THE EFFECTS OF AGE AND SEX ON BIOMECHANICAL ASYMMETRIES OF THE LOWER LIMBS DURING A WALK, RUN AND SIDE-CUT TASK

by

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This thesis is dedicated to:

My former right knee. RIP

Table of Contents

List of Tables

List of Figures

Abstract

INTRODUCTION: The anterior cruciate ligament (ACL) is an important contributor to knee joint stability during athletic maneuvers such as a side-cut. ACL rupture is associated with short- and long-term consequences that place a heavy burden on the health care system. 70-80% of all ACL injuries are non-contact in nature and are 2-8X more prevalent in females. The concept of limb lateralization may explain differences in dynamic control between the lower limbs and may be evidence for the unconscious preference toward greater loading of one limb versus the other, thus resulting in asymmetry of lower limbs. The purpose of this study was threefold: i) to identify if asymmetry existed beyond a clinically accepted 10% threshold for peak hip and knee joint flexion and abduction moments as well as peak hip and knee flexion angles and flexion angles at initial contact (IC) in an athletic population across age and sex, as well as during walk, run, and side-cut tasks, ii) identify the proportion of each population that experienced greater than 10% (>10%) asymmetry for each of the biomechanical variables of interest, and iii) to identify if differences in asymmetry exist across age and sex to further understand if asymmetry may function as an etiological risk factor for ACL injury. **METHODS:** Bilateral data was collected for 122 healthy high-performance cutting sport athletes. Four populations were identified based on age and sex (pre-pubescent males/females; post-pubescent males/females). Mean peak hip and knee internal joint flexion and abduction moments and mean peak hip and knee flexion angles and flexion angles at initial contact were calculated for stance phase of over-ground walking, running and side-cut tasks. Right and left limbs were reclassified as greater or lesser to prevent obscuring absolute asymmetry. Calculated asymmetry measures were subject to a 2x2 ANOVA to detect statistically significant differences among groups. The proportions of participants experiencing >10% asymmetry were calculated for each population and differences between populations was tested using a Chi-Square Test. Confidence intervals for the proportion of subjects with >10% asymmetry between limbs were estimated based on the binomial distribution. **RESULTS:** The percentage of asymmetry for peak extension and peak abduction moments as well as flexion angles at instant of contact during all tasks were greater than expected for all populations. At least 27% of the total population had >10% asymmetry across all variables and across all tasks. Age effects were noted for peak hip flexion and hip flexion angles at initial contact for all tasks, peak knee flexion angle (pKFA) and knee flexion angle at initial contact (KFA_IC) during the cut task, and peak knee extension moment (pKEM) during the walking task. In all cases, pre-pubescent athletes displayed greater asymmetry than post-pubescent athletes. Main effects of sex were noted for KFA_IC during the walk task and pKEM during the running task. In both cases, males displayed a greater asymmetry than females. No interaction effects were found. Differences in proportions of participants experiencing >10% asymmetry were found for pKFA during the walk and cutting tasks. Differences in the proportion of athletes exhibiting >10% asymmetry were found for pKFA during walk and cut tasks. **CONCLUSION:** Findings of this study may have important implications on gait evaluations, particularly in clinical and research settings where asymmetry is used as an outcome. The high proportion of the healthy population exhibiting >10% asymmetry suggests additional research is required to determine acceptable levels of lower limb kinematic and kinetic asymmetry in a healthy population as well as for return to play criteria. High variance for each variable among groups may have been a limiting factor for identifying age and sex effects.

List of Abbreviations and Symbols Used

ACL	anterior cruciate ligament
AI	asymmetry index
ANOVA	analysis of variance
BW	body weight
cm	centimeters
EMG	electromyography
FP	force platform
GA	Gait Asymmetry
GRF	ground reaction force
Hz	hertz
Kg	kilogram
LCL	lateral collateral ligament
m	meters
MCL	medial collateral ligament
mm	millimeters
msec	millisecond
MVIC	maximum voluntary (isometric) contraction
m/s	meters per second
N	Newton
N*m	Newton-meter
NSI	Normalized Symmetry Index
PCL	posterior cruciate ligament
RI	Ratio Index
SA	Symmetry Angle
yrs	years

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Chapter 1 – Introduction

1.1 Statement of the Problem

Anterior cruciate ligament (ACL) injury is one of the most common and traumatic sports-related injuries to the knee joint. ACL injury continues to receive widespread attention throughout medical and sporting communities due to the severity and the potential effects associated with the injury. Though many envision these injuries to be the result of a devastating blow to the knee or body during sport, it is less commonly known to the general public that approximately 70-80% of ACL injuries occur in the absence of contact and are usually the result of a combination of several factors when performing high-risk sporting maneuvers such as sudden deceleration, abrupt cutting to change direction and jump landing.[1–7] These maneuvers are thought to elicit increased mechanical load on the ACL and can force the lower-limb into a high risk injury position called the 'position of no return,' which is categorized by internal rotation at the hip, along with hip adduction as well as knee valgus, external tibial rotation and subtalar pronation.[8] The most concerning aspect of noncontact ACL injuries is the one-off nature of the injury, where the athlete sustains the injury during a maneuver they have performed safely countless times over their career.[9]

Numerous short-and-long-term consequences are associated with ACL injuries, including an increased risk of re-injury to both the previously injured limb and the uninjured limb[10–12] and perhaps most concerning, the development of early onset osteoarthritis (OA).[13–17] Rupture of the ACL also requires reduced participation in sport, and lengthy and rigorous rehabilitation following an invasive surgical procedure.[18–20] With participation in sport increasing annually, so too is the rate of ACL injury. It is currently estimated that 400,000 ACL injuries occur annually in the United States alone.[21,22] A population adjusted estimate of ACL reconstructions indicated that reconstruction rates, in the United States have increased by 37% between 1994 and 2006.[23] Thus the resulting annual cost of ACL reconstruction is estimated to be in excess of two billion dollars.[24,25]

Unfortunately for females, the epidemiological study by Buller et al. (2015) also indicated an increased proportion of females undergoing ACL reconstruction between 1994 and 2006.[23] This provided more evidence of a sexual dimorphic incidence of ACL rupture. An extensive number of studies have revealed that females exhibit a 2-8 times greater likelihood of sustaining a noncontact ACL injury compared to their male counterparts, particularly in sports which involve ballistic maneuvers (e.g. cutting, decelerating and jumping) such as football, volleyball, soccer and basketball.[2,26,35,27–34] Collectively, these studies show little evidence which demonstrates a sex disparity in ACL injury rates in prepubescent athletes[26,36,37], which is in harsh contrast to pubescent[30,32,34] and collegiate level or post pubescent athletes.[29,31] Research by Griffin et al. (2006) reported that the highest frequency of noncontact ACL injuries occur between the ages of 15 and 25 years of age.[38] Therefore, pubescent and post-pubescent females participating in high-level, dynamic sports seem to be at the greatest risk of sustaining a noncontact ACL rupture.

Various studies that have attempted to further the understanding of the mechanism of injury often compare specific anatomical characteristics that differ between males and females. Intercondylar notch width,[39–43] Q-angle,[44–48] excessive foot pronation[43,49–51] and joint laxity[42,43,52–54] have all been discussed as potential risk factors for ACL injury. Hormonal differences between males and females may further explain the sexual dimorphic rates of ACL injury at the onset of puberty. Several studies have investigated the menstrual cycle[55–58] and ligamentous laxity[59–62] as potential contributing factors to injury incidence, however, these studies vary widely in their methodology and thus the findings tend to be contradictory.

While the categorization of risk factors drastically varies throughout the literature, risk factors that are potentially modifiable offer the greatest opportunity for reducing the incidence of ACL injury, particularly in females. Many laboratory studies have compared the biomechanics and/or neuromuscular control strategies of the lower limb between sexes during high-risk ballistic sporting maneuvers such as single and/or double leg jump landings,[34,35,63–70] or cutting to change directions.[71–75] However, outside of epidemiological studies, the effects of age and pubertal maturation on lower limb biomechanics and/or neuromuscular control during drop-jumps has been studied less extensively.[34,35,63,67] Moreover, the effect of age on lower limb biomechanics and/or neuromuscular control during cutting maneuvers remains vastly unstudied.

