduction angle (Hewett et al., 2010) and internal varus moment (Dempsey et al., 2007; Jamison et al., 2012), both of which predict ACL injuries (Hewett et al., 2005). In the current study on crossover cutting, the unanticipated condition resulted in decreased trunk and pelvic flexion side flexion in the direction of travel. If we examine the trunk position relative to the pelvis in particular, the unanticipated condition led to increased trunk side flexion away from the direction of the cut relative to the pelvis. These may have the opposite effect to that observed in unanticipated side cutting: it may displace the centre of mass medial to the hip and knee joint centres resulting in the observed modifications at the hip, knee and ankle. It specifically led to greater hip abductor and external rotator moments, greater knee extensor and valgus moments and smaller ankle plantarflexor and invertor moments, albeit with trivial to small effects. Similar to previous research (Kim et al., 2014) we found greater abductor moments throughout the weight acceptance phase during unanticipation, which may facilitate the stabilization and reorientation of the centre of mass and contribute to the increased internal knee valgus moments. In contrast to previous findings (Kim et al., 2014), participants in the current study generated greater hip external rotator moments during unanticipation, which contribute to the rotational control required to re-orientate the centre of mass in the direction of cut (Andrews et al., 1977). Adequate hip control is critical given that a decrease in hip abductor (Claiborne et al., 2006) and external rotator (Lawrence et al., 2008) strength are related to poor frontal plane knee control and are predictors of ACL injury (Khayambashi et al., 2015). The unanticipated condition resulted in increased knee extensor moment, as previously reported (Besier et al., 2001; Kim et al., 2014), which may lead to increased ACL strain (Markolf et al., 1995) particularly if combined with increased loading in the frontal and transverse planes (Oh et al., 2012; Shin et al., 2011). Greater internal knee valgus moment increases ACL strain (Oh et al., 2012) if coupled with an increased internal rotator moment. However, there was no increase in knee rotator moment in the current study. The increased hip external rotator moment observed may be essential to prevent increased ACL strain that would result from the combined increase in knee extensor, valgus and rotator moments. Any potential increase in ACL strain may be, at least partially, offset by the increased varus positioning of the knee throughout, which is theorized to minimize ACL injury risk (Cortes et al., 2014). Increased exposure to unanticipated conditions may improve athletes' technique during cutting activities (Kipp et al., 2013). This, in combination with the results of the current study, suggests that the effect of trunk and pelvic control exercises in unanticipated conditions on cutting technique should be investigated with a view to refining current injury prevention programmes.

We can be confident that fatigue was induced by the HIIP as participants reported high RPEs and heart rates similar to the 187 beats/min ± 9 reported in soccer players during match play (Table 0.1) (Krustrup et al., 2010). They also ran slightly longer distances than those reported in soccer (Mohr et al., 2003) (approximately 289.5 m *versus* 219 m ± 8 in 5 minutes). These findings, coupled with the increased circuit-completion times (Table 0.1), indicate that participants were fatigued by the HIIP. However, the HIIP did not lead to alterations in approach speed or pre-contact preparation times (Table 0.1). From this we can conclude that any post-HIIP biomechanical changes were due to the effects of the HIIP. However, performing the crossover cutting manoeuvre post-HIIP only resulted in trivial to small effects primarily in the sagittal plane. Participants performed the cross cutting manoeuvre post-HIIP in a more extended position with decreased anterior pelvic tilt and displayed risk factors for ACL injury which have been previously reported post fatigue, such as decreased hip and knee flexion (Cortes et al., 2014; McGovern et al., 2015; Potter et al., 2014). On one hand, the decreased knee extensor and internal rotator moments and subsequent power absorption (Table 0.4, Figure 0.4) may offset any potentially hazardous knee kinematics. However, there were no changes in the GRF post-HIIP. Therefore the decreased knee moments and power absorption may result in an increase in energy dissipation by the non-contractile tissues, such as the ACL, and thereby increase injury risk. The findings of the current study suggest that a short, high intensity exercise protocol does not notably increase potential ACL strain resulting from knee joint moments. However, it may lead to an increased risk of lower limb injury due to the increased loading of non-contractile tissues. Further studies investigating the effects of different exercise protocols on the biomechanics of the crossover cutting manoeuvre should be investigated.

We hypothesized that the biomechanical risk factors (such as increased knee valgus angle (Hewett et al., 2005), internal varus (Hewett et al., 2005), extensor (Markolf et al., 1995), and external rotator moments (Oh et al., 2012; Shin et al., 2011), and altered trunk kinematics (Jamison et al., 2012; Shimokochi et al., 2013) proposed to increase ACL loading would be greater during the performance of unanticipated cutting manoeuvres post-HIIP. The results of the current study do not support this hypothesis as no interaction was observed which is consistent with previous research (Collins et al., 2016; Khalid et al., 2015). However, it contradicts the studies of Borotikar et al., (2008) and McLean and

Samorezov (2009) which found that ACL injury risk is accentuated in a fatigued and unanticipated condition during side cutting (Borotikar et al., 2008; McLean and Samorezov, 2009)(Borotikar et al. 2008, McLean, Samorezov 2009) (Borotikar et al. 2008, McLean, Samorezov 2009). These conflicting findings may largely be explained by the different fatiguing protocols used and the fact that Borotikar et al. (Borotikar, Newcomer et al. 2008) and McLean and Samorezov (McLean, Samorezov 2009) analysed females during the side cutting manoeuvre while we examined males during the crossover cutting manoeuvre. These conflicting findings may largely be explained by the different fatiguing protocols used and the fact that Borotikar et al. (Borotikar, Newcomer et al. 2008) and McLean and Samorezov (McLean, Samorezov 2009) examined female participants. Firstly, the protocols used by Borotikar et al. (Borotikar, Newcomer et al. 2008) and McLean and Samorezov (McLean, Samorezov 2009) required participants to alter between squatting and cutting tasks until they were no longer able to complete body weight squats. Similar to Collins et al. (Collins et al. 2016) and Khalid et al. (Khalid et al. 2015), we utilized a fatiguing protocol that more closely mimicked the physiological demands of a field-sport leading to similar findings despite differences in the duration of the fatiguing protocol (60 mins (Collins et al., 2016) *vs* 8 mins approximately). Also, Borotikar et al. (Borotikar, Newcomer et al. 2008) and McLean and Samorezov (McLean, Samorezov 2009) both used discrete point analysis whereas continuous data analysis, which was used in the present study, has been reported to be superior in identifying biomechanical features that describe an outcome measure (Richter, O'Connor et al. 2014, Pataky, Robinson et al. 2013).Therefore, the results of the current study suggest that the effects of unanticipation and fatigue do not combine to increase risk factors for ACL strain and the possibility of fatigue failure (Lipps, Wojtys et al. 2013). This suggests that there is no added risk of performing unanticipated crossover cutting manoeuvres following a bout of high intensity exercise, as has been recently found with the side cutting manoeuvre (Collins et al., 2016, Khalid et al., 2015). It may also have implications for screening protocols as it suggests the effects of fatigue and unanticipation can be separately assessed. However, these suggestions should be treated with caution given the low level of analysis of ACL mechanisms of injury during crossover cutting (Cochrane et al., 2007), the HIIP may lead to increased dissipation of the GRF by non-contractile tissues and the fact the results of the current study only are applicable to male participants experienced in field-sports following a brief exercise protocol.

There are a number of limitations to the current study. Firstly, while analysing patterns of kinetics and kinematics rather than discrete points is more informative, it makes comparison with other studies difficult. Secondly, the HIIP is a relatively short protocol that may not replicate cumulative fatigue experienced during field-sports. Repetitions of the HIIP and subsequent trials interspersed with longer recovery periods may better simulate a game situation. Thirdly, the results of the current study may only be applicable to males given the gender differences observed in cutting biomechanics (Landry et al., 2008, Landry et al., 2009). Finally, the unanticipated condition may have reduced ecological validity as it was not match specific.

6.6 Conclusion

In conclusion, the HIIP may have resulted in a small decrease in ACL injury risk. Performance of unanticipated crossover cutting manoeuvre resulted in potentially detrimental trunk and pelvic biomechanics from an injury perspective. This led to frontal and transverse plane hip and sagittal plane knee modifications, with the potential to increase ACL strain. The combination of fatigue and unanticipation did not increase the risk of ACL injury. Future research should directly investigate the effect of trunk and hip control exercises in unanticipated conditions to determine their effect on the biomechanics of the crossover cut with a view to refining ACL injury prevention programmes.

6.7 Link to Chapters 7 and 8

It has been proposed that performance of unanticipated high risk activities, such as cutting, in a fatigued state may place an athlete at an elevated risk of ACL injury. Study 4 did not find that the performance of unanticipated crossover cutting in a HIIP-induced fatigue state resulted in an elevated biomechanical risk of ACL injury. It was also found that HIIP-induced fatigue alone did not elevate risk. However, study 4 demonstrated that unanticipated condition resulted in greater knee loading which may increase the risk of ACL injury. The unanticipated condition resulted in altered trunk and pelvic kinematics which may have contributed to the greater hip and knee joint loading observed. As athletes perform crossover and side cutting regularly during athletic activities, the suggestion that performance of unanticipated high risk activities, such as cutting, in a fatigued state may place an athlete at an elevated risk of ACL injury also applies to side cutting. Therefore, it is equally important to investigate the effects of unanticipation and fatigue on side cutting, which is conducted in study 5 (chapter 7).

The results of the current study, study 4, demonstrate that unanticipation results in altered trunk and pelvic biomechanics and increased hip and knee joint loading compared with the unanticipated condition. Therefore, improved hip and trunk control, particularly in response to unanticipated conditions, may improve the biomechanics of crossover cutting activities in terms of ACL injury risk. Therefore, study 6 (chapter 8) will investigate the effect of a trunk and hip control exercise programme on the biomechanics of anticipated and unanticipated side and crossover cutting.

Chapter 7 An Investigation of the Effects of High Intensity, Intermittent Exercise and Unanticipation on Trunk and Lower Limb Biomechanics during a Side Cutting Manoeuvre using Statistical Parametric Mapping

Study 5

Study 5: An investigation of the effects of high intensity, intermittent exercise and unanticipation on trunk and lower limb biomechanics during a side cutting manoeuvre using statistical parametric mapping

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STATEMENT OF CONTRIBUTION: Kieran Moran was the research supervisor for this study. Siobhan O'Connor and Chris Richter assisted in the data collection and analysis.

7.1 Abstract

Anterior cruciate ligament (ACL) injuries frequently occur during side cutting manoeuvres when fatigued or reacting to the sporting environment. Trunk and hip biomechanics are proposed to influence ACL loading during these activities. However, the effects of fatigue and unanticipation on the biomechanics of the kinetic chain may be limited by traditional discrete point analysis. We recruited twenty-eight male, varsity, Gaelic footballers (21.7 ± 2.2 years; 178.7 ± 14.6m; 81.8 ± 11.4kg) to perform anticipated and unanticipated side cutting manoeuvres pre- and post- a high intensity, intermittent exercise protocol (HIIP). Statistical parametric mapping (repeated-measures ANOVA) identified differences in phases of trunk and stance leg biomechanics during weight acceptance. Unanticipation resulted in less trunk flexion ($p < 0.001$) and greater side flexion away from the direction of cut ($p < 0.001$). This led to smaller (internal) knee flexor and greater (internal) knee extensor ($p = 0.002 – 0.007$), hip adductor ($p = 0.005$) and hip external rotator ($p = 0.007$) moments. The HIIP resulted in increased trunk flexion ($p < 0.001$) and side flexion away from the direction of cut ($p = 0.038$) resulting in smaller (internal) knee extensor moments ($p = 0.006$). One interaction effect was noted demonstrating greater hip extensor moments in the unanticipated condition post-HIIP ($p = 0.025$). Results demonstrate that unanticipation resulted in trunk kinematics considered an ACL injury risk factor. A subsequent increase in frontal and transverse plane hip loading and sagittal plane knee

loading was observed, which may increase ACL strain. Conversely, HIIP-induced trunk kinematic alterations resulted in reduced sagittal plane knee, and subsequent ACL, loading. Therefore adequate hip and knee control is important during unanticipated side cutting manoeuvres.

7.2 Introduction

Noncontact ACL injuries in males commonly occur in field sports with intermittent bursts of high intensity exercise and rapid changes of direction (Hootman et al., 2007). They typically occur during the weight acceptance (WA) phase (Koga et al., 2010) of side cutting manoeuvres (Walden et al., 2015) when the trunk and pelvis rotate around the stance limb (Andrews et al., 1977). This may be due to the combination of internal knee extensor (Weinhandl et al., 2013), varus and external rotator moments which significantly increase ACL strain and injury risk (Markolf et al., 1995; Oh et al., 2012; Shin et al., 2011). Notably, ACL injury rates are up to 20 times higher in matches than training (Larruskain et al., 2017). As they often occur in response to the sporting environment (Walden et al., 2015) and as injury occurrence increases as the first and second halves progress in soccer (Ekstrand et al., 2011), unanticipation (Kim et al., 2014; Weinhandl et al., 2013) and fatigue (Borotikar et al., 2008; McLean and Samorezov, 2009; Shultz et al., 2015) have been proposed as ACL injury risk factors.