It has been demonstrated that high-risk ballistic athletic maneuvers place increased load on the ACL and thus increase the potential for ACL injury.[76–81] Though these increased loads have been shown to increase the load in sagittal, frontal, or transverse planes, the greatest loading of the ACL occurs as a result of loading the knee joint simultaneously across multiple planes.[82–85] In addition, unanticipated ballistic maneuvers have been shown to alter lower limb biomechanics in a laboratory setting when replicating high risk maneuevers[86–90] as well as in video and meta analyses.[74,91] This alteration is likely to increase mechanical load of the knee joint, and potentially lead to the perplexing one-off nature of ACL injury. However, research indicates that noncontact ACL injuries during athletic maneuvers are the result of altered or deficient biomechanics not only at the knee, but also the hip, trunk, and ankle.[29,92–94]

Research by Brophy et al. (2010) suggested limb dominance, as determined by operational kicking limb (dominant) versus the support limb (non-dominant) serves as an etiological factor regarding ACL rupture whereby females are more likely to injure their non-dominant limb, and males their dominant limb. They suggested the discrepancy was a result of underlying sex-based anatomical difference or differences in neuromuscular coordination patterns during cutting maneuvers. Similarly, the concept of lateralization has been previously used to define differences between the operational and support limb. Given that its well understood that lateralization can lead to task-specific roles of the support and operational limb, it is conceivable that developed bilateral strength differences and/or differences in muscle recruitment may lead to the unconscious preference toward greater loading of one limb versus the other, thus resulting in asymmetry of lower extremity biomechanics. Notwithstanding, symmetry of lower limb biomechanics is often assumed in healthy individuals and deficiencies from perfect symmetry are believed to be the product of pathology[95]. Bilateral strength measures of symmetry are also used in a clinical setting to define goals for return-to-play rehabilitation in ACL reconstruction patients[12,20] as well as other lower-limb injuries,[96] thus it is logical to use that same clinical measure of symmetry to define asymmetry in a healthy population. Despite the aforementioned assumption of symmetry, previous research presents contradictory findings.[97] Confounding results have also been reported in investigations of joint moment asymmetry.[98–101] Only one biomechanical study has attempted to use a clinically acceptable definition of asymmetry to quantify joint moment asymmetry in healthy individuals.[95] Findings from Lathrop-Lambach et al. noted that 55-

69 percent of healthy individuals experienced greater than 10% asymmetry in external knee adduction moments during normal gait. It has yet to be determined in the literature if ballistic movements demonstrate similar biomechanical asymmetries or if these movements display similar prevalence of asymmetry across age and sex.

To date, there are no biomechanical laboratory studies in the literature that have made simultaneous age and sex comparisons of joint kinematic and kinetic (angles and moments) asymmetries between the lower limbs in an athletic population during a walking, running or cutting task. As such, it is apparent that there is a need for development of a comprehensive age and sex comparative study that simultaneously measures and compares asymmetries of lower limb biomechanics during both primarily unidirectional walking and running tasks, as well as during more ballistic multidirectional athletic maneuvers such as a side-cut task.

1.2 Rationale

This study was designed to address four main areas that were identified through careful review of the existing literature on noncontact ACL injuries. The four areas that will be discussed in the following sections include: i) age of participating athletes at risk of injury, ii) biomechanics of the lower limb, iii) asymmetries of lower limb biomechanics as defined by a clinically accepted return-to-play threshold and iv) asymmetry as a potential risk factor for, and a potential explanation of the one-off nature of ACL injury.

This study will be the first to use a clinically relevant definition of asymmetry to investigate lower limb biomechanics for a healthy athletic population during gait. In addition, this will be the first study to use that same clinically relevant definition to investigate lower limb biomechanics for an athletic population during sport-specific athletic tasks such as running, and side-cutting. Finally, this study will assess asymmetry of the lower limbs for walk, run and side-cut tasks across age and sex.

1.2.1 Age of Participating Athletes

Although females have been shown to exhibit greater ACL injury rates for both pubescent,[30,32,34] and post-pubescent athletes,[29,31] no sex disparity in ACL injuries rates has been found in pre-pubescent athletes.[26,36,37] An epidemiological study by Buller et al.

(2015) discovered a 304 percent increase in rate of female ACL reconstructions from 1994 to 2006, with the largest proportion of these reconstructions occurring in those aged 15 to 34 years of age.[23] Despite the acknowledgement of increased ACL injury rates in females at the onset of puberty, few study designs incorporate multiple pubertal categories and/or age ranges into their analyses. To our knowledge, only three studies have investigated sex, pubertal stage and/or multiple age ranges as well as lower limb biomechanics during jump landing maneuvers,[34,35,63] and only one study during a side-cut maneuver.[102] None of these studies investigated lower limb asymmetry during the maneuvers. Inclusion of age as a variable when examining lower limb asymmetry during high-risk ballistic maneuvers may provide important information on each pubertal category, both across sex and within sex. It could provide identification of specific biomechanical criterion that predispose pubescent and post-pubescent athletes, and specifically females, to ACL injury.

1.2.2 Biomechanics of the Lower Limb During Side-Cut Maneuvers

Noncontact ACL injuries account for the majority of all ACL injuries and occur frequently during ballistic sport-specific maneuvers which involve a combination of deceleration and change of direction, as seen in a side-cut.[71–75,89] Retrospective video analyses indicate that a common body position associated with noncontact ACL injury occurs when the tibia is externally rotated and the knee is close to full extension, the foot is planted and deceleration occurs, causing a dynamic valgus.[2,103] Dynamic valgus, a term used throughout this text, is categorized as femoral adduction, or movement of the distal femur toward the midline of the body, in combination with tibial abduction or movement of the distal tibia away from the midline of the body. The link between dynamic valgus knee loading and resultant increases in ACL strain has been demonstrated previously in cadaveric and computer modelling experiments.[82,104–106] These studies are imperative if we are to understand and interpret movement biomechanics during laboratory studies that do not directly measure ACL loading or use ACL injury as an outcome. There have been several laboratory studies addressing the sex bias of ACL injury; many of which, investigate specific discrete variables including hip, knee and ankle joint angles and moments. Though this is a study focusing on risk factors for ACL injury and the knee joint, it is important to understand that the hip, knee and ankle are links in a kinetic chain, thus, all lower-limb joints are important in ACL research and are nicely summarized in a review by Hewett et al.[43] Several biomechanical factors across all planes of motion have been associated with increased knee joint loading and increased risk of noncontact ACL injury.[76–83,85] Though these increased loads have been associated with abnormalities in sagittal, frontal, or transverse plane biomechanics, the greatest loading of the ACL has been shown to occur as a result of loading the knee joint simultaneously across multiple planes.[82–85]ACL loads have been shown to increase as a result of large anterior tibial shear forces,[77] reduced knee flexion,[79,80,82] and increased knee adduction moments and internal rotation moments.[78,82,83,85] At the hip, studies have linked greater hip adduction and internal rotation to dynamic valgus positions during cutting,[107,108] as well as reduced hip flexion (moment and angle), to increased risk of ACL injury.[73,109,110,111] At the ankle, greater eversion angles have been noted for females during a side-cut task,[109,112] which may lead to increased internal rotation of the tibia and dynamic valgus. Additionally, rearfoot and dorsiflexed cutting and landing techniques have been linked with other sagittal plane biomechanical deficits, particularly reduced hip and knee flexion.[113] This erect movement posture may result in the lower limb being incapable or less effective in absorbing ground reaction force (GRF) loads without buckling or collapse of the knee joint.

Though these studies have identified biomechanical deficits that may increase loading to the knee and more specifically to the ACL, no studies have examined hip and knee flexion-extension angles and internal joint moments as they relate to asymmetry between limbs for an athletic population. Nor have any studies attempted to examine these potential biomechanical asymmetries across sex and age to provide insight into the sex bias of ACL injury.

1.2.3 Asymmetries of Lower Limb Biomechanics

It is widely assumed that rudimentary locomotion such as human gait is symmetrical in healthy individuals. A recent publication by Lathrop-Lambach et al. (2014) reported, however, that 55-69 percent of their studied population experienced greater than 10% asymmetry – the clinically relevant return to play criterion[114–116] – in adduction and flexion moments (external) at the knee and hip during healthy adult human gait. Results of that study compliment previous work which had suggested the occurrence of asymmetry and limb dominance in lower-limb joint moments during gait in adult humans without impairment.[97,101] Evidence of bilateral asymmetry in joint moments have also been reported during ballistic maneuvers such as cutting and landing maneuvers, however, this study only studied high school female athletes.[117] As such, it is unclear if asymmetries and characteristics of limb dominance or limb lateralization persist across age and sex groupings. Is asymmetry present for both pre-pubescent and pubescent females as well as in males, and are those asymmetries equal or vastly different between groups?

1.2.4 Asymmetry as a Risk Factor and the One-off Nature of ACL injury

It is well understood that a number of biomechanical factors linked to lower limb motion and knee joint loading have been associated with increased risk of noncontact ACL injury.[76–83,85] Coupled with this, replication of unanticipated ballistic maneuvers in the laboratory have been shown to alter lower limb biomechanics.[86–90] These studies have shown that ballistic athletic tasks such as a side-cut can elicit abnormalities in joint kinematics and kinetics in the sagittal, frontal, or transverse planes, with the greatest loading of the ACL occurring as a result of loading the knee joint simultaneously across multiple planes.[82–85] Thus, simultaneous multi-planar joint loading is a potential driving force for increased ACL injury risk.

The limb dominance theory states that there is an imbalance of muscular recruitment patterns and muscular strength between legs, which leads to differences in dynamic control between limbs.[118,119] Similarly, the concept of lateralization has been previously used to define differences between the operational and support limb. Given that its well understood that lateralization can lead to task-specific roles of the support and operational limb, it is conceivable that developed bilateral strength differences and/or differences in muscle recruitment may lead to the unconscious preference toward greater loading of one limb versus the other, thus resulting in asymmetry of lower extremity biomechanics. If it is understood that: i) the ACL is placed under increased injury risk during multi-planar loading of the knee joint, ii) that the knee joint is largely controlled and regulated by the neuromuscular system, and iii) that differences in muscle strength and muscle recruitment can lead to asymmetry of the biomechanics of the lower extremity, then it is conceivable to believe that asymmetry of lower extremity biomechanics could be a contributing factor to the perplexing 'one-off' nature associated with ACL injury, whereby the ACL is injured while performing a maneuver the athlete may have performed safely countless times prior to injury.

1.3 Purpose

The underlying theme of this study was to perform a comprehensive analysis of lower limb asymmetry of peak sagittal and frontal plane hip and knee internal joint moments, as well as peak sagittal plane hip and knee joint angles for high performance athletes across sex and age during walk, run and side-cut tasks. Sagittal plane joint angles were also analyzed at initial contact. The purpose was to: i) identify if asymmetry existed beyond a clinically accepted 10% threshold for lower limb biomechanical variables of interest in an athletic population of varying age and sex during a walk, run and side-cut task, ii) identify the proportion of each sub population (pre/post-pubescent males, pre/post-pubescent females) that experiences greater than 10% asymmetry for each of the biomechanical variables of interest, and iii) to identify if differences in asymmetry exist across age and sex within an athletic population, to further the understanding of the sex bias of ACL injury as well as recognize how asymmetry may function as an etiological risk factor for ACL injury.

1.4 Objectives

The objectives of this study were:

1) Use a clinically relevant definition of asymmetry to determine if bilateral asymmetry exists in the kinematics and kinetics of the lower-limb for an athletic population in an over-ground walking and running task.

2) Use a clinically relevant definition of asymmetry to determine if bilateral asymmetry exists in the kinematics and kinetics of the lower limb within an athletic population for a ballistic sport-specific maneuver such as an unanticipated side-cut task, that more closely resembles a game-like situation where the ACL is most likely to be injured via a noncontact mechanism.

3) Determine the proportion of participants that experience bilateral asymmetry for each of the biomechanical variables of interest that exceeds the clinically relevant 10% threshold for walk, run and side-cut tasks.

4) Determine the effects of age and sex on the magnitude of asymmetry present in the biomechanical variables of interest for walk, run and side-cut tasks.

1.5 Hypotheses

The hypotheses for this study were:

1) All athletic populations will display bilateral asymmetry in lower limb biomechanics that exceeds the clinically relevant 10% threshold for over-ground walking, running and side cut tasks.

2) Sex and age differences in the magnitude and proportion of lower limb biomechanical asymmetry will exist for a ballistic side-cut maneuver.

3) Post-pubescent females will exhibit the greatest percentage asymmetry for the biomechanical variables of interest during the side-cut task.

4) Pre-pubescent males and pre-pubescent females will exhibit similar levels of asymmetry that are less than both post-pubescent groups.

1.6 Structure of Thesis

Chapter 1 provides a brief overview of ACL injury and the risk factors associated with the injury, including limb asymmetry. The statement of the problem and rationale of this study emphasizes the importance of analyzing the biomechanics of the lower limb asymmetries during ballistic athletic maneuvers. The objectives and hypotheses are also clearly stated within the chapter.

Chapter 2 includes a comprehensive literature review on the ACL anatomy and, more broadly speaking, the knee, the functional role of the ACL, ACL injury mechanisms, sex bias, intrinsic and extrinsic risk factors, etiology of the ACL injury, as well as the role of asymmetry in the lower-limb. Studies specifically relating to these topics are addressed with emphasis on the biomechanics of the lower limb while performing ballistic athletic maneuvers.

Chapter 3 provides a description of the methodology and analysis techniques used in this study. The participants and recruitment of participants are discussed as well as the electromyography (EMG) and movement analysis setup. Data acquisition and the instrumentation used in this study are described in detail, in addition to the walking, running, cutting and jumping protocols. Finally, a detailed description of how calculations of asymmetry were performed, as well as how changes in asymmetry were assessed is presented.

Chapter 4 is a complete manuscript within this thesis titled, "The Effects of Age and Sex on Biomechanical Asymmetries of the Lower Limbs During a Walk, Run and Side-cut Task." This manuscript addresses the prevalence of asymmetry in the biomechanics of the lower limbs during walking, running and unanticipated side-cut maneuvers.

Chapter 5 concludes the thesis by summarizing the findings and presenting the limitations of the present study.

1.7 Significance of Research

Symmetry of lower limb biomechanics is often assumed in a healthy population, and deviations from this are considered to be indicative of pathology[95]. In a clinical setting, functional symmetry is widely used to define goals in rehabilitation and measure the efficacy of treatment. However, Lathrop-Lambach et al. (2014) showed that 55-69 percent of healthy individuals experienced greater than 10% asymmetry in external knee adduction moments during gait. Furthermore, functional asymmetry is also widely used in return to play protocols for recovering athletes.[114–116] However, before we can determine if asymmetry of joint kinematics and kinetics during walking would be an appropriate standard for rehabilitation and return to play for athletes, we must first determine how asymmetrical joint biomechanics are in younger and healthy athletic populations.

This research study will be the first to use a clinically relevant definition of asymmetry to investigate lower limb biomechanics for a healthy young athletic population during gait. This will also be the first study to use a clinically relevant definition of asymmetry to investigate lower limb biomechanics for the same athletic population during ballistic sport-related athletic maneuvers such as running and a side-cut.

Identifying and describing the effects of sex and age on asymmetry of lower limb biomechanics could provide important and highly relevant insight into the one-off nature of ACL injury. Understanding how healthy athletes of different sex and age perform ballistic athletic maneuvers, as well as any underlying trends in asymmetry could enhance our ability to design prevention programs that specifically target a particular age group, and/or sex during their athletic career. This study also has relevance to current ACL rehabilitation programs which strive to achieve functional symmetry between the affected and unaffected limb. If a high proportion of healthy high-performance athletes exhibit greater than 10% asymmetry in joint kinematics and joint kinetics, it begs the question if 10% is an appropriate threshold for biomechanical asymmetry as it relates to a return to play setting.