Laboratory based studies have found that biomechanical risk factors for ACL injury are more evident during unanticipated than anticipated side cutting (Besier et al., 2001; Borotikar et al., 2008; Mornieux et al., 2014). This may be due to the altered trunk position observed during unanticipated side cutting (Mornieux et al., 2014; Patla et al., 1999) which is directly associated with increased knee joint loading (Donnelly et al., 2012; Jamison et al., 2012; Shimokochi et al., 2013). Furthermore, deficits in trunk control (Zazulak et al., 2007) and hip strength (Khayambashi et al., 2015) predict ACL injury risk highlighting the importance of trunk and hip control in ACL injury prevention programmes and rehabilitation post-surgical repair. However, the association between trunk kinematics and risk factors for ACL injury has not been sufficiently investigated (Shultz et al., 2015).

Biomechanical variables associated with non-contact ACL injury risk become more evident during side cutting manoeuvres following short-term exhaustive protocols (Borotikar et al., 2008; Tsai et al., 2009). However, there is a lack of research investigating the effects of temporary fatigue, which commonly occurs during field-sports after bouts of high intensity intermittent exercise (Knicker et al., 2011). In addition, the combination of unanticipation and fatigue results in increased knee loading during side cutting in short term exhaustive protocols (Borotikar et al., 2008; McLean and Samorezov, 2009) but it has not been found following protocols which more closely mimic temporary fatigue evident in field-sports (Collins et al., 2016; Khalid et al., 2015). In summary, to date our understanding ACL injuries is limited by the fact that the combined effects of fatigue and unanticipation on trunk, pelvic and ankle biomechanics during side cutting manoeuvres have not been investigated.

In addition, the majority of previous research has used discrete point analysis (DPA), which may limit our understanding of the combined effects of fatigue and unanticipation on side cutting manoeuvres (Shultz et al., 2015). DPA extracts discrete or singular measures of a particular variable (such as peak knee abductor moment). This can over simplify the original highly multivariate datasets (Pataky et al., 2013) leading to the analysis of less than 5% of the data (Richter et al., 2014). In contrast, statistical parametric mapping (SPM) analyses the whole kinematic and kinetic waveform (Pataky et al., 2013). It also negates the need to preselect discrete measures to be analyzed, which is a potential source of bias in DPA (Pataky et al., 2015). Finally, SPM avoids the selection of measures that are not always functionally comparable across all subjects due to temporal variations, which may result in false conclusions (Richter et al., 2014). For these reasons, SPM will improve our understanding of the biomechanics (Pataky et al., 2013; Richter et al., 2014) of the trunk, pelvis and lower limb during the weight acceptance phase of the side cutting manoeuvre.

The aims of this study were as follows. To investigate the effects of (a) a high intensity, intermittent exercise protocol (HIIP) (pre-HIIP versus post-HIIP), and (b) the state of anticipation (anticipated versus unanticipated condition), on the biomechanical patterns of the kinetic chain during the weight acceptance phase of the side cutting manoeuvre. It was hypothesized that there would be (a) significant main effects for the post-HIIP and unanticipated conditions and (b) no significant interaction between the effects of the HIIP and unanticipation.

7.3 Methods

7.3.1. Experimental Approach to the Problem

A repeated measures design was used to assess the effect of the independent variables of anticipation (unanticipated versus anticipated) and high intensity, intermittent exercise (pre-HIIP versus post-HIIP) on the biomechanics of the WA phase of the side cutting manoeuvre. The HIIP used in the current study results in impaired dynamic balance (E. Whyte et al., 2015) which is an important component in unanticipated cutting manoeuvres. Subjects were required to attend a familiarization session between 2-7 days prior to the data collection session in a university biomechanics laboratory.

7.3.2 Participants

A power analysis using previous findings (Borotikar et al., 2008), revealed a minimum requirement of 19 single-sex subjects to achieve a 95% statistical power (alpha level > 0.05). To allow for potential dropout, twenty eight male, collegiate Gaelic footballers participated in this study (age, 21.71 ± 2.16 years; height 178.71 ± 14.64m; body mass 81.82 ± 11.44kg). Gaelic football is a high intensity, contact field sport requiring athletes to run, sprint, jump, catch, kick and perform multiple, rapid changes of direction (Murphy et al., 2012). Players cover on average almost 9000 m during a game with 1500 m of that at high speed running. Players also perform 181 accelerations on average during the course of a game (Malone et al., 2017). Although officially an amateur sport, Gaelic football training and match schedules and intensities are similar to professional sports (Murphy et al., 2012). The collegiate match injury rate is 25.1 per 1000 athlete exposures (AE) (O'Connor et al., 2016) which is higher than collegiate rugby (22.5 injuries per 1000 AE) (Kerr et al., 2008) and soccer (18.8 injuries per 1000 AE) (Agel et al., 2007). 72% of injuries are lower limb injuries (O'Connor et al., 2016) with an ACL injury incidence of 0.14 per 1000 hours (Murphy et al., 2012) which is up to twice that reported in soccer (Larruskain et al., 2017). Subjects were required to be injury free with no history of lower limb ligamentous reconstructive surgery and participating in Gaelic football on at least three occasions per week. All subjects provided written informed consent, and the study was approved by the local Research Ethics Committee.

7.3.2 Experimental Procedure

During the familiarization session, subject height, weight, leg dominance, leg length, and knee and ankle widths were recorded (Table 7.1). Following this, the maximum standing long jump distance was assessed. Subjects were permitted to use their arms to assist with the test and the original test (Fernandez-Santos et al., 2015) was modified as subjects were instructed to "begin by standing on two feet and then jump as far forwards as you can, landing on your dominant (preferred kicking) foot. The best of three jumps was taken as their maximum jump distance (2.36 ± 0.30m). In order to normalize the effort of the side cutting manoeuvre, participants jumped 70% of their maximum jump distance and then performed the side cut (Whyte et al., 2018). After this, side cutting to 45°, stop jumping and cross cutting to 45° were explained and demonstrated to the subjects. The angle of side cutting manoeuvre was 45°as the majority of non-contact ACL injuries occur during a side cut between 30-60°^{0}(Walden et al., 2015).The stop jump task was included in the experimental procedure to limit the subjects' ability to predict the unanticipated cutting manoeuvres. Subjects began 70% of their maximum jump distance away from the centre of the force platform, then jumped forwards to the centre of the force platform and performed the manoeuvre. This standardized approach (Whyte et al., 2018) was implemented to minimize alterations in approach speed, a limiting factor in previous side cutting manoeuvres (Brown et al., 2014), and potential HIIP-induced alterations in maximal jump distance. Only data for the side cutting was analysed for this study. Subjects practiced these for a minimum of ten minutes until they were comfortable in the execution of the manoeuvres. Subjects were also instructed in, and practiced a minimum of five circuits of, the HIIP.

Table 7.1 Participant characteristics

Age (years)	21.7 ± 2.2
Height (cm)	178.7 ± 14.6
Leg length (cm)	93.2 ± 6.0
Weight (kg)	81.8 ± 11.4
Years of practice/competition at elite level*	4.8 ± 2.1
Number of subjects right leg dominant	23
Number of subjects left leg dominant	5

*Elite level = Intercounty and/or collegiate level

Baseline heart rate after ten minutes of quiet sitting was collected during the data collection session using a Polar heart rate monitor (model FT1; Polar Electro Inc., Lake Success, NY, USA). Subjects warmed-up by jogging for five minutes through the HIIP circuit and then dynamically stretching major lower limb muscles.

7.3.2.1 Side cutting manoeuvre

Subjects performed the side cutting manoeuvre on their preferred kicking leg as 74% of non-contact ACL injuries occur on this limb in males (Brophy et al., 2010). Correct angle of cut was ensured by using four sets of light gates (Smartspeed™, Fusion Sports, Australia) with light gate sensors and reflectors positioned 1 meter (m) from the ground (Figure 0.1). Gate 1 was positioned at 20% of the subject's maximum jump distance from the starting position to prevent any arm movements activating light gate 1 during unanticipated manoeuvres and to provide sufficient time for the subjects to complete the manoeuvre. The other three positioned at 2 meters distance from the centre of the force plate at 45^0 to the left, straight ahead and 45^0 to the right. The manoeuvre to be performed was indicated by the activation of light gates 2, 3 or 4 the order of which was randomized using a random number generator (www.random.org) when gate 1 was broken. Manoeuvres altered between anticipated and unanticipated conditions. In order to allow subjects to plan the anticipated side cutting manoeuvre, gate 1 was manually broken a minimum of three seconds before execution of the manoeuvre. In the unanticipated condition, light gate 2, 3 or 4 was activated when a subject jumped through gate 1. VICON motion data capture was also triggered as the subject jumped through gate 1 by using a custom made integration circuit (JTEC Ltd, Dundalk, Ireland) with a consistent delay of 0.012 seconds. The time from breaking gate 1 to the initial force plate contact was called the pre-contact preparation time and recorded. Subjects rested for a minimum of 1 minute to minimize any potential fatiguing effect during pre-HIIP trials.

7.3.2.2 The high intensity, intermittent exercise protocol

The HIIP has been previously described (Whyte et al., 2015). Subjects were requested to perform the HIIP at maximal effort. Each HIIP circuit began with subjects sprinting forwards 5 meters (m), performing a 90° change of direction and another 5m forward sprint before back-pedalling 5m. This was repeated 4 times. Subjects then performed 10 2-legged jumps over 30cm hurdles and 10 side stepping exercises. Finally subjects completed 4 5m lateral shuffles. Subjects were given a 30 second break before repeating the circuit. The HIIP was discontinued when subjects reported 18 on the Borg 6–20 rating of perceived exertion (RPE) scale (Borg, 1970). Heart rate was monitored throughout. HIIP circuit time was recorded using infrared timing gates (model TC; Brower Timing Systems, Draper, UT, USA).

7.3.2.3 Data collection and processing

A 12 camera Vicon Motion analysis system (Oxford metrics Ltd., Oxford, Great Britain) and the Vicon plug-in-gait marker set recorded three dimensional trunk and lower extremity movements. The plug-in-gait marker set consisted of 16 lower limb (Kim et al., 2014) and 4 trunk markers place at the spinous processes of C7 and T10, the sternal notch and xiphisternum with two additional pelvic markers placed midway between the anterior and posterior superior iliac spines bilaterally. Ground reaction force (GRF) data was recorded using an AMTI force platform (2000Hz) (BP-600900; Advanced Mechanical Technology Inc., Watertown, Massachusetts, USA.). Simultaneous collection of motion and force data was controlled at 250Hz using Nexus VICON software (version 1.8.5; Vicon, Oxford, Great Britain). Lower limb kinetic data were generated using inverse dynamics. Only the WA phase was analysed, defined as the period from initial contact to first trough in vertical GRF (Besier et al., 2001). For this study, trunk, pelvic, hip, knee and ankle data were extracted. Kinetic measures (GRF and net, internal hip, knee and ankle moments and work) were exported for analysis and normalized to body mass. Negative joint work values represent energy absorption by the muscle-tendon unit and are associated with eccentric muscular activity (Winter, 2009).

7.3.3 Statistical Analyses

The effects of the independent variables (HIIP and anticipation) on the dependent variables (Table 7.2) were assessed using statistical parametric mapping (SPM). In contrast to the traditional method of taking singular measures from dependent biomechanical variables to represent intricate human biomechanics, SPM analyses the entire dependent variable. This may afford us a greater understanding of complex biomechanics, such as the side cutting manoeuvre, and the effects of anticipation and a HIIP thereon (Shultz et al., 2015). SPM (Pataky et al., 2013) calculated the test statistic for each data point of each examined variable. The effects of HIIP (pre-HIIP versus post-HIIP), the state of anticipation (anticipated versus unanticipated) and any interaction effects were determined by completing a 2 way repeated measures ANOVA on each data point. Similar to univariate analysis, SPM provides a test statistic field (F value) and an evaluation of the significance of this field. SPM also takes variable covariance and field smoothness over a range of points along the variable into account when calculating the test statistic (F value) and *P* value allowing the identification of significant phases. Following this, a randomized field theory correction ensured that significant findings were not down to chance (Adler, 2007;

Pataky et al., 2013). A planned comparison between the anticipated crossover cut pre-HIIP and the unanticipated crossover cut post-HIIP was conducted where interactions existed. Partial eta squared effect sizes (η^2) were calculated in a point-by-point matter and classified as small 0.01- 0.06; medium 0.06 – 0.14; and large >0.14 (Pallant, 2010).

Paired sample t-tests were used to assess the physiological effects of the HIIP by comparing the first and final HIIP circuit completion times as well as heart rates at rest and post-HIIP. Paired sample t-tests also investigated an effect on approach velocities (calculated by dividing the distance from gate 1 to the force plate (Figure 0.1) by the pre-contact preparation time). Data processing and statistical analyses were performed in MATLAB (R2015a, Math- Works Inc., USA). An alpha level was set a *priori* at 0.05 for all analysis.