Chapter 2 - Review of the Literature

2.1 Anatomy of the Knee

It is essential to have a fundamental understanding of the anatomy of the knee to conduct an in-depth analysis of the biomechanical and neuromuscular factors acting at the knee during ballistic athletic maneuvers that often put the knee joint at risk for ACL injury. This modified hinge-type joint is capable of primarily flexion and extension but is also capable of abduction/adduction (valgus/varus), as well as internal and external rotation.[120,121] The knee joint is comprised of two synovial joints; the tibiofemoral joint and the patellofemoral joint. The patellofemoral joint is comprised of the patella that lies superficially on the patellar groove of the femur. The patellofemoral joint acts to allow extension by dissipating 15-30% of the force produced during quadriceps contraction by transferring the force further from the axis of rotation.[120,122] The tibiofemoral joint involves the articulation of the distal femoral condyles and proximal aspect of the tibia, or tibial plateaus.

Unlike the hip, the knee joint relies primarily on the surrounding structures for stability, rather than gaining stability from the congruency of distal and proximal boney segments of the joint. The supporting structures around the knee joint include the surrounding musculature which acts as a dynamic stabilizer, as well as the ligaments, menisci and joint capsule which act passively to stabilize the knee joint. Injury or pathologies that affect both the passive and dynamic stabilizing structures can produce abnormal movement patterns. Thus, the role of the neuromuscular system to adequately control the timing of contraction of the musculature surrounding the knee is imperative to maintaining knee joint stability during athletic maneuvers.

2.1.1 Passive Stabilizers

The menisci (Figure 2.1) are fibrocartilaginous, crescent-like articular structures located on the surface of the tibial plateau that function mainly to distribute and absorb compressive loads.[123] The menisci have been shown to absorb between 50-70% of the load during weight bearing.[123] They are mostly avascular, leading to complications when injuries occur to these structures. In addition to force dissipation, the menisci serve to maintain the joint space and also improve the concavity of the tibial plateaus, thereby increasing the congruency of the joint.[123,124] The medial meniscus is lunar shaped and attaches anteriorly to the intercondylar area of the tibia and posteriorly to the posterior intercondylar area and strongly adheres to the medial collateral ligament (MCL) attachment.[121] The lateral meniscus serves a similar function but is circular and smaller than that of the medial meniscus. The lateral meniscus does not adhere to the tibial plateau as firmly as the medial meniscus and thus allows for greater range of movement and less risk of injury.[121]

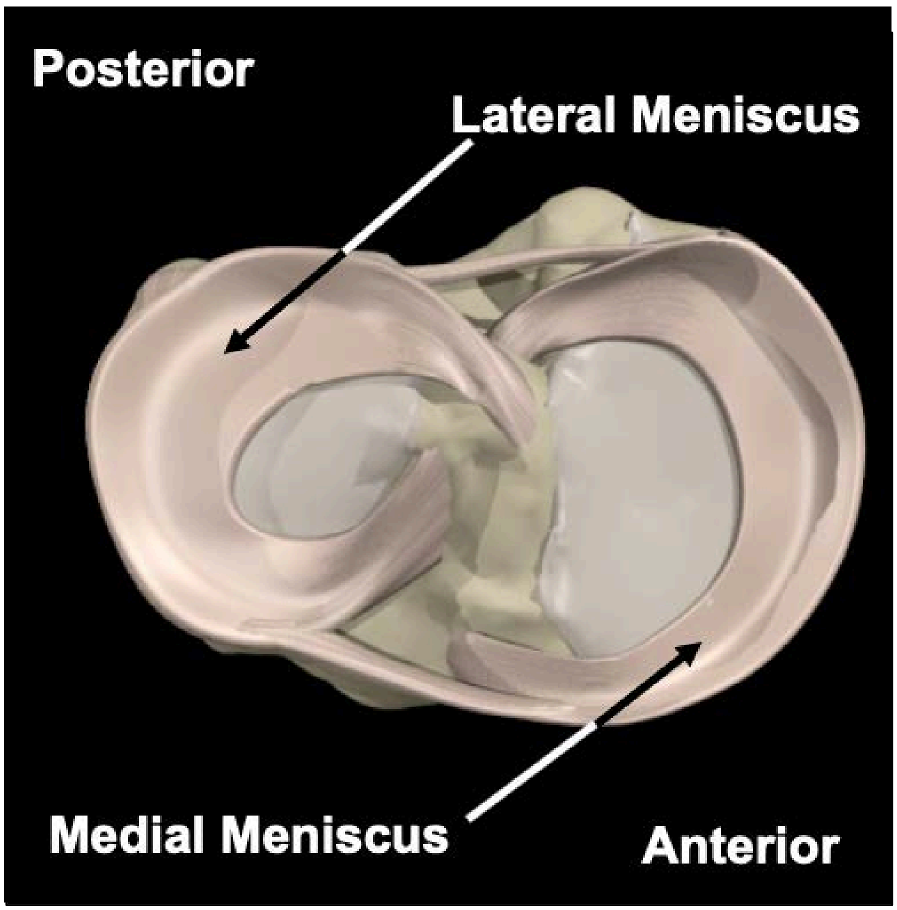

Figure 2.1. Superior view of the right tibial plateau, showing the menisci and cross section of the cruciate ligaments (Interactive Knee 1.66 © 2009 Primal Pictures Ltd.)

Stability of the knee joint is also enhanced by the passive joint capsule which surrounds the joint and attaches to the bony structures nearby. The capsule is created by expansions of the vastus medialis, vastus lateralis and iliotibial band merging with the collateral ligaments of the knee joint.[121] The five external or extracapsular ligaments of the knee that are responsible for reinforcing the joint capsule itself include the patellar ligament, lateral collateral ligament (LCL), medial collateral ligament (MCL), oblique popliteal ligament and arcuate popliteal ligament. The patellar ligament refers to the distal part of the quadriceps tendon that runs superficially over the apex of the patella and attaches at the tibial tuberosity.[121] This ligament serves as the attachment point for the medial and lateral patellar retinacula and aponeurotic expansions of the vastus medialis and vastus lateralis. Combined, these play a crucial role in proper patellar alignment. The Q-angle is a clinical term that refers to knee alignment involving the patella. This angle is created from the line directed from the tibial tuberosity, to the midline of the patella and a second line from the anterior superior iliac spine of the ilium and midpoint of the patella.

The two collateral ligaments that contribute to medial and lateral stability of the knee are the MCL and LCL (Figure 2.2). Both ligaments are in tension at full knee extension. The MCL originates at the medial epicondyle of the femur and attaches to both the medial condyle and superior part of the medial surface of the tibia.[121] The deep fibers of this ligament insert into the medial meniscus. The MCL functions primarily to restrict knee valgus or abduction. Weaker than the LCL, the MCL is more susceptible to injury, subsequently putting the medial meniscus at risk due to the attachment of its deep fibers. The LCL originates at the lateral epicondyle of the femur and runs inferiorly, inserting on the lateral surface of the fibular head and this ligament functions to resist knee varus or adduction. [121]

The oblique popliteal ligament is an expansion of the semimembranosus tendon at the posterior aspect of the medial condyle of the tibia and acts to reinforce the posterior joint capsule. This ligament spans the intercondylar fossa and inserts at the posterior aspect of the joint capsule. The arcuate popliteal ligament functions similarly by also strengthening the posterior joint capsule. It inserts at the posterior fibular head and spans over the posterior surface of the knee joint.[121]

The ACL and posterior cruciate ligament (PCL), along with the medial and lateral menisci, are located within the articular joint of the knee and are referred to as the intra-articular knee ligaments (Figure 2.2). The two cruciate ligaments cross obliquely within the middle of the joint capsule. The ACL originates at the anterior intercondylar surface of the tibia and inserts superiorly, posteriorly and laterally on the posterior part of the medial side of the lateral condyle of the femur. The ACL primarily prevents anterior translation of the tibia on the femur but also functions to prevent internal knee rotation, hyperextension, valgus, and varus movement to a lesser extent. The PCL is the stronger of the two cruciate ligaments and serves to prevent posterior displacement of the tibia on the femur and hyperflexion. The PCL is the main stabilizer during weight bearing and flexed knee positions.[121]