Table 7.2 Dependent variables included in the statistical analysis

Variable	Sagittal	Frontal	Transverse
Trunk Kinematics	Flexion/Extension	Ipsi/contralateral flexion	Ipsi/contralateral rotation
Trunk on Pelvis Kinematics	Flexion/Extension	Ipsi/contralateral flexion	Ipsi/contralateral rotation
Pelvic Kinematics	Anterior/Posterior Tilt	Ipsi/contralateral flexion	Ipsi/contralateral rotation
Hip Kinematics	Flexion/Extension	Abduction/Adduction	Internal/External rotation
Hip Moments	Flexor/Extensor	Abductor/Adductor	Internal/External rotator
Hip Work	Eccentric/Concentric Flexor/Extensor	Eccentric/Concentric Abductor/Adductor	Eccentric/Concentric Internal/External rotator
Knee Kinematics	Flexion/Extension	Abduction/Adduction	Internal/External rotation
Knee Moments	Flexor/Extensor	Abductor/Adductor	Internal/External rotator
Knee Work	Eccentric/Concentric Flexor/Extensor	Eccentric/Concentric Abductor/Adductor	Eccentric/Concentric Internal/External rotator
Ankle Kinematics	Dorsi/Plantarflexion	Abduction/Adduction	Internal/External rotation
Ankle Moments	Dorsi/Plantarflexor	Abductor/Adductor	Internal/External rotator
Ankle Work	Eccentric/Concentric Flexor/Extensor	Eccentric/Concentric Abductor/Adductor	Eccentric/Concentric Internal/External rotator

7.4 Results

7.4.1 Physiological Analysis

Subjects completed 6.59 ± 1.84 circuits of the HIIP, with heart rates at completion of 184.4 ± 5.6 beats/min versus 64.8 ± 5.6 beats/min at rest. Circuit-completion times increased from the initial to the final circuit (46.8 seconds ± 4.2 versus 49.9 seconds ± 4.3; $p < 0.001$). There was no difference in approach velocities pre- and post-HIIP (8.15 $m.s^{-1}$ ± 1.52 and 8.39 $m.s^{-1}$ ± 1.73 respectively; p=0.15) or pre-contact preparation times (0.14 ms ± 0.02 and 0.12 ms ± 0.02; $p = 0.14$).

7.4.2 Biomechanical Analysis

For concise reporting of results, only significant findings of the extended key phases are reported below with the respective percentage of the weight acceptance phase over which the difference occurred.

7.4.2.1 Interaction effect

There was only one interaction effect with trivial effect size which demonstrated that the combined effects of the unanticipated and post-HIIP conditions resulted in an increase in hip extensor moment (p=0.025, η^2=0.018, 71-80%).

7.4.2.2 Main effect for HIIP on the kinematics of the side cutting manoeuvre

Performance of the side cutting manoeuvre post-HIIP resulted in greater trunk flexion ($p < 0.001$, $\eta^2 = 0.038$, 1-100%), and greater trunk side flexion away from the direction of cut ($p = 0.038$, $\eta^2 = 0.017$, 1-88%) and relative to the pelvis ($p = 0.039$, $\eta^2 = 0.022$, 1-75%) compared to the pre-HIIP cuts. It also resulted in less anterior pelvic tilt ($p = 0.049$, $\eta^2 = 0.008$, 81-100%), decreased hip flexion ($p = 0.033$, $\eta^2 = 0.019$, 27-100%), decreased knee flexion ($p = 0.037$, $\eta^2 = 0.013$, 73-100%) (Table 7.3 and Figure 7.1).

7.4.2.3 Main effect for HIIP on the kinetics of the side cutting manoeuvre

Smaller internal knee extensor moments (p = 0.006, η^2 = 0.013, 70-98%) and greater internal ankle external rotator moment (p = 0.045, η^2 = 0.050, 1-6%) were observed post-HIIP compared with pre-HIIP (Table 7.3 and Figure 7.1).

7.4.2.4 Main effect for anticipation on kinematics of the side cutting manoeuvre

Unanticipation resulted in greater trunk side flexion (p < 0.001, η^2 = 0.142, 1-100%) and pelvic side flexion (p < 0.001, η^2 = 0.143, 1-100%) away from the direction of cut, and a decrease in both trunk flexion (p < 0.001, η^2 = 0.051, 1-100%) and anterior pelvic tilt (p < 0.001, η^2 = 0.072, 1-100%). There was also greater trunk rotation in the direction of cut (p = 0.049, η^2 = 0.027, 77-100%) and less trunk rotation away from the direction of cut relative to the pelvis (p = 0.045, η^2 = 0.32, 50-100%). At the hip joint, performance of the unanticipated side cutting manoeuvre led to less hip flexion (p = 0.049, η^2 = 0.008, 1-16%), greater abduction (p = 0.022, η^2 = 0.050, 1-91%), less external rotation (p = 0.043, η^2 = 0.005, 1-32%), followed by greater internal rotation (p = 0.048, η^2 = 0.009, 82-100%).

During unanticipated side cuts, participants demonstrated greater knee flexion (p = 0.045, η^2 = 0.013, 84-100%) and greater knee internal rotation (p = 0.045, η^2 = 0.004, 51-69%) compared with the anticipated condition. At the ankle, the unanticipated condition gave rise to greater plantarflexion (p < 0.001, η^2 = 0.015, 15-100%) (Table 7.4 and Figure 7.2).

Table 7.3 Variables significantly affected by a high intensity, intermittent exercise protocol (post versus pre) during the weight acceptance phase of a side cutting manoeuvre

Variable	**Sagittal**	**Frontal**	**Transverse**
Trunk Kinematics	↑ Flexion	↑ Ipsilateral flexion	
Trunk on Pelvis Kinematics		↑ Ipsilateral flexion	
Pelvic Kinematics	↓ Anterior Tilt		
Hip Kinematics	↓ Flexion		
Knee Kinematics	↓ Flexion		
Knee Moments	↓ Extensor		
Ankle Moments			↑ External rotator

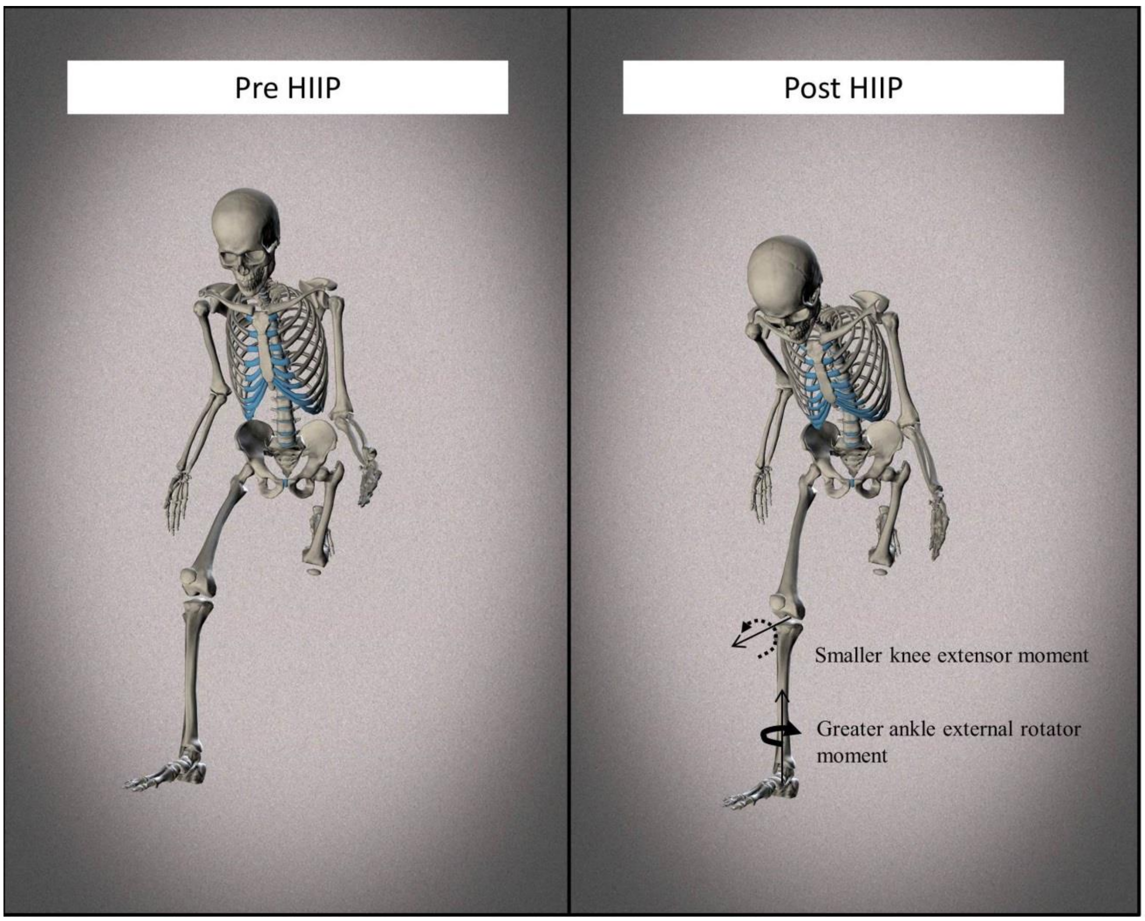

Figure 7.1 The effect of the HIIP on the biomechanics of the weight acceptance phase of a side cutting manoeuvre

Table 7.4 Variables significantly affected by anticipation (unanticipated versus anticipated) during the weight acceptance phase of a side cutting manoeuvre

Variable	Sagittal	Frontal	Transverse
Trunk Kinematics	↓ Flexion	↑ Ipsilateral flexion	↑ Contralateral rotation
Trunk on Pelvis Kinematics			↓ Contralateral rotation relative to pelvis
Pelvic Kinematics	↓ Anterior Tilt	↑ Ipsilateral flexion	
Hip Kinematics	↓ Flexion	↑ Abduction	↑ Internal rotation
Hip Moments		↑ Adductor	↑ External rotator
Hip Work			↑ Eccentric External rotator
Knee Kinematics	↑ Flexion		↑ Internal rotation
Knee Moments	↑ Extensor		↓ External rotator
Knee Work	↑ Eccentric Extensor		
Ankle Kinematics	↑ Plantarflexion		
Ankle Moments	↓ Plantarflexor		↓ External rotator

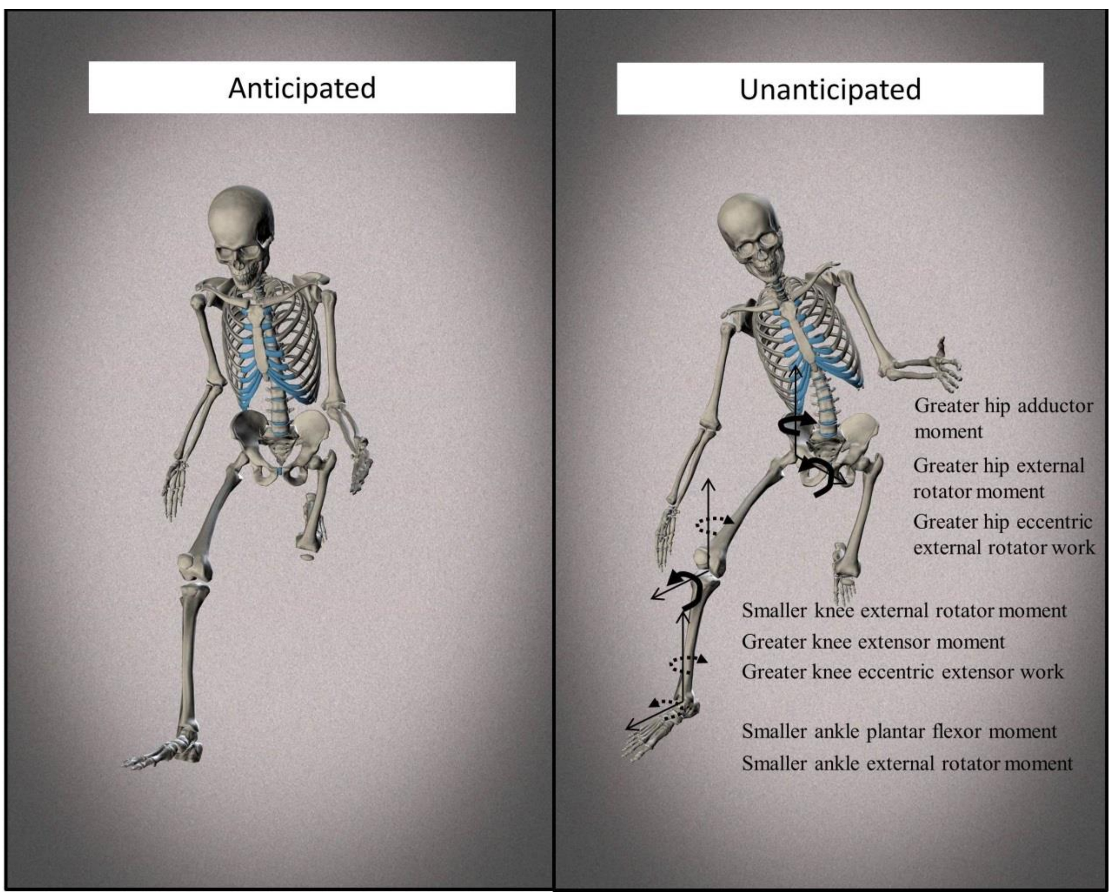

Figure 7.2 The effect of unanticipation on the biomechanics of the weight acceptance phase of a side cutting manoeuvre