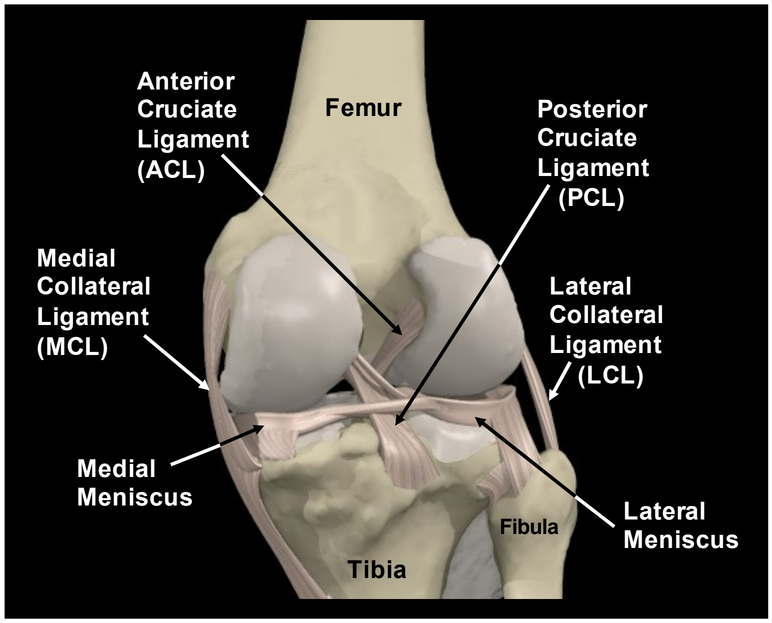

Figure 2.2. Posterior view of the passive stabilizers of the right knee showing the two menisci, four primary knee ligaments and main bones excluding the patella (Interactive Knee 1.66 © 2009 Primal Pictures Ltd.)

2.1.2 Dynamic Stabilizers

As a result of the knee's relatively incongruent articular surfaces, the role of the neuromuscular system to adequately control the timing of muscle contraction in the musculature surrounding the knee is imperative to maintaining knee joint stability during athletic maneuvers. Figure 2.3 shows the three main muscle groups that cross the knee joint: the gastrocnemii (medial and lateral), the hamstrings (biceps femoris, semitendinosus, and semimembranosus) and the quadriceps (vastus lateralis, vastus medialis, vastus intermedius and rectus femoris). The quadriceps muscles are responsible for guiding extension of the knee joint. As the knee joint comes to full extension the tibia undergoes a small rotation relative to the femur. This rotation serves to lock the knee into its fully extended and most stable position. At this point the articular surfaces of the femur and tibial plateau are at their highest congruency and the primary ligaments surrounding the knee are rigid or tight.[121] Conversely, the hamstring group – referring to the biceps femoris, semitendinosus, and semimembranosus – are the primary muscles responsible for flexion as they cross the knee joint. Each of the hamstring muscles originate at the ischial tuberosity and insert on the lateral aspect of the fibula head in the case of the biceps femoris and at the medial aspect of the tibial condyle in the case of the semitendinosus and semimembranosus. Internal and external rotation of the knee joint is also guided by the hamstring muscle group. The biceps femoris guides external rotation of the knee joint while internal rotation of the knee joint is facilitated through the muscles which insert on the medial aspect of the tibial condyle, including the semitendinosus, semimembranosus, gracilis and sartorius. The gastrocnemii also function as weak flexors, however, their primary role is to stabilize the tibia in synergy with the soleus and plantaris, thus possibly serving a protective role to the knee joint.[121,125]

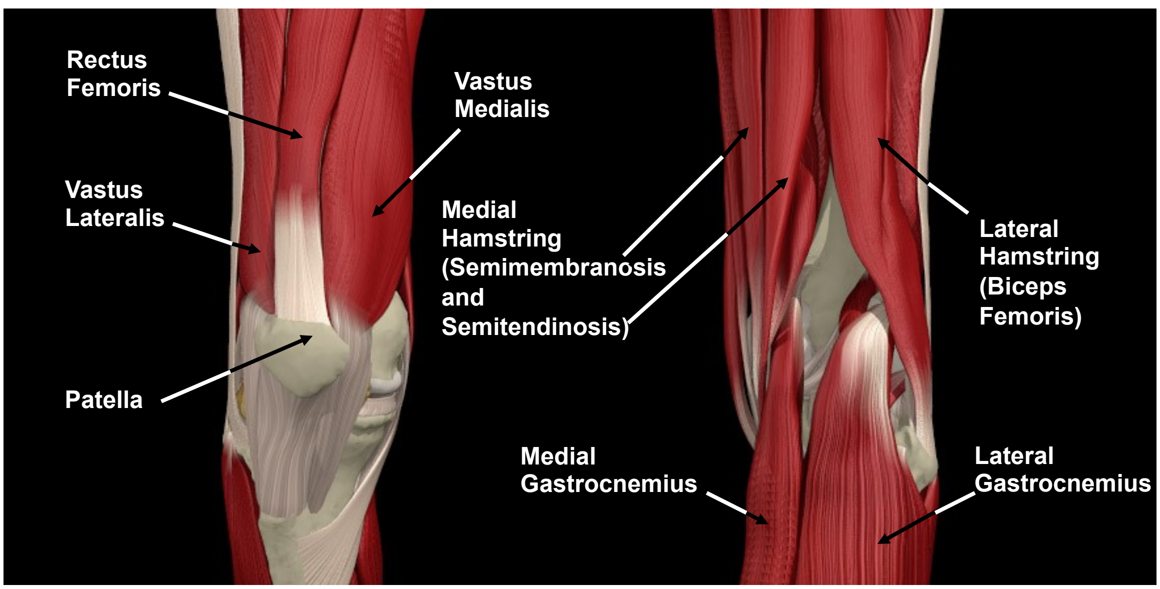

Figure 2.3. Musculature surrounding the right knee joint. Left: Anterior view; Right: Posterior view (Interactive Knee 1.66 © 2009 Primal Pictures Ltd.)

2.2 The Anterior Cruciate Ligament

The ACL is one of four primary knee ligaments found in the knee joint complex and is one of the most commonly injured structures in sports involving ballistic movements such as cutting, deceleration or jump landings.[43,73,126] A firm understanding of the anatomical characteristics of the ACL is crucial for comprehending the function of the ligament as well as the mechanisms in which it can be injured.

2.2.1 ACL Anatomy

The ACL is a band-like structure of dense connective tissue located centrally within the knee joint. The ACL originates at the medial aspect of the femoral condyle and inserts slightly anterior and lateral to the medial tibial eminence. The ACL has a spiral orientation and is referred to as a cruciate ligament because it forms an "X", or "cross" with the other ligament located centrally in the knee joint, the PCL. The cruciate nature of the ACL and PCL is critical to constraining joint motion and providing additional stability within the knee joint. Although the ACL is located intra-articularly, the ligament has its own synovial envelope that allows the ligament to be extrasynovial. Phenotypic characteristics such as length and width are dependent on individual anatomy, however, the ACL is often found to be an average of 31-38 mm in length, with the tibial attachment being slightly wider (approximately 12 mm) to increase stability at the tibial insertion point.[127,128]

It has been widely accepted within the literature that the ACL can be divided into two distinct groups or bands, each having a different function throughout the range of motion.[128] The anteromedial band inserts on the anteromedial aspect of the tibia and originates proximally on the anterior aspect of the femoral attachment site. Conversely, the posterolateral bundle originates at the distal posterior portion of the femoral attachment site, while inserting at the posterolateral aspect of the tibial attachment site. The anteromedial and posterolateral bundle bands move relative to one another as the knee is flexed and extended. During knee flexion, the posterolateral bundle band has been shown to be relaxed while the anteromedial band is under greater tension. Conversely, under full extension, the anteromedial band is relaxed while the posterolateral bundle band is under tension.