7.4.2.5 Main effect for Anticipation on the Kinetics of the Side Cutting Manoeuvre

Performance of the side cutting manoeuvre in the unanticipated condition resulted in greater internal hip adductor moments (p = 0.028, η^2 = 0.005, 34-55%; p < 0.001, η^2 = 0.045, 63-100%), greater internal hip external rotator moments (p = 0.007, η^2 = 0.050, 75-100%) and greater eccentric hip work in the transverse plane (p = 0.004, η^2 = 0.124, 87-100%) compared with the anticipated condition. Likewise it led to smaller internal flexor and greater extensor knee moments (p = 0.007, η^2 = 0.060, 20-48%; p = 0.002, η^2 = 0.032, 63-100%) and smaller internal knee external rotator moments (p = 0.045, η^2 = 0.040, 33-38%), with greater negative knee work in the sagittal plane (p = 0.048, η^2 = 0.027, 70-72%). Finally, unanticipation resulted in smaller internal ankle plantarflexor (p < 0.001, η^2 = 0.075, 10-41%; p = 0.036, η^2 = 0.036, 92-100%) and external rotator (p = 0.011, η^2 = 0.020, 77-100%) moments (Table 7.4 and Figure 7.2).

7.5 Discussion

The results of our study largely supported our hypotheses that the HIIP and unanticipated condition would have significant effects on the biomechanics of the side cutting manoeuvre, without any additional effects present in the unanticipated side cutting manoeuvre post-HIIP.

Unanticipation resulted in alterations to trunk and pelvic kinematics with strong to small effects which were accompanied by increased frontal and transverse plane hip loading and increased sagittal plane knee loading which may lead to an increase in ACL loading towards the end of the WA phase. The HIIP led to altered trunk and pelvic kinematics and changes in sagittal plane knee loading that decreases ACL loading, with small to trivial effects. Also, only one trivial interaction effect was noted demonstrating that the combined effects of the unanticipated and post-HIIP conditions resulted in increased eccentric hip extensor moment. In order to discuss and interpret these results, it will be necessary to first describe the biomechanics of the side cutting manoeuvre in general (i.e. in the anticipated condition, pre –HIIP) in terms of ACL loading and subsequently the effects of unanticipation and the HIIP thereon. The discussion will also generally discuss the relationship between the kinetic chain and potential ACL loading, and the specific effects of unanticipation and the HIIP thereon.

7.5.1 The Effect of Unanticipation and the HIIP on Knee joint Biomechanics

Unanticipation resulted in increased knee loading in the sagittal planes, decreased knee loading in the transverse plane with no changes in the frontal plane. In the sagittal plane, it led to a reduced, stabilizing (internal) knee flexor moment and an increased (internal) knee extensor moment (20-48% of phase, $\eta^2 = 0.032$; 63-100% of phase $\eta^2 = 0.060$) which will, as demonstrated by Weinhandl et al. (Weinhandl, Earl-Boehm et al. 2013), increase ACL strain during unanticipated side cutting. However, sagittal plane loading alone cannot cause ACL rupture (McLean, Huang et al. 2004). Although, unanticipation did not lead to any alterations to frontal plane loading and it resulted in smaller (internal) knee external rotator moments (33-38% of the phase, $\eta^2 = 0.40$), there was no reduction in transverse plane loading during the latter part of the WA phase when greater extensor loading (63-100%) occurred, indicating that there is a greater potential for increased ACL strain from 65% of the WA. Conversely, subjects demonstrated smaller knee extensor moments post-HIIP (70-98% of phase, $\eta^2 = 0.013$), supporting previous research (McLean, Samorezov 2009) . This reduction may offset any detrimental effects of the reduced hip and knee flexion post-HIIP (27-100% of phase, $\eta^2 = 0.019$; 73-100% of phase, $\eta^2 = 0.013$ respectively). In summary, these findings suggest that unanticipation increases potential for ACL strain during the latter part of the WA phase, whereas the HIIP decreases this potential.

The absence of greater knee frontal plane loading in the unanticipated condition conflicts with many other studies, whereas the effect of unanticipation on transverse (internal) knee loading is inconclusive (Brown, Palmieri-Smith et al. 2009). In the current study, the unanticipated condition resulted in smaller (internal) knee external rotator moments (33-

38% of the phase, $\eta^2 = 0.40$), demonstrating a reduction in potential ACL strain (Oh et al. 2012). Differences may be due to differences in experimental set-up. For example our study did not have continued straight line running (Besier et al. 2001, Dempsey, Lloyd et al. 2009) as a possible manoeuvre in the unanticipated condition and so subjects may have been prepared to decelerate. The fact that the altered trunk kinematics did not lead to greater frontal plane loading suggests the hip strategy adopted is essential to avoid the increases in loading considered necessary to cause ACL injury (Oh et al. 2012, Shin et al. 2011).

The successful completion of cutting is achieved through a combination of feed-forward and feedback control. Provided there is sufficient time, feed-forward control allows for retrieval and implementation of the neuromuscular patterns from an internal programme based upon prior experience of the task (Kandel 1999). Feedback control, in contrast, allows for the modification and refinement of a task. The interaction between these two methods of neuromuscular control is important during unanticipated tasks (Seidler, Noll et al. 2004). However, this interaction requires a certain amount of time due to the delay in feedback loops. Therefore, the ability of participants to implement neuromuscular programmes to maintain optimal kinematic and kinetic control during the unanticipated side cutting manoeuvres may be reduced. This may, at least partially, account for the observed changes in the unanticipated condition. Altered neuromuscular control patterns, in particular the timing of quadriceps and hamstring EMG activity, have been proposed to be risk factors for ACL injuries as it can result in suboptimal stabilisation of the knee joint (Myer, Ford et al. 2009, Zebis, Andersen et al. 2009). It has also been demonstrated that performance of side cutting in the unanticipated condition results in increased gluteal EMG activity in the pre-landing phase (Meinerz, Malloy et al. 2015). In line with the findings of the current study, this suggests the importance of hip joint control in response to altered trunk, and centre of mass position. This may play an important role in avoiding excessive loading of the ACL during unanticipated side cutting. This may be critical given that reduced trunk control (Zazulak, Hewett et al. 2007) and hip strength (Khayambashi et al. 2015) predict ACL injury. Although static hip and lumbopelvic strengthening programmes have not found to affect the biomechanics of side cutting (Jamison, McNeilan et al. 2012), the findings of the current study suggest the necessity of incorporating exercises in unanticipated conditions with a view to refining injury prevention programmes.

7.5.2 The Effect of Unanticipation and the HIIP on Trunk Kinematics

In the current study, subjects generally landed in trunk flexion, side flexion away from, and rotation towards, the direction of cut, all of which increased throughout the phase, as previously described (Jamison, Pan et al. 2012, Patla et al. 1999). Increasing trunk flexion leads to greater (internal) hip extensor moments (Frank, Bell et al. 2013, Shimokochi et al. 2013) acting to decelerate the centre of mass in the sagittal plane. Trunk flexion also explains the increasing (internal) knee external rotator moment and (internal) eccentric knee extensor moment, which contributes to centre of mass deceleration and, controlled knee flexion (Frank et al. 2013). The eccentric (internal) ankle plantarflexor moment may also assist deceleration from 20% of the phase. The altered sagittal plane moments in the unanticipated condition and post-HIIP can be explained by altered trunk kinematics. The smaller, stabilizing (internal) knee flexor moment and greater (internal) knee extensor moments observed in the unanticipated condition can result from decreased trunk flexion and anterior pelvic tilt throughout the entire WA phase ($\eta^2 = 0.051$ and 0.072) (Shimokochi et al. 2013) and possibly from the smaller ankle plantarflexor moments (10-41% of phase, $\eta^2 = 0.075$; 92-100% of phase, $\eta^2 = 0.036$). The smaller knee extensor moments post-HIIP, on the other hand, may be due to the increase in trunk flexion (1-100% of phase, $\eta^2 = 0.038$) (Shimokochi et al. 2013). As previously observed (Mornieux, Gehring et al. 2014, Jamison, Pan et al. 2012, Dempsey et al. 2009), participants generally performed the WA phase of the side cutting manoeuvre with increasing side flexion away from the direction of travel. This was due to increasing trunk on pelvic side flexion throughout the phase, resulting in a lateral deviation of the centre of mass. Consequently, an internal adductor moment must be generated in the stance limb to maintain equilibrium. A side flexed trunk is related to increased (internal) knee varus moment during side cutting (Jamison et al. 2012). However, the initial (internal) knee varus moment preceded maximum trunk side flexion, as previously reported (Jamison et al. 2012). Counterintuitively, as trunk side flexion increased, the knee varus moment actually decreased and became a larger (internal) valgus moment. This was primarily due to the increased loading of the hip as there were no increases observed at the knee or ankle. The largely isometric (internal) hip adductor moment acted to stabilize the pelvis for the first 70% of the phase, facilitating the action of the lumbopelvic musculature to control the centre of mass (Mornieux, Gehring et al. 2014). Following this, an eccentric (internal) abductor moment controlled pelvis side flexion in the direction of cut which may represent the beginning of changing of direction.

Throughout the entire WA phase, unanticipation resulted in greater trunk side flexion and pelvic lateral flexion away from the direction of cut with strong effect sizes (η^2=0.142, 0.143). This posture was controlled by a greater (internal) hip adductor moment (34-55% of phase, η^2 = 0.28), as reported previously (Kim, Lee et al. 2014) whereas increased trunk side flexion away from the direction of cut post-HIIP (1-88% of phase, η^2 = 0.017) were not sufficient to cause any frontal plane changes at the hip, knee or ankle. The stabilizing adductor moment continues for longer in the unanticipated condition while the hip is in a position of greater abduction (1-91%. η^2 = 0.050) (Kim et al. 2014, Lee, Lloyd et al. 2013, T. N. Brown et al. 2009). The fact that the altered trunk kinematics did not lead to greater frontal plane loading suggests the hip strategy adopted is essential to avoid the increases in loading considered necessary to cause ACL injury (Oh et al. 2012, Shin et al. 2011) during unanticipated side cutting activities. This highlights the importance of the hip control during lateral deviation of the trunk and may partially explain the ACL injury predictive value of reduced trunk control (Zazulak et al. 2007) and hip strength (Khayambashi et al. 2015). It also indicates the importance of developing hip control and strength during side cutting activities with changes in trunk position in order to prevent or rehabilitate ACL injuries.

Generally, subjects performed the side cutting manoeuvre with slight trunk rotation in the direction of cut at initial contact, which increased throughout the phase which has been found to decrease ACL strain (Dempsey, Lloyd et al. 2007). Therefore, the increasingly internal rotated position of the knee, similar to previously reported (Borotikar et al. 2008, Kim et al. 2014), and (internal) knee external rotator moments from 30% of the phase onwards increase ACL strain (Oh et al. 2012, Yasuda, Ichiyama et al. 2008). This may be due to the increasing trunk flexion and (internal) hip internal rotator moment (Frank et al. 2013). During the unanticipated condition, subjects demonstrated smaller (internal) knee external rotator moments (33-38% of the phase, $\eta^2 = 0.40$), suggesting a reduction in potential ACL strain (Oh et al. 2012). The reason for this decrease is not clear. Although there is greater trunk rotation towards the direction of cut (77-100% of phase, $\eta^2 = 0.027$) (Dempsey et al. 2007) and greater (internal) external rotator moment at the hip (75-100% of the phase, $\eta^2 = 0.050$) (Frank et al. 2013) which are associated with smaller external rotator moments, they occur later in the WA phase and have small effect sizes. It may be that the reduction in trunk flexion observed in the current study is related to a decrease in transverse plane loading of the knee (Frank et al. 2013). There were no effects of the HIIP on transverse plane biomechanics apart from a greater (internal) ankle external rotator moment (1-6% of the phase, $\eta^2 = 0.050$) the reason for which is uncertain. In summary, the greater hip internal rotation angles (82-100% of the phase, $\eta^2 = 0.009$) and greater eccentric (internal) hip external rotator moments may indicate a greater challenge to control the rotation of the trunk and pelvis during unanticipated side cutting manoeuvres, again highlighting the importance of hip control during this activity.