2.2.2 Innervation and Vascularization

Branches of the middle genicular artery provide the majority of the ACL's vascular supply, with medial and lateral inferior genicular arteries supplying the distal position of the ligament to a lesser extent. The genicular artery runs posterior to the knee joint and as such, passes through the posterior aspect of the knee joint capsule. Within the joint capsule, the artery branches to support the synovial plexus, from which small vessels travel to the soft tissues found within the intercondylar notch, such as the ACL.[121,128,129] With the advent of injection techniques and immunochemistry, the areas of insertion of the ACL have been determined to be avascular. As such, with a diminished blood supply the ACL has a poor ability to heal in the event of a lesion or injury.[128]

Overall homeostasis and healing of the ACL is regulated by a small group of mechanoreceptors that heavily innervate the ligament. These receptors act by converting mechanical stresses of the ligament into nerve action potentials. The mechanoreceptors that innervate the ACL include, primarily, Ruffini receptors which function as stretch receptors as well as free nerve endings which act as nociceptors. Pacini receptors, which detect compression forces, also innervate the ligament but to a much lesser extent. Shultz et al. (1984) was the first group to provide a description of mechanoreceptors of the human ACL and suggest possible proprioceptive function.[130] It has since been shown that mechanoreceptors are able to track acceleration at the initiation and termination of movements as well as monitor joint motion, position and angle of rotation.[131–134] Furthermore, the ACL mechanoreceptors have been hypothesized to sense the limits of knee motion and influence the neuromuscular system by either the inhibition and/or stimulation of agonist muscles and antagonist muscles.[135,136] Johansson et al. (1991) has also demonstrated that the ACL mechanoreceptors influence muscle tension and thus can modify overall knee joint stiffness.[137]

2.2.3 Functional Role of the ACL

The cruciate ligaments play an essential role in the biomechanics of the knee joint by both stabilizing and guiding the joint through the six degrees of motion; three rotations and three translations. The rotations that occur include flexion/extension, valgus/varus (abduction/adduction), and internal/external rotation, while translations occur anterioposteriorly, medialolaterally and in compression/distraction.[120,138] The main function of the ACL is to act as a primary restraint of excessive anterior translation of the tibia relative to the femur. Secondarily, the ACL functions as a stabilizer of internal rotation of the tibia relative to the femur, helps prevent hyperextension, and plays a small role in resisting varus/valgus stress on the knee joint.

In addition, the cruciate nature of the ACL fibres in conjunction with the shape of the femoral condyles facilitate the "screw-home" mechanism of the knee joint by assisting with the external rotation of the tibia throughout the latter 20 degrees of extension.[127,138]

2.3 ACL Injury Prevalence, Etiology and Emerging Sex-Bias

Since the inception of Title IX of the United States Educational Assistance Act of 1972, female participation in high school sports has increased by over 1100%.[139] With participation in sport increasing annually, so too is the number of traumatic knee injuries. Knee injuries make up as much as 22% of all injuries sustained, with ACL injury being among the most common.[140] A 2006 review by Griffin estimates that between 250,000 and 400,000 ACL injuries occur annually in the United States alone.[22] A population adjusted estimate of ACL reconstructions indicated that ACL reconstruction rates, in the United States alone, have increased by 37% between 1994 and 2006.[23] This increased injury prevalence in combination with a 2-8 times greater ACL injury rate in females[27,28,31] has resulted in an overall sizeable increase in female ACL injuries for female athletes 14-23 years of age.[141] In North America, basketball, soccer, volleyball and football show the highest rates of ACL injury,[142] as each of these sports involve some degree of ballistic maneuvers (e.g. cutting, decelerating and jumping) that put the ACL at risk of injury.[2,26,29–36] Factors such as anticipation[86] and limb dominance[141] may also play a role in the susceptibility of the ACL to injury during these maneuvers.

Evidence also exists that ACL rupture is associated with accelerated development of knee osteoarthritis in the affected knee.[13–16,143] Due to the traumatic nature of the injury, and subsequent instability of the knee joint, ACL injury often requires surgical intervention and extensive rehabilitation for the affected individual. As such, the implications of ACL injury are therefore burdensome not only to the affected individual but to the health care system as well. The resulting annual cost of ACL reconstruction and subsequent rehabilitation is estimated to be in excess of 2 billion dollars annually in the USA.[24,25]

2.3.1 ACL Injury Etiology and Mechanisms

What is most troublesome, is that 70-80% of ACL injuries occur via noncontact mechanisms where there is no direct blow to the knee joint.[2,81] Thus, an understanding of noncontact ACL injury mechanisms is imperative if future ACL injuries are to be reduced or prevented. Within the literature, five ballistic sporting maneuvers have been defined as noncontact ACL injury inducing and these include: planting and cutting, straight knee landing, sudden deceleration, pivoting and one-step stop landings with knee hyperextension.[144] Literature has demonstrated that these high-risk ballistic athletic maneuvers increase three types of excessive knee joint loading that place additional strain on the ACL[76–81] including, anterior tibial shear force,[77,79,81] (ii) internal knee abduction moments[78,79,81] and (iii) internal tibial rotation moments.[77,79,81] Though increased loads can be caused by abnormalities in sagittal, frontal, or transverse planes, the greatest loading of the ACL has been shown to occur as a result of loading the knee joint simultaneously across multiple planes.[82–85] Despite the understanding of knee joint loading across multiple planes, the magnitude of loading that most frequently results in noncontact ACL injury is less understood and remains controversial.

Video analysis studies have demonstrated markedly similar knee kinematics between individuals during ACL injury. Though the capability to accurately characterize lower-limb biomechanics from video is limited, the literature provides evidence that supports a multiplanar mechanism of injury.[145] At the time of initial contact, individuals land in neutral frontal plane alignment, with their knee near full extension and slight external tibial rotation.[145] Within the first 100 milliseconds of initial contact, the time frame in which injury is most prevalent, the tibia begins to externally rotate relative to the femur and the knee joint undergoes dramatic abduction or valgus collapse. In a cadaveric study by Kiapour et al. (2014), multiplanar loading of the knee joint generated significantly greater strain in the ACL and was able to reproduce ACL rupture in nearly 90% of the specimens.[81] Thus, it is imperative to develop and implement neuromuscular training regimens that address this injury mechanism by attenuating excessive multiplanar knee joint loadings, or deficiencies in knee joint kinematics during athletic maneuvers.

2.3.2 Maturation and Emergence of a Sex Bias

Collectively, the literature shows little evidence to demonstrate a sex bias in ACL injury rates in pre-pubescent athletes,[26,36,37,146] which is in direct contrast to pubescent[30,32,34] and collegiate level or post-pubescent athletes.[29,31] The 2006 study of Mihata et al. (2006) examined data from the National Collegiate Athletic Association (NCAA) Injury Surveillance System, to investigate ACL injuries of basketball and soccer athletes over a 15-year period.[33] The results of this study found that ACL injury rates for females were nearly threefold compared to their male counterparts. This review supports previous research that indicated females experience a greater ACL injury rate than males following the onset of puberty.[27–29,31]

However, knee injuries occur frequently across all ages including pediatric and pre-pubescent athletes. Gallagher et al. (1984) found that 63% of injuries to children aged 6-12 years old were classified as joint sprains with the majority occurring at the knee.[26] Andrish (2001) postulated that the rise in the sex disparity of ACL injury incidence parallels with the onset of adolescence due to the rapid musculoskeletal growth that occurs for children at this time of development.[37] During this growth spurt, the long bones of the lower extremity, including the femur and tibia, have been shown to grow at a rapid rate for both males and females.[147] This growth translates into increased potential for higher torques at the knee joint by creating longer lever arms both on the distal and proximal end of the joint. This growth also leads to a change in center of mass of the athlete, leading to greater un-coordination as muscular control of the body becomes more challenging. The increase in center of mass is also associated with decreased balance in athletes during ballistic sport maneuvers, subsequently causing movements to become more difficult for the neuromuscular system to control, thus increasing injury risk. Prior to puberty, little sex disparity exists,[146] and ACL injuries only account for 0.2% of all knee injuries to girls and boys between the ages of 5-10 years of age.[148] Thus, differences in pubertal development have be attributed to the increased ACL injury rates in females particularly after the onset of puberty.