7.5.3 The combined effects of Unanticipation and the HIIP on the Biomechanics of a Side Cutting Manoeuvre

There was only one interaction effect observed in the current study with an increase in eccentric hip extensor moment in the unanticipated condition post-HIIP. Although this only had a trivial effect size, it again demonstrates the importance of adequate hip joint control during unanticipated side cutting manoeuvres when fatigued, supporting recent research (Collins, Almonroeder et al. 2016, Khalid, Ian Harris et al. 2015). Our findings conflict with previous findings (Borotikar, Newcomer et al. 2008, McLean, Samorezov 2009) for a number of reasons. Firstly, we examined males. Secondly, we used a whole body, high intensity intermittent programme to mimic the demands of field sports instead of a squatting and jumping protocol (Borotikar, Newcomer et al. 2008, McLean, Samorezov 2009). Nevertheless, the results of our study highlight the importance of adequate hip control in performance of the side cutting manoeuvre with the combined effects of fatigue and unanticipation.

7.5.4 Limitations of the Current Study

There are a number of limitations in the current study. Firstly, comparison of our results with previous studies is limited as we analysed the *pattern* of kinetics and kinematics rather than discrete points. However, we used previous findings from studies using DPA in conjunction with cadaveric and modelling studies as a basis for the interpretation of our results. Secondly, a jump forwards and side cut manoeuvre rather than a run and side cut manoeuvre may not be considered to be game specific. However, it is similar to technique previously reported (Borotikar, Newcomer et al. 2008, McLean, Samorezov 2009) and normalises the effort of the side cutting manoeuvre. In addition, the HIIP used may not replicate the cumulative fatigue that is experienced during field sports. Finally, the unanticipated condition may not be ecologically valid as responding to a light is clearly not game specific. However, the experimental procedure limited spatial anticipation (the ability to predict what will happen) to three possibilities and temporal anticipation (predicting when the event will happen) was standardized for each subject. Also, the stimulus and response are highly compatible (i.e. light flashing to the left indicates the subject must run to the left). Therefore the unanticipated condition used in the current study may help to reduce the previously identified influence of experience on the side cutting manoeuvre (Lee et al. 2013).

7.6 Conclusion

Unanticipation and the HIIP altered trunk kinematics with subsequent lower limb biomechanical modifications. A hip and knee strategy controlled potentially detrimental trunk and pelvic kinematics in the unanticipated condition with a potential increase in ACL loading and subsequent injury risk. In contrast the HIIP led to a reduction in potential ACL loading and subsequent injury risk. When combined, that is performance of unanticipated side cuts post-HIIP, there was no additional effect on ACL injury risk. However, given that unanticipated actions will occur prior to fatigue, these results demonstrate the importance of the hip and knee in controlling potentially dangerous trunk and pelvic kinematics during the side cut in the unanticipated condition in particular and the importance of developing hip strength and control for prevention and/or rehabilitation of ACL injuries in Gaelic footballers (Khayambashi et al. 2015). It also suggests the importance of incorporation of unanticipated side cutting drills with an emphasis on good hip and knee alignment in to practice sessions (Dempsey et al. 2009). Future research should determine the impact of hip and trunk control exercises in the unanticipated condition, on the biomechanics of the side cutting manoeuvre.

7.7 Link to Chapter 8

Study 5 investigated the effect of both anticipation and a HIIP and on the biomechanics of side cutting. Similar to the findings from study 4, study 5 did not find any combined effect of anticipation and a HIIP on the biomechanics of side cutting. In addition it was found that HIIP-induced fatigue alone did not elevate ACL injury risk. However, study 5 demonstrated that performance of side cutting in the unanticipated condition resulted in greater knee loading which may increase the risk of ACL injury. The unanticipated condition resulted in altered trunk and pelvic kinematics which may have contributed to the greater hip and knee joint loading observed.

Study 5 also found that the hip joint plays a greater role during unanticipated side cutting compared with anticipated side cutting. Therefore, exercises aimed at improving dynamic trunk and hip control, particularly when combined with unanticipated conditions may improve the biomechanics of side cutting in terms of ACL injury risk. Study 6, (chapter 8) will investigate the effect of a 6 week dynamic core stability programme on the biomechanics of side and crossover cutting in anticipated and unanticipated conditions.

Chapter 8 Effects of a Dynamic Core Stability Programme on the Biomechanics of Cutting Manoeuvres: A Randomised Controlled Trial

Study 6

Study 6: “Effects of a Dynamic Core Stability Programme on the Biomechanics of Cutting Manoeuvres: A Randomised Controlled Trial.”

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STATEMENT OF CONTRIBUTION: Kieran Moran was the research supervisor for this study. Siobhan O’Connor and Chris Richter assisted in the data collection and analysis.

Author’s note: This study examined the effect of a dynamic core stability programme on the biomechanics of cutting manoeuvres. Although not included in the final publication, the effect of the dynamic core stability programme on trunk endurance tests was also investigated. The details relating to the methods, results and discussion of this is included in appendix 1.

8.1 Abstract

Deficits in trunk control predict ACL injuries which frequently occur during high risk activities such as cutting. However, no existing trunk control/core stability programme has been found to positively affect trunk kinematics during cutting activities. This study investigated the effectiveness of a 6-week dynamic core stability programme (DCS) on the biomechanics of anticipated and unanticipated side and crossover cutting manoeuvres. Thirty-one male, varsity footballers participated in this randomised controlled trial. Three-dimensional trunk and lower limb biomechanics were captured in a motion analysis laboratory during the weight acceptance phase of anticipated and unanticipated side and crossover cutting manoeuvres at baseline and 6-week follow-up. The DCS group performed a DCS programme three times weekly for 6 weeks in a university rehabilitation room. Both the DCS and control groups concurrently completed their regular practice and match play. Statistical parametric mapping and repeated measures analysis of variance were used to

determine any group (DCS *vs* control) by time (pre *vs* post) interactions. The DCS resulted in greater internal hip extensor (p = 0.017, η^2 = 0.079), smaller internal knee valgus (p = 0.026, η^2 = 0.076) and smaller internal knee external rotator moments (p = 0.041, η^2 = 0.066) during anticipated side cutting compared with the control group. It also led to reduced posterior ground reaction forces for all cutting activities (p = 0.015 - 0.030, η^2 = 0.074 - 0.105). A 6-week DCS programme did not affect trunk kinematics but it did reduce a small number of biomechanical risk factors for ACL injury, predominantly during anticipated side cutting. A DCS programme could play a role in multimodal ACL injury prevention programmes.

8.2 Introduction

Anterior cruciate ligament (ACL) injuries lead to profound short and long term consequences, with only 55% returning to sport after ACL injury (Ardern et al., 2014) and an increased incidence of osteoarthritis by middle age (Oiestad et al., 2010). Therefore injury prevention programmes (IPPs) are of primary importance. While ACL IPPs successfully reduce injury rates in specific studies (Gilchrist et al., 2008; Hewett et al., 1999; Walden et al., 2012), there has been no apparent reduction in ACL injuries in athletes participating in at-risk sports in the last decade (Ardern et al., 2014). A meta-analysis by Sugimoto et al., (2015) (Sugimoto et al., 2015) identified that core stability exercises increased the efficacy of ACL IPPs. Additionally, it was suggested that elements of successful ACL IPPs, such as a core stability programme, may be sufficient to reduce ACL injuries (Norcross et al., 2016). However, the effect of core stability programmes on biomechanical risk factors for noncontact ACL injury is poorly understood (Norcross et al., 2016; Pappas et al., 2015). Only one study has examined this during cutting activities (Jamison, McNeilan et al., 2012), when ACL injuries are likely to occur (Krosshaug et al., 2007; Walden et al., 2015). The study by Jamison et al. (Jamison et al., 2012), using a static core stability programme, did not find any effect on trunk kinematics or knee loading during unanticipated side cutting proposedly due to the exclusion of exercises targeting the dynamic control of the centre of mass (Jamison et al., 2012). No randomised controlled trial has investigated the effect of a dynamic core stability programme with perturbations on the biomechanics of unanticipated and anticipated side and crossover cutting manoeuvres.

Biomechanical risk factors for noncontact ACL injury during high risk activities such as cutting include reduced trunk control (Hewett et al., 2009; Hewett et al., 2010; Zazulak et al., 2007), less hip and knee flexion angles (Leppanen et al., 2017), greater internal[4] knee adductor moments and valgus angles (Hewett et al., 2005), and larger vertical and posterior ground reaction forces (GRFs) (Leppanen et al., 2017; Sell et al., 2007). Also, cadaveric studies demonstrate that ACL strain is increased by combined internal knee extensor (Markolf et al., 1995), adductor and especially, external rotator moments (Oh et al., 2012). Positive associations between trunk kinematics and these potentially detrimental kinetics have been identified. Specifically, trunk flexion with internal knee extensor moment (Shimokochi et al., 2013), trunk side flexion with internal knee adductor moment (Dempsey et al., 2007; Jamison, Pan et al., 2012; Mornieux et al., 2014) and trunk rotation (Dempsey et al., 2007) with internal knee external rotator moment. These associations, and the fact that reduced trunk control predict noncontact ACL injuries (Zazulak et al., 2007), form a logical basis for the inclusion of trunk control exercises in ACL IPPs (Gilchrist et al., 2008; Hewett et al., 1999; Walden et al., 2012). However, there is a dearth of studies investigating the effects of such programmes on trunk control during cutting (Pappas et al., 2015).

The absence of an effect of the core stability programme employed by Jamison et al. (Jamison et al., 2012) may also be contributed to by the method of analysis employed, i.e. discrete point analysis (DPA). DPA analyses feature reduced kinematic and kinetic waveforms (e.g. peak moment) which over simplify the original datasets (Pataky et al., 2013), leading to the analysis of less than 5% of the data (Richter et al., 2014). In addition,

[4] An internal joint moment is a measure of the body's resistance (for example muscular contractions) to external moments generated by the ground reaction force and inertial forces. For example, during the landing phase from a jump, an external knee flexor moment is resisted by an internal extensor moment generated by the quadriceps. All joint moments are reported as "internal" moments in the current manuscript. When discussing findings from previous studies which originally reported results as "external" moments, the authors have converted these results to "internal" moments to ensure consistency throughout the manuscript.

and most importantly, analysing discrete points (e.g. peak knee varus moment) can result in the comparison of points during different functional tasks (Richter et al., 2014). Statistical parametric mapping (SPM), on the other hand, is more effective at identifying biomechanical characteristics of complex movements (Richter et al., 2014). It analyses the original variables rather than predetermined discrete points (e.g. peak knee flexion angle or peak knee varus moment) and takes into account the influence of time and the influence that different variables have on each other (Pataky et al., 2013). No randomised controlled trial has investigated the effect of a core stability programme on the biomechanics of unanticipated and anticipated side and crossover cutting manoeuvres using SPM.

This study aims to investigate the effect of a 6-week DCS programme on the pattern of trunk and lower limb biomechanics during anticipated and unanticipated side cutting and crossover cutting manoeuvres. It is hypothesised that the intervention will (1) improve trunk kinematics associated with biomechanical risk factors for ACL injury, (2) subsequently ameliorate biomechanical risk factors for ACL injuries and (3) that a greater number of effects will occur in the unanticipated than the anticipated condition.

8.3 Materials and Methods

8.3.1 Study Design and Participants

We performed a randomised controlled trial on a University's Gaelic football academy. All male, collegiate footballers were assessed for eligibility. The inclusion criteria were that participants were over 18 years of age, currently injury-free and participating in varsity level Gaelic football on at least three occasions per week. The exclusion criterion was a history of lower limb ligamentous reconstructive surgery. In order to remain as an active participant in the study, participants were not allowed to miss more than two consecutive sessions and three in total. A concealed randomisation method (more details below) was used to assign study participants to the DCS or control groups. All participants provided written informed consent, and the study was approved by the local Research Ethics

Committee. Participants were required to attend a familiarisation session and two data collection sessions.

8.3.2 Intervention Procedures

Core stability exercises from successful ACL IPPs formed the basis of the DCS. The programme was developed by the lead author (EW), a certified Athletic and Rehabilitation Therapist (ARTC) and physiotherapist with over 15 years sports medicine experience. It is described according to the Template for Intervention Description and Replication (TIDier) checklist and guide (Hoffmann et al., 2014). It progressively challenged the participants' core stability defined as the ability to control trunk excursions while maintaining a functional position of the pelvis (Weltin et al., 2016) by modifying the predominantly static, original exercises to include controlled movements and perturbations.

8.3.2.1 Dynamic core stability programme details

The DCS consisted of transversus abdominis activation, bridge (Walden et al., 2012), side and prone plank (Steffen et al., 2008), lunges (Gilchrist et al., 2008) and trunk curl exercises (Hewett et al., 1999) (Table 8.1). The largely static, trunk control exercises were modified in a number of ways. Phase 1 exercises incorporated controlled movement of the trunk, hips and arms in the sagittal, frontal and transverse planes, where appropriate. Phase 2 required the participant to perform the exercises on unstable surfaces (wobble board or swiss ball) while phase 3 incorporated perturbations to the unstable surface. The DCS was progressed from one phase to the next after two weeks. The instructor (EW) gave individualised feedback to ensure maintenance and recovery of good postural alignment.

8.3.2.2 Implementation of Intervention

Participants randomised to the DCS group were required to attend 3 sessions per week for 6 weeks in addition to their normal team activities. If a participant was unable to attend the group session, an alternative session was arranged within 48 hours of the missed session.

The exercise programme was delivered and supervised by the primary author. Participants completed on average 17.4 ± 0.5 intervention sessions over the 6 weeks. Each session consisted of 8-10 exercises repeated 20 times with three sets each. Each session lasted for between 10-14 minutes (Table 8.1).

Table 8.1 Overview of dynamic core stability programme

Exercise	Progressions	Phase 1	Phase 2 (session numbers)	Phase 3
TA activation	TA activation	1,2		
	TA Activation with arm movements	3-4		
	TA activation with arm and leg movements	5-6		
Trunk curl	Trunk curls	2, 4, 6	8, 10, 12	14, 16, 18
	Trunk curls with rotations	2, 4, 6	8, 10, 12	14, 16, 18
Dynamic Bridge	Dynamic bridge	2, 4, 6	8, 10, 12	14, 16, 18
	Dynamic bridge with knee extensions	2, 4, 6	8, 10, 12	14, 16, 18
Dynamic prone plank	Shortened dynamic prone plank	2, 4	8, 10	14, 16
	Full dynamic prone plank	2, 4, 6	8, 10, 12	14, 16, 18
	Plank walk out	4, 6	8, 10, 12	14, 16, 18
Dynamic side plank	Shortened dynamic side plank	1, 3	7,9	13, 15, 17
	Full dynamic side plank	3, 5	7, 9, 11	13, 15, 17
	Full dynamic plank with trunk rotations	3, 5	7, 9, 11	13, 15, 17
Lunges	Forwards lunges with handheld weights	1, 3, 5	7, 9, 11	13, 15, 17
	Backwards lunges with handheld weights	3, 5	7, 9, 11	13, 15, 17
	Sideway lunges with handheld weights	3, 5	7, 9, 11	13, 15, 17

TA – Transversus abdominis

The control group completed their regular team practice and match play over the 6 weeks. To avoid any possible exercise contamination, the DCS group completed the intervention outside of scheduled team practice. Attendance at team practices and matches was verified by the team management. The DCS and control groups attended on average 22.3 ± 2.1 and 21.9 ± 1.8 team practices and matches, respectively.

8.3.3 Outcomes

The primary outcomes of this study were sagittal, frontal and transverse plane: (a) kinematics of the trunk, hip, knee and ankle, (b) Internal moments of the hip, knee and ankle, and (c) GRFs. These were measured during anticipated and unanticipated side and crossover cutting manoeuvres, both pre and post an intervention (DCS versus control). All testing took place in a motion analysis laboratory.

8.3.4 Test Protocol

Familiarisation sessions were conducted between 21 and 14 days prior to the start of the intervention. The pre-intervention sessions were completed 4 to 10 days prior to the start of the intervention and post testing was completed within a maximum of 10 days following intervention.

8.3.4.1 Experimental procedure

During the familiarisation session participant measurements (height, mass, leg dominance, leg length, knee and ankle widths, and maximum horizontal jump distance (157.2 ± 13.6cm) were recorded. Crossover and side cutting manoeuvres, and stop jumps were explained and demonstrated to the participants with only data for the side and crossover cutting manoeuvres analysed in this study. Participants were given a minimum of ten minutes to practise the tasks until they were comfortable with, and accurate in (as determined by the lead author), their execution.

Participants completed a standardized warm-up consisting of a 10 minute light jog and 5 minutes of lower limb dynamic stretching prior to data collection session. Participants performed a horizontal jump equalling 70% of their maximum jump distance before landing on their dominant leg and performing the side cutting or crossover cutting manoeuvres at 45^0 to their dominant side or a stop jump and continued for 2 metres through a set of speed gates at maximum effort (Figure 0.1). Light gates were positioned one metre from the ground with light gate 1 and the force platform positioned at 20% and 70% of the maximum jump distance from the starting point, respectively. When the light beam of light gate 1 was broken, the task to be performed was indicated by random activation of light gate 2, 3 or 4. The order of the light gate activation was randomized using a random number generator (www.random.org). A custom made circuit (JTEC Ltd, Dundalk, Ireland) triggered VICON motion data capture with a consistent delay of 0.012 seconds as gate 1 was broken. In the unanticipated condition, participants broke the light beam from gate 1 as they jumped through it. For the anticipated condition, gate 1 was manually broken a minimum of three seconds before execution of the task, allowing the participant sufficient time to plan their task. Tasks altered between anticipated and unanticipated conditions.

8.3.4.2 Data collection and processing

A 12 camera Vicon Motion analysis system (Oxford metrics Ltd., Oxford, Great Britain) and the Vicon plug-in-gait marker set recorded three dimensional trunk and lower extremity movements during the entire stance phase of the cutting manoeuvre (from the instant the force platform reading exceeded 10N until it dropped below 10N). The Vicon plug-in-gait marker set consisted of 16 lower limb markers, 4 trunk markers and an additional tracking marker midway between the anterior and posterior superior iliac spines bilaterally. GRF data was recorded using an AMTI force platform at 2000Hz (BP-600900; Advanced Mechanical Technology Inc., Watertown, Massachusetts, USA.). Force and marker trajectory data were filtered using a zero lag, fourth order, Butterworth technique (15 Hz cut off frequency) (Kristianslund et al., 2012). Nexus VICON software (version 1.8.5; Vicon, Oxford, Great Britain) synchronised motion and force data at 250Hz and used inverse dynamics to generate lower limb kinetic data. Internal hip, knee and ankle joint moments were calculated with inverse dynamics (Winter, 2009) and projected onto the joint axes according to the anatomical coordinate system of the distal segment.

8.3.5 Randomisation and blinding

The lead author (EW) used a computer (www.sealedenvelope.com) generated four block randomisation pattern to allocate participants to the control or DCS groups. It was not possible to blind the participants during the study or the lead author (EW) who delivered the intervention. However, the other authors who assisted in the data collection (SOC, KM) and completed the statistical analysis (CR) were blinded to the group allocation.

8.3.6 Sample Size

A power analysis was performed for sample size estimation based upon trunk and pelvic kinematic data reported by Weltin et al., 2016. (Weltin et al., 2016). It revealed a minimum requirement of 12 participants to achieve a 95% statistical power with an alpha level of greater than 0.05.

8.3.7 Statistical Analysis

Prior to data analysis, data curves were normalised by landmark registration to the average occurrences of negative peak power (19% of stance) the beginning of the concentric phase (63% of stance and posterior peak power (82% of stance) (Franklyn-Miller et al., 2017). Statistical parametric mapping (SPM) (Pataky et al., 2013) analysed each point of the variable and subjected it to a two-way repeated measures ANOVA to identify any group (DCS *vs* control) by time (pre *vs* post) interaction effects. SPM provides a test statistic field (F value) over a range of points along the variable, allowing the identification of significant phases. Following this, a randomized field theory correction ensured that significant findings were not due to chance (Pataky et al., 2013). A Tukey-Kramer post hoc analysis was used to correct for multiple comparisons, and an alpha level was set a *priori* at 0.05 for all analysis. Partial eta squared effect sizes (η^2) were classified as small 0.01- 0.06; medium 0.06 – 0.14; and large >0.14 (Pallant, 2010). For reporting purposes we define the stance phase as being 100%, and for a concise and meaningful discussion of results, only significant findings during the first 30% of the stance phase are reported and discussed. The first 30% of the stance phase represents the weight acceptance phase (Dempsey et al., 2007) which is the period when ACL injuries predominantly occur (Krosshaug et al., 2007; Walden et al., 2015).

Table 8.2 Participant characteristics

Demographics	DCS (n=15)	Control (n=16)
Age (years)	22.05 (1.47)	21.76 (1.59)
Height (cm)	180.71(6.29)	180.16 (5.62)
Weight (kg)	78.5 (8.34)	79.13 (10.24)

Values are group mean (SD), RCS = Dynamic Core Stability

8.4 Results

During November and December 2014, 110 male athletes from a University's Gaelic football academy were assessed for eligibility. 26 declined to participate and 39 did not respond. 9 subjects were excluded due to current injury and a further 4 reported a history of lower limb reconstructive surgery. In total, 32 participants were randomised to either the dynamic core stability (DCS) (n=16) or the control groups (n=16). Between the group allocation and baseline data collection group, 1 participant sustained an injury and was unable to participate in the study leaving a total of 15 and 16 in the DCS and control group, respectively. Participant flow is displayed in Figure 8.1 with baseline characteristics in Table 8.2.

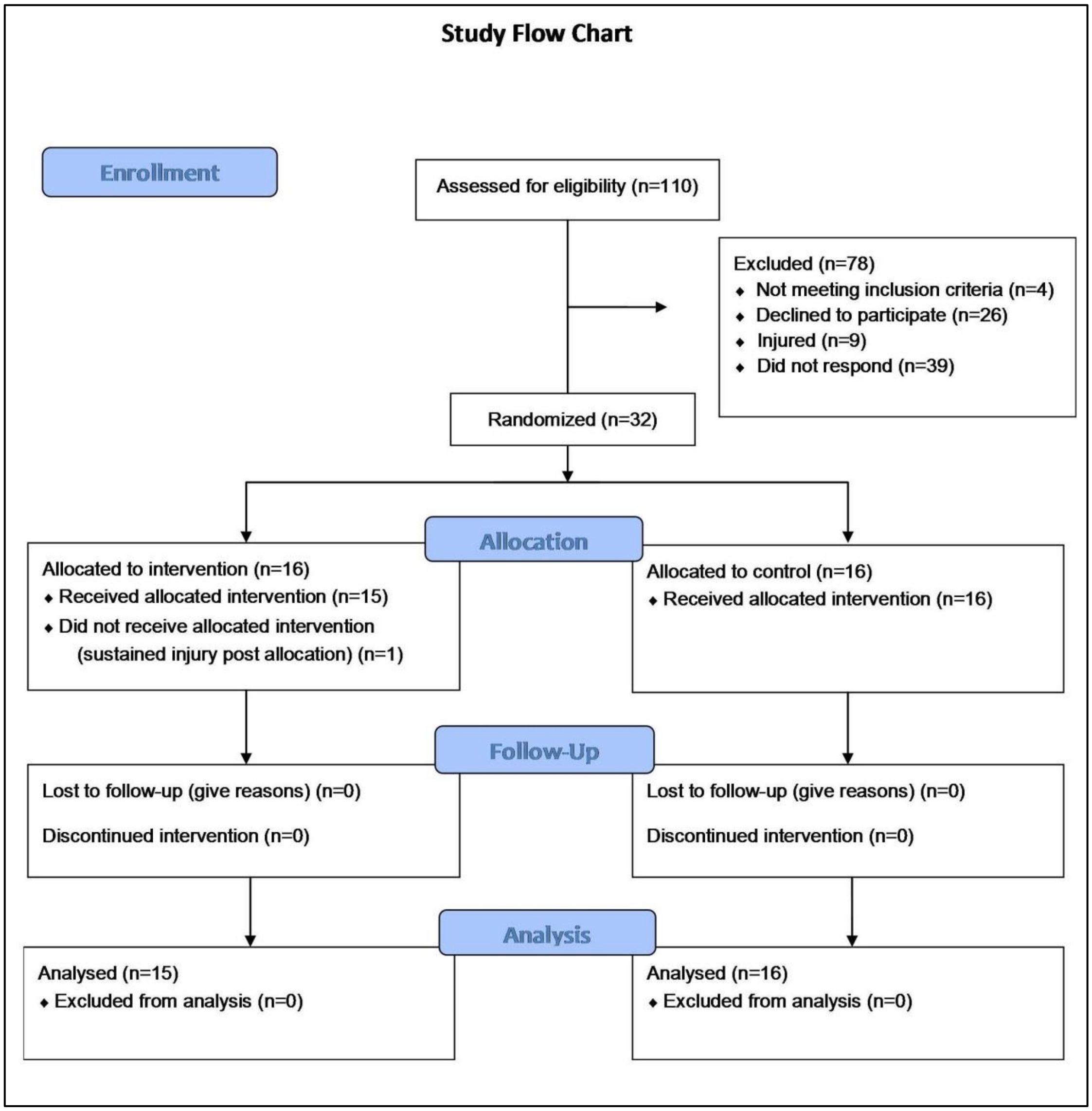

Figure 8.1 Flow of Participants through the Intervention (DCS, Dynamic core stability)

8.4.1 Primary Outcomes and Analysis

8.4.1.1 *Between group analysis of the side cutting manoeuvre*

A number of group (DCS *vs* control) by time (pre *vs* post) interaction effects were observed for side cutting manoeuvres (see Table 8.3 and Figure 8.2). For the anticipated condition, greater internal hip extensor moments (p = 0.017, η^2 = 0.079, 24-28% of stance phase), smaller internal knee valgus (p = 0.026, η^2 = 0.076, 18-25% of stance phase) and external rotator (p = 0.041, η^2 = 0.066, 15-20% of stance phase) moments were observed in the DCS group post-intervention. Also, a smaller posterior GRF was observed for the anticipated (p = 0.025, η^2 = 0.074, 11-30% of stance phase) and unanticipated (p = 0.030, η^2 = 0.081, 15-19% of stance phase) side cutting manoeuvres in the DCS group post-intervention. Significant interaction effects were not observed for the other biomechanical variables investigated.