Following the onset of adolescence, Kellis et al. (1999) demonstrated a correlation between an increase in power and strength and chronological age in males, while females demonstrated little correlation or change in neuromuscular contributions throughout adolescence.[149] The study, cross-sectional in nature, investigated changes in vertical jump height for both male and female basketball players aged 13-30 years old.[149] Results indicated that female basketball players did not increase vertical jump performance by a statistically significant margin, while males showed a statistically significant increase in vertical jump performance with increasing age.[149] Additional findings from the study showed that females often demonstrated decreased dynamic knee stability compared to males following their rapid growth spurt.[149] These results suggest that the rapid increase in height and lower extremity bone length, coupled with a lack of power and strength development leads to deficits in neuromuscular control for adolescent females. This lack of neuromuscular control may expose the passive and active stabilizers of the knee to greater forces and torques during ballistic maneuvers, which could account for biomechanical alterations and increase the risk of ACL injury in adolescent females.[147]

Supporting these results is a recent 2015 study, where Hewett and colleagues hypothesized that the rapid increases in lower extremity bone length in the absence of sufficient neuromuscular adaptation are related to a decrease in dynamic knee stability in female athletes. The study examined the correlation between growth, as measured by peak height velocity, and high joint load biomechanics that result in poor dynamic knee stability, as measured by peak knee abduction moment, on 865 adolescent female soccer and basketball athletes over a two year span.[35] Results from this study showed that peak knee abduction moment increased for both sexes after the onset of puberty, indicating a loss of dynamic knee stability. However, males regained dynamic knee stability following a neuromuscular spurt at 91% of adult stature (the point of peak height velocity), whereas dynamic knee control in females continued to decrease.[35] This study also showed that pre-adolescent athletes displayed a similar quantity of knee valgus loading between both males and females, however, females displayed greater valgus loading at late adolescence.[35] It can be postulated that the increase in peak knee abduction moments during adolescence is attributed to altered neuromuscular control of lower extremity biomechanics in the coronal plane. This may be caused by altered neuromuscular control and firing patterns of the adductors and abductors of the knee and hip joints. Hewett and colleagues stated that this demonstrates a clear relation between maturation and the tendency for high-risk female athletes to preferentially utilize an increased frontal plane loading strategy as opposed to a sagittal plane load absorption strategy via increased muscle activation in flexion-extension during ballistic sporting maneuvers.[35] In combination, frontal plane loading and increased peak knee abduction moments can inhibit desirable knee joint biomechanics, destabilize the joint and increase load and injury risk to the ACL in adolescent female athletes.[35]

Together, these studies support the notion that increases in lower extremity bone length (which occur primarily during adolescence), paired with the absence of sufficient neuromuscular adaptation and control are related to a decrease in dynamic knee stability and increased knee joint loading for adolescent females. Thus, these studies highlight the fact that anatomical changes during adolescence may underlie biomechanical changes such as knee joint loading and may contribute to the increased risk of ACL injury in pubescent and post-pubescent females in particular.

2.4 Risk Factors for ACL Injury

Research aimed at identifying ACL injury risk factors may aid in improving the current understanding of injury mechanisms, sex disparity and injury prevention strategies. Numerous factors are speculated to increase the risk of sustaining a noncontact ACL injury, and can be categorized as intrinsic or extrinsic. Both extrinsic and intrinsic risk factors may account for differences in asymmetry, the effects of anticipated and unanticipated cutting and ultimately influence the injury rates between sex. While it is beyond the scope of this literature review to describe in detail all the risk factors that have been identified within the literature, many are worth highlighting and will be discussed below.

2.4.1 Intrinsic Risk Factors

Intrinsic risk factors are those which originate due to factors within the body and may be unique to each body, including the physiological and anatomical differences between individuals. As such, intrinsic risk factors can be sex specific and are generally considered uncontrollable. These factors can be classified as anthropometric, biomechanical, neuromuscular, fatigue, genetic and hormonal influences.

Anthropometric

Anthropometric differences between individuals, especially between sexes, that have been suggested to put athletes at risk of ACL injury include Q-angle, femoral intercondylar notch width, joint laxity and posterior tibial slope.[150–152]

The Q-angle is determined by measuring two lines superimposed on the lower extremity. One line runs from the midpoint of the patella to the center of the tibial tuberosity, while the other is measured from the anterior superior iliac spine to the center of the patella.[153] It is commonly used to measure the resultant forces on the patellar tendon by quantifying the force of pull created by the quadriceps on the patellar tendon.[153] Thus, an individual's Q-angle will vary through their lifetime based on pubertal changes. While pre-pubescent athletes show little variance in Q-angle between sex, post-pubescent female athletes have been shown to have a greater Q-angle due to an anatomically greater hip width relative to leg length.[153,154] Conventionally, the Q-angle is believed to place the knee in a position of greater knee valgus which has been proposed to predispose those with a high Q-angle to knee injury. Work by Ferber et al. (2003) suggested that a greater dynamic Q-angle in females is associated with a larger hip internal rotation angle in combination with greater knee abduction angles during movement.[155]

The link between Q-angle and increased knee injury continues to be debated in the literature. Hertel et al. (2004) concluded that knee injury rates are not related to Q-angle, and that static Q-angle is not a predictive measure of ACL injury risk during dynamic tasks.[156] However, previous work which looked at a cohort of recreational basketball players found the average Q-angle for athletes which sustained a knee injury was much greater than those who remained uninjured.[157]

Femoral intercondylar notch width has also been linked to possible ACL injury. Anatomically, the ACL is situated within this small notch originating at the medial aspect of the femoral condyle and inserting slightly anterior and lateral to the medial tibial eminence. It has been proposed that a smaller notch width may be indicative of a smaller and weaker ACL, thus leading to failure of the ACL.[39–42,158] Smaller notch width dimensions have been widely reported as a sex specific trait among females[40,42] and are thus thought to play a role in the sex bias shown in ACL injury rates. Shelbourne et al.

(1998) suggested the absolute size of the ACL predisposes those with a narrow notch width to ACL injury, not the narrow notch width itself.[40] Intuitively, this hypothesis holds merit as a smaller ACL would have less mechanical strength than a larger one, and thus would rupture sooner under the same loading conditions. However, results from Uhorchak et al. (2003) showed that notch width itself contributes to the incidence of ACL injuries.[42] Measurement techniques of notch width – radiographs, MRI, calipers in cadavers – vary significantly between studies and have thus been attributed to the variability of results between studies. As such, the true role of the intercondylar notch width remains unclear as to whether it affects ACL injury incidence.

Since 1970 is has been postulated that joint laxity may predispose athletes to knee injuries.[159] More recent studies have supported this hypothesis, finding that loose jointed individuals suffer more ligamentous knee injuries.[42,52–54,160] Quatman et al. (2008) examined the effects of pubertal status on generalized joint laxity in a population of male and female athletes.[160] Results from this study demonstrated that joint laxity scores were greater in post-pubertal females compared to males, despite showing no differentiation between sex in pre-pubertal groups.[160] These authors hypothesized that the structural and physiological changes that occur during puberty, including alterations in passive joint restraints, may affect the severity, type and incidence of injuries in the pubescent adolescent population. Additionally, it has been reported that joint laxity may contribute to higher ACL injury incidence as the subsequent hyperextension and valgus motion of the joint results in greater-than-normal loads on the ACL.[94]

Tibial slope is defined as the angle between the line perpendicular to the tibial axis and the posterior inclination of the tibial plateau.[151] Recent evidence suggests a strong relationship between the angle of posterior tibial slope and ACL injury risk.[151,152,161–163] Dejour et al. (1994) used radiological tests and determined that every 10 degree increase in tibial slope results in 6 mm of anterior tibial translation during a monopodal stance test.[161] A radiographic study by Marouane et al. (2014) also found that a 10 degree increase in tibial slope results in 6 mm of anterior tibial translation in a stance test and 3.5 mm in the Lachman Test.[163] Meanwhile, a study using MRI by Beynnon et al. (2014) found there is an astonishing 21.7% increase in noncontact ACL injury rate for each degree increase in the lateral tibial slope in females.[152] Together these results support the notion that increased posterior tibial slope should be considered as an important risk factor for ACL injury during activities with compression forces.