8.4.1.2 Between group analysis of the crossover cutting manoeuvre

A number of group (DCS *vs* control) by time (pre *vs* post) interaction effects were also observed for crossover cutting manoeuvres (see Table 8.4 and Figure 8.3). Smaller posterior GRFs were observed during the anticipated (p = 0.029, η^2 = 0.105, 1-9% of stance; p = 0.015, η^2 = 0.103, 15-36% of stance) and unanticipated (p = 0.023, η^2 = 0.078, 14-31% of stance) conditions in the DCS group post-intervention. Finally, greater ankle dorsiflexion (p = 0.017, η^2 = 0.079, 13-19% of phase) was evident in the DCS group post-intervention. There were no significant interaction effects for the other biomechanical variables analysed.

8.4.1.3 Adverse events

No adverse events related to the DCS programme were reported during the intervention period.

Table 8.3 The effect of a dynamic core stability programme on the weight acceptance phase biomechanics of anticipated and unanticipated side cutting manoeuvres

Variable	Condition	Effect	phase	*p*	η^2
Trunk kinematics (°)	ANT and UNA	NS			
Pelvic Kinematics (°)	ANT and UNA	NS			
Hip Flexor/extensor Moment ($NmBW^{-1}$)	ANT	Greater extensor moment in DCS	24-28%	0.017	0.079
Knee valgus/varus moment ($Nm\ BW^{-1}$)	ANT	Smaller valgus moment in DCS group	18-25%	0.026	0.076
Knee rotator moment ($Nm\ BW^{-1}$)	ANT	Smaller external rotator moment in DCS	15-20%	0.041	0.066
Ankle kinetics and kinematics	ANT and UNA	NS			
Anterior-Posterior GRF ($N\ BW^{-1}$)	ANT	Smaller posterior GRF in DCS	11-30%	0.025	0.074
	UNA	Smaller posterior GRF in DCS	15-19%	0.030	0.081

ANT – Anticipated; UNA – Unanticipated; DCS –Dynamic core stability group; CON – Control group; NS – Non significant

Table 8.4 The effect of a dynamic core stability programme on the weight acceptance phase biomechanics of anticipated and unanticipated crossover cutting manoeuvres

Variable	Condition	Effect	phase	*p*	η^2
Trunk rotation (°)	UNA and ANT	NS			
Pelvic rotation (°)	UNA and ANT	NS			
Hip kinematics and kinetics	UNA and ANT	NS			
Knee kinematics and kinetics	UNA and ANT	NS			
Ankle plantarflexion/dorsiflexion angle (°)	ANT	Less plantarflexion in DCS	13-19%	0.027	0.098
Anterior-posterior GRF ($N\ BW^{-1}$)	ANT	Smaller posterior GRF in DCS	1-9%	0.029	0.105
	ANT	Smaller posterior GRF in DCS	15-30%	0.015	0.103
	UNA	Smaller posterior GRF in DCS	14-30%	0.023	0.078

ANT – Anticipated; UNA – Unanticipated; DCS – Dynamic core stability group; CON – Control group; NS – Non significant

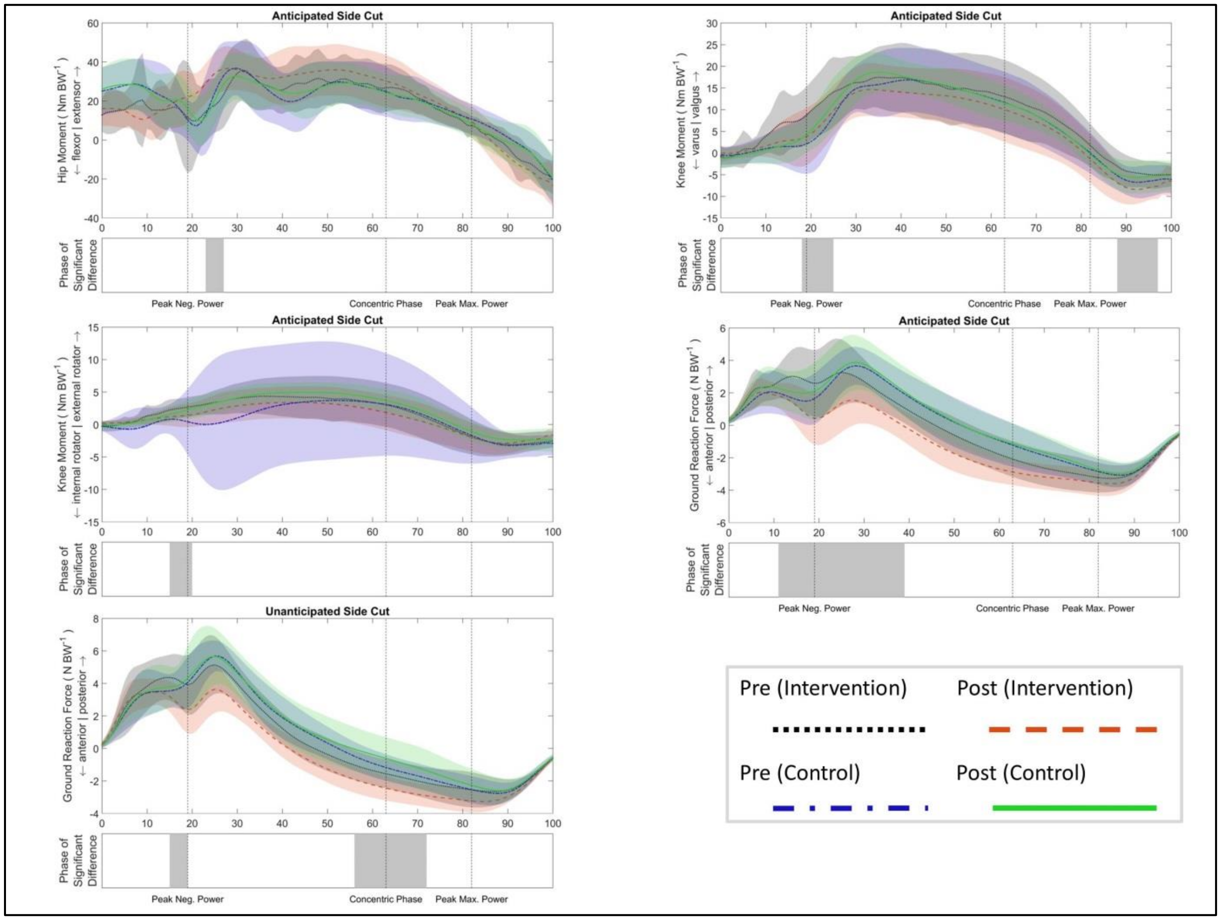

Figure 8.2 The effect of a dynamic core stability programme on the hip biomechanics of side cutting manoeuvres

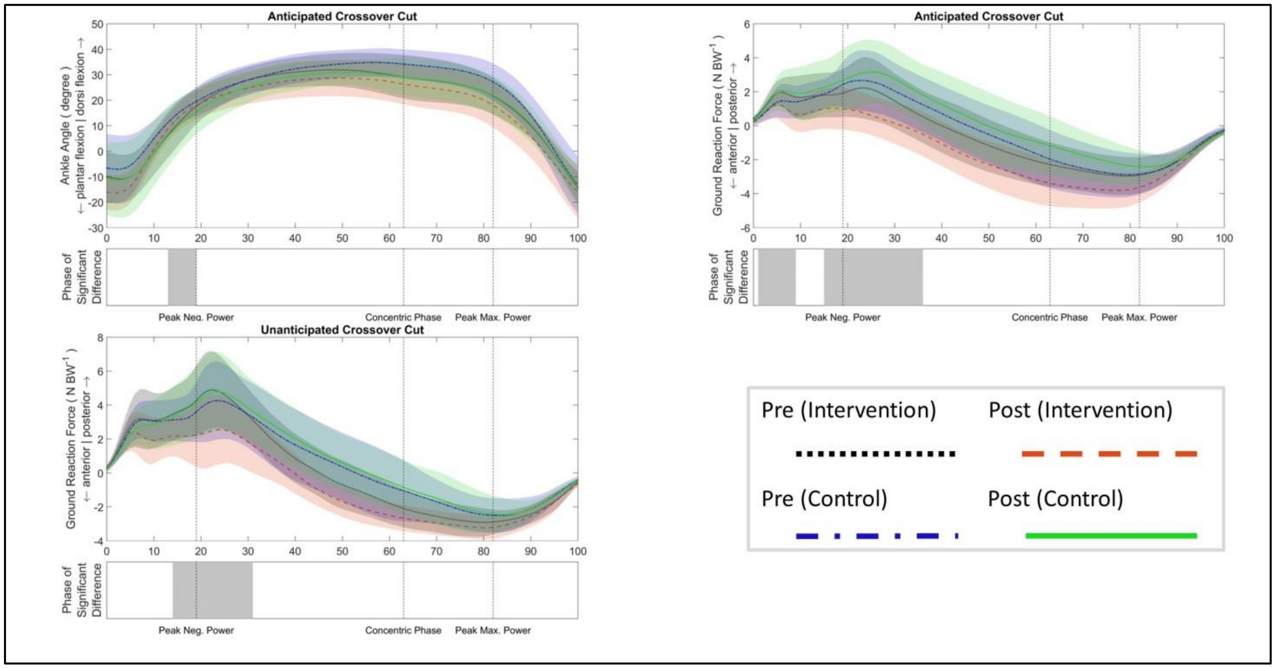

Figure 8.3 The effect of a dynamic core stability programme on the biomechanics of crossover cutting manoeuvres

8.5 Discussion

The aim of this study was to determine if a 6-week DCS intervention affected trunk kinematics and subsequent lower limb biomechanics during anticipated and unanticipated side and crossover cutting manoeuvres. We hypothesised that incorporation of dynamic and perturbative elements to trunk control/core stability exercises from successful IPPs would improve trunk kinematics associated with biomechanical risk factors for ACL injury and a subsequent amelioration of lower limb biomechanical risk factors for ACL injury. Also, due to the specificity of training, we believed that the incorporation of the perturbation element would lead to greater effects in the unanticipated condition compared with the anticipated condition. Despite the fact the secondary outcomes demonstrated that DCS group significantly improved their core endurance, the first hypothesis can be rejected as we found that the DCS intervention did not affect trunk kinematics. The second hypothesis was partially supported as the DCS resulted in a small number of positive effects on biomechanical risk factors for ACL injuries, particularly during anticipated side cutting manoeuvres. Finally, our third hypothesis can be rejected as there were a greater number of effects observed in the anticipated condition. In short, we found that in varsity level male athletes a DCS programme had a small number of effects on the biomechanics of cutting manoeuvres, which are deemed high risk activities for ACL injuries (Krosshaug et al., 2007; Walden et al., 2015). These findings may have important implications for ACL IPPs.

Similar to other interventions that targeted improved core strength (Jamison et al., 2012) or perturbation training (Weltin et al., 2016), the DCS did not lead to improved trunk kinematics despite an increase in core endurance times. Trunk kinematics during side cutting have previously been investigated (Dempsey et al., 2007; Mornieux et al., 2014) although they are less well understood during crossover cutting. A trunk roll strategy, incorporating trunk side flexion and rotation, is used to redirect the centre of mass in the new direction of travel during side cutting (Mornieux et al., 2014; Patla et al., 1999) and is greater during unanticipated side cutting manoeuvres (Lee et al., 2013). This trunk posture is commonly observed at the time of ACL injury (Hewett et al., 2009) and associated with increased internal knee varus loading (Dempsey et al., 2007; Mornieux et al., 2014), which can predict ACL injuries (Hewett et al., 2005). Therefore studies have investigated techniques to reduce side flexion away from the direction of cut (Dempsey et al., 2009; Jamison et al., 2012; Weltin et al., 2016). To date, a 6-week programme specifically targeting technique correction rather than core stability is the only successful intervention to reduce trunk side flexion away from the direction of travel and knee varus loading during anticipated and unanticipated side cutting (Dempsey et al., 2009). However, it is important to point out that although statistically significant, there was only an average reduction of 0.6° and 3.5° in trunk side flexion during unanticipated and anticipated side cutting respectively. Following the absence of a positive effect from their 6-week static core stability exercise programme, Jamison et al. (Jamison et al., 2012) proposed that the incorporation of a dynamic core stability programme with perturbations would yield improved results. There are a number of potential reasons for the absence of an effect on trunk kinematics in the current study. Firstly, as we examined elite male athletes experienced in a field sport that requires regular changes of direction, it could be that they had sufficient control prior to the intervention. Perhaps most importantly, the trunk roll technique may be necessary to perform a cutting manoeuvre, particularly when time is restricted (Mornieux et al., 2014). Functionally, it may be more important that athletes are able to control, rather than limit, trunk roll while simultaneously maintaining good alignment and stability of the lower limb. An inability to do this may cause overload of the ACL and explain why a reduction in hip strength predicts ACL injury (Khayambashi et al., 2015). Future studies should screen athletes for suboptimal limb control during cutting activities and investigate the effect of core stability programmes in athletes deemed to have suboptimal control.

While no effect on trunk kinematics was observed, a potential beneficial effect of the DCS common to all cutting activities (side and crossover cutting, anticipated and unanticipated) was a reduction in posterior GRFs (Tables 8.3 and 8.4). While there has been some debate on the topic (van den Bogert and McLean, 2006), increased posterior GRF is considered a risk factor for ACL injuries as it is associated with (Chappell et al., 2005; Yu et al., 2006), and predicts (Sell et al., 2007), proximal anterior tibial shear force which directly loads the

ACL (Markolf et al., 1995). This is supported by studies which found that ACL injuries tend to occur during decelerating activities (Krosshaug et al., 2007; Walden et al., 2015) when there is a greater posterior GRF and that ACL strain significantly increases during decelerating activities in vivo (Cerulli et al., 2003). Therefore, the findings of the current study suggest that a decrease in posterior GRF is a beneficial effect of the DCS. This may be achieved by a variety of adaptive strategies due to the large number of biomechanical degrees of freedom along the kinetic chain, thus masking any significant changes in the kinetics or kinematics of the trunk and lower limb (Richter et al., 2014).

The DCS also positively altered two biomechanical risk factors for ACL injuries at the knee during anticipated side cutting manoeuvres (Tables 8.3 and 8.4). It led to smaller internal knee external rotator and internal knee valgus moments during 15-20% and 18-25% of stance respectively, both of which would act to decrease ACL strain (Oh et al., 2012). These beneficial effects of the DCS on knee loading during anticipated side cutting may be explained in a number of ways. Firstly, the increase in hip extensor moment (Table 8.3) could increase the stability of the femur thereby reducing loading of the knee as previously reported (Shimokochi et al., 2013). The increase in internal hip extensor moments observed in the current study is not surprising given that many of the exercises included in the DCS targeted the hip musculature. Secondly, during unanticipated tasks, participants had to position their foot and lower limb in such a position as to allow the completion of any of the possible tasks (side cut, crossover cut or stop jump). On the other hand, during anticipated side cutting, participants had sufficient time to implement a pre-planned cutting strategy which would allow for altered foot positioning (Patla et al., 1999) and lower limb alignment. For example, if during anticipated side cutting, participants placed their stance foot in a less internally rotated position relative to the knee, this, combined with a decreased posterior GRF, may result in a decreased internal knee external rotator moment. Foot position relative to progression of cut was not analysed in the current study and should be included in future investigations given the influence of foot position and orientation on knee joint loading (Donnelly et al., 2016) during side cutting. Also, the intensive nature of the DCS programme may have significant implications for its implementation. It requires a significant time commitment from participants and supervision from a sports medicine professional. This may affect the levels of adoption and compliance, low levels of which have been found to negatively affect the IPPs outcomes (Joy et al., 2013; Norcross et al., 2016; Sugimoto et al., 2012). If adoption and compliance was considered to be an issue, the effects of a condensed DCS programme may warrant future investigation.

Finally, the DCS also led to a decrease in ankle plantarflexion angle during anticipated crossover cutting. Previous studies have found that reduced plantarflexion angle during side cutting is related to an increase in knee varus loading (Donnelly et al., 2016), and potential ACL strain. However, this relationship has not been explored in crossover cutting. As there were no changes at the knee joint, it is difficult to interpret the potential affects from the change in dorsiflexion angle.

There are a number of limitations to the current study. Participants' varsity team training/competition was recorded to ensure participants completed at least three sessions per week. However, control group participants may have completed additional sessions outside of the varsity team environment. Also, as the cutting task was preceded by a jump in order to standardise effort and the unanticipated condition was created using the light gate systems, ecological validity of the study may be reduced as they are not game specific. It is possible that cutting following higher velocity running in response to a sporting stimulus would be more sport realistic, however it is important to standardise approach speed to allow comparisons (Brown et al., 2014). Finally, it was not possible to blind all the researchers and participants to the group allocation. However, all researchers, with the exception of the primary author, were blinded.

8.6 Conclusion

A DCS programme did not alter trunk kinematics during anticipated or unanticipated side and crossover cutting manoeuvres despite leading to an increase in core endurance. It did, however, lead to a decrease in posterior GRF in all cutting manoeuvres, which may reduce anterior tibial shear force and ACL strain. During anticipated side cutting, it also resulted in decreased internal knee valgus and external rotator moments, which are biomechanical risk factors for ACL injuries. The results of this study demonstrate that a DCS leads to biomechanical improvements predominantly during anticipated side cutting. These findings support the practice of including core stability exercises as part of a multimodal ACL IPP, rather than a standalone IPP.

Chapter 9 Thesis Summary, Conclusions and Future Recommendations

9.1 Summary and Conclusions

The review of literature conducted in this thesis identified that there was a deficit in the understanding of the effects of high intensity, intermittent exercise on biomechanical risk factors for ACL injuries during cutting. Subsequently, the HIIP developed for this thesis, and used in the first five studies, led to similar physiological demands to those seem in soccer and importantly, it negatively affected dynamic postural control (study 1). As deficits in dynamic postural control predict injury, this suggested that athletes may be at a greater risk of injury following the HIIP. However, the HIIP did not affect biomechanical risk factors for ACL injuries during the vertical drop jump (study 2). This may have been due to the fact that fact that the vertical drop jump was not demanding enough as it is a 2 legged activity and does not consist of any decision making element. It is important to note that noncontact ACL injuries often occur during unanticipated single legged cutting and landing activities when athletes are responding to the sporting environment. The combination of performing unanticipated cutting when fatigued is proposed to place athletes at a particularly high risk of ACL injury. Given that the performance of unanticipated cutting tasks requires significant neurocognitive function and deficits in neurocognitive function are associated with ACL injury, the effect of the HIIP on neurocognitive function was examined. The HIIP was found to negatively affect aspects of neurocognitive function (study 3). This, in combination with the detrimental effects observed on dynamic postural control, suggested that athletes would be at greater risk of injury during high risk of activities post-HIIP that required substantial neurocognitive function, such as unanticipated cutting.

The combination of the HIIP and unanticipation did not increase the magnitude of biomechanical risk factors for ACL injuries during the weight acceptance phase of side or crossover cutting (studies 4 and 5). This indicates that performing unanticipated cutting activities following high intensity, intermittent exercise does not increase the risk of ACL injury above that due to the HIIP or unanticipation alone. Indeed, the HIIP demonstrated a small reduction in the magnitude of biomechanical risk factors for ACL injuries without a reduction in performance as assessed by the time to complete the cutting tasks. This suggests that athletes adopt an ACL protective strategy following the HIIP. On the other hand, unanticipation resulted in altered biomechanics during side and crossover cutting which increased potential ACL loading. The strongest effects of unanticipation were observed in sagittal and frontal plane trunk kinematics which have been demonstrated to increase potential ACL loading. This may be due to the trunk roll technique which is employed when there is limited time to plan and execute a change of direction such as unanticipated cutting. In order to redirect the centre of mass in the new direction of travel during cutting, individuals move the centre of mass to the side of the base of support by side flexing the trunk, before bringing the centre of mass anterior to the base of support and advancing it in the new direction of travel by flexing and rotating the trunk. During side and crossover cutting, the alterations in trunk kinematics were accompanied by increased loading of the hip joint in the frontal and transverse planes and increased knee joint loading in the sagittal plane. As it is necessary for athletes in field sports to perform unanticipated cutting tasks in response to the sporting environment, the ability of the trunk, hip and knee musculature to safely control the re-direction of the centre of mass is critical. Therefore, exercises to improve dynamic trunk and hip control may be beneficial in ameliorating biomechanical risk factors for ACL injuries during anticipated and unanticipated side and crossover cutting.

The dynamic core stability programme developed for this thesis aimed to improve trunk and hip strength and control during planned movements and in response to perturbations. Although it did not alter trunk kinematics during anticipated or unanticipated side and crossover cutting, it led to a reduction in a small number of risk factors for ACL injuries, particularly during anticipated side cutting (study 6). This suggests that there are some limited beneficial effects to the incorporation of a dynamic core stability programme for ACL IPPs.

9.2 Future Recommendations

Based upon the findings of this thesis, there are a number of areas that warrant further investigation.

- Firstly, this thesis focussed on male athletes due to the lack of knowledge on biomechanical risk factors for ACL injuries in males. However, the review of literature highlighted the deficiencies of research on the effects of both fatigue and anticipation on cutting and the effect of ACL IPPs on cutting biomechanics in general. The studies employed in this thesis could be repeated in females with the aim of identifying risk factors during cutting and the development and evaluation of interventions to address any risk factors identified.
- Although the fatigue protocol used in this study negatively affected dynamic postural control and neurocognitive function, it did not detrimentally affect the biomechanics of the vertical drop jump or cutting. The impact of other ecologically valid fatigue protocols should be investigated.
- The effect of different interventions on cutting should be investigated. This should include programmes aimed at improving trunk roll technique during unanticipated cutting under increasingly challenging situations.
- The advent of improved sensor technology, data storage and data mining processes can facilitate advanced research in this area. Firstly, sensor technology and data mining methods should be used to investigate the biomechanics of high risk activities during actual sporting activities rather than attempted recreations in biomechanical laboratories. In the future it may also allow for epidemiological studies to take place whereby sensors are continually worn by athletes during sporting events. This would facilitate the recording of loading during actual ACL injuries, providing invaluable information regarding risk factors for, and prevention of, ACL injuries.

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Chapter 11 Appendix 1

Additional information for Chapter 8 ""Effects of a Dynamic Core Stability Programme on the Biomechanics of Cutting Manoeuvres: A Randomised Controlled Trial."

11.1 Additional Outcomes

The secondary outcomes of the study were core muscle endurance testing as assessed by the timed side plank, prone plank and trunk flexor endurance tests.

11.2 Additional Experimental Procedures

Following a 30 minute rest period, participants' core endurance was assessed using the timed side plank, extensor endurance and trunk flexor endurance tests as previously described (Jamison et al., 2012; McGill et al., 1999). For the side plank test, participants were required to rest on their side on a firm rubber mat. They placed one foot on top of the other and rested on their hip and elbow, with their opposite arm placed in approximately 90° abduction with the elbow extended. They were then instructed to lift their hip off the mat and maintain that position for as long as possible. The duration of this was recorded and time was stopped when any part of the participant's body, apart from the ipsilateral foot and elbow, touched the ground. For the extensor endurance test, participants were positioned prone on an examination table with their anterior superior iliac spine at the edge of the table and their ankles secured to the table. The test began when the participant assumed a horizontal position with arms crossed over chest. An inclinometer was positioned at the level of the tenth thoracic vertebra. Time was stopped when the participant deviated from horizontal by more than 10°. Finally, the flexor endurance test was completed by positioning the participant on the examination plinth with the feet resting on the plinth, the knee and hips at approximately 90° of flexion. The participant was support in this position by a wedge. Timing began when the wedge was moved 10 cm away from the participant's trunk and it was stopped when the participant's back touched the wedge.

11.3 Additional Statistical Analysis

The secondary outcome of core endurance results was subjected to a two-way repeated measures ANOVA to identify any group (DCS *vs* control) by time (pre *vs* post) interaction effects.

11.4 Additional Outcomes and Analysis

A number of group (DCS *vs* control) by time (pre *vs* post) interaction effects were observed for core endurance test demonstrating a significant increase in flexor endurance, extensor endurance and side plank times in the intervention group (Table 11.1) over the course of the intervention compared with the control group.

Table 0.1 The effect of a dynamic core stability programme core endurance tests

		DCS group		**Con Group**		**Interaction effect**	
		Pre	**Post**	**Pre**	**Post**	***p***	**η^2**
Core Endurance Tests (seconds)	Flexor endurance	85.4 (35.7)	137.6 (32.9)	80.0 (27.1)	84.9 (27.3)	<0.001	0.405
	Extensor endurance	80.4 (36.5)	104.6 (27.6)	89.3 (21.8)	93.5 (19.9)	0.003	0.271
	Left side Plank	82.4 (17.8)	101.5 (23.2)	87.4 (14.6)	91.2 (14.5)	0.003	0.264
	Right Side Plank	80.2 (14.6)	102.8 (19.4)	85.3 (13.1)	88.5 (12.8)	0.001	0.422