Biomechanical

Rupture of the ACL occurs when the forces applied to the ligament exceed the structural properties in which the ligament can withstand. Dynamic factors thought to influence ACL loading are knee kinematics (joint angles) and kinetics (moments). Numerous biomechanical risk factors related to lower limb kinematics and kinetics have been linked to increased ACL loading and noncontact ACL injury risk.[78,80–83,85,164] Studies using video analysis have observed ACL injuries occurring with limited knee flexion, combined with valgus collapse and tibial rotation relative to the femur.[4,103,145] Experimental studies, both in vivo and in vitro, support these observations where knee valgus and tibial rotation moments[78,82,83,85] and reduced knee flexion[79,80,82] have been shown to increase loading of the ACL. It is important to remember that the knee is only one part of the kinetic chain; therefore, other segments of the lower limb, including the ankle, hip and trunk, may play a role in ACL injury.

The biomechanics of the frontal plane, specifically at the knee are widely reported in the literature. In vitro, in vivo and computer modelling experiments have all demonstrated a link between dynamic knee valgus and increase in ACL loading and strain.[82,104,105] In a prospective and combined biomechanical-epidemiologic study by Hewett et al. (2005), the authors noted knee abduction angles were eight degrees greater during a drop jump task for female adolescent athletes who went on to experience ACL injury. In addition, this study showed the future ACL injury sample experienced higher peak knee abduction moments as well, thus leading the group to infer that both abduction angles and moments were significant predictors of future ACL injury risk.[150] Greater knee abduction angles have also been noted in female athletes during a cutting maneuver.[5,112]

Frontal plane contributions of the hip and trunk may put the knee in a position of dynamic valgus, characterized in the literature as a bodily position in which the knee collapses medially from excessive valgus and/or internal-external rotation.[4,165] Positions of dynamic valgus (or knee abduction) occur as a product of tibial abduction with femoral adduction and internal rotation; all of which have been associated with peak knee moments during cutting tasks.[107,108] Fukuda et al. (2003) demonstrated that torques caused by dynamic valgus can increase anterior tibial translation, resulting in significant strain and load on the ACL.[166] Additionally, hip angles during landing have been shown to be key determinants for impact forces at the knee joint.[92]

Few studies have reported their results on frontal plane ankle biomechanics during ballistic maneuvers. Interestingly, two studies have shown consistent results in that female athletes display greater peak ankle eversion than males during a cutting task.[109,112] Intuitively, excessive eversion of the ankle joint may account for increased rotation of the tibia and thereby lead to a dynamic valgus collapse of the knee joint during these activities. Despite these results, the role of frontal plane ankle biomechanics is relatively unexplored. Future research should look to expand this field to fully understand the impact of frontal plane biomechanics of the ankle joint and its influence on noncontact ACL injury risk.

Several studies have focused on sagittal plane biomechanics and specifically knee flexion angles when performing sport tasks. Hip and ankle joint flexion during the stance phase of a ballistic athletic maneuver allows for greater force dissipation and therefore less resultant force to be dissipated solely by the knee joint. As such, an erect lower limb posture – as a result of reduced hip and knee flexion along with reduced plantar flexion – may reduce the ability of the body to mitigate forces acting at the knee. This hypothesis has been supported by several in vitro[79,82] and in vivo[80] studies that demonstrate increased peak knee strain occurs when knee flexion is reduced. Further research indicates that the ACL is most vulnerable to rupture within the first 30 degrees of flexion.[2,3,82,103,167] A review by Shimokochi et al. (2008) examined both in vitro and in vivo studies and concluded ACL injury prevention programs need to promote an increased knee flexion angle during sudden deceleration tasks.[168]

Elsewhere in the kinetic chain, but also in the sagittal plane, studies have shown that elite female soccer players exhibit reduced external hip flexion moments and hip flexion angles during an unanticipated cutting maneuver.[73,109] These results were supported by additional literature showing similar results during a jump landing task.

Female soccer players over the age of 13 years were reported to have decreased hip flexion angles compared to males in both a drop jump and stop jump task.[110,111] Landing with decreased hip and knee flexion is thought to increase the risk of ACL injury because greater peak forces are transmitted to the knee in a more upright position.[169] Recent research examining this correlation showed hip and knee flexion-extension angular velocity were strongly correlated to peak vertical ground reaction forces.[170]

Sagittal plane movement at the ankle has also been shown to have biomechanical implications elsewhere in the kinetic chain. Landing with the rearfoot, or in a more dorsiflexed position, has been associated with less hip and knee flexion than a plantarflexed forefoot landing.[113] Intuitively, if the ankle is more plantarflexed at contact, the ankle joint, and the surrounding musculature such as the gastrocnemii, can absorb more of the force than if the landing occurs with increased dorsiflexion. When landing in dorsiflexion, that link (the ankle) can be removed from the lower limb model, leaving a two-segment model with those segments aligning distally and proximally to the knee joint. Thus, this 2-segment approach may be incapable of absorbing ground reaction force loads without the associated gastrocnemii activation, which could lead to buckling at the knee joint due to excess loading.

Both internal and external rotational motions and moments have been identified as possible contributors to the ACL injury mechanism. At the hip joint, greater internal rotation maximum displacement has been observed in female collegiate athletes than in males.[171] This finding is generally considered to be the result of weak hip musculature control by the female hip joint. Significantly lower gluteal activation in female athletes has been noted in jump landings, suggesting females have less control over the hip joint during sport-specific maneuvers.[172] Furthermore, training of the gluteal muscles has been suggested as a method to decrease hip internal rotation and may be effective in preventing knee valgus positioning.[172,173]

At the knee, Markolf et al. (1995) showed that the application of a knee internal rotation moment increased knee joint loading more than an applied external rotation moment when combined with an anterior shear force.[82] Larger ACL strain has also been measured during a weight bearing task with an applied internal rotation moment in comparison to loading due solely to weight bearing.[21] These results suggest that the combination of normal weight bearing and the introduction of an internal rotation moment may place the ACL at increased risk of injury due to this increased ACL strain. The studies reporting an increase in ACL loading become particularly intriguing given that transverse plane motion has been observed in studies using video analysis during the moment of injury.[103,145] Together, these results infer that transverse plane knee biomechanics act in combination with the biomechanics of the knee joint in other planes, to place increased strain on the ACL and thus increase injury risk.

Though these increased loads can be caused by abnormalities in sagittal, frontal, or transverse plane biomechanics, it is important to understand that rupture of the ACL is multifactorial and is likely the result of deficits in biomechanics across multiple joints and in all planes. The greatest loading of the ACL occurs as a result of loading the knee joint simultaneously across multiple planes[82–85] and generally arises in dynamic settings where athletes are under deceleration during a cut, change in direction or jump landing. The ACL becomes loaded when an anterior directed force is applied to the tibia. Sakane et al. (1997) showed the relative proportion of anterior shear force, transferred to the ACL as tensile force, at various flexion angles.[167] Results from this study showed that nearly 82% of the original applied anterior shear force could be observed as tensile force in the ACL at flexion angles less than 30 degrees. The relative percentages decreased as knee flexion angle increased.[167] These results are in agreement with previous work that showed the ACL is the major anatomical restraint against anterior shear forces applied to the tibia.[174] The findings pertaining to anterior shear force raise an important question regarding the musculature surrounding the knee joint, as anterior forces may be influenced by forces generated by the quadriceps, hamstrings and gastrocnemii in a dynamic setting.

Neuromuscular

Neuromuscular control refers to the interaction between the neural and muscular systems to coordinate and control movement and posture of the body. As highlighted by the previous section, there has been extensive research surrounding movement strategies that may influence ACL injury risk. Likewise, there is considerable research examining the underlying neuromuscular factors that contribute to biomechanical deficits or potential injurious movement patterns. Poor neuromuscular control of the musculature surrounding the hip and knee act to provide support against external loads on both the passive and dynamic stabilizers of the joint, and help to stabilize the lower extremity joints during dynamic movements.[175] As such, it is important to consider the activation strategies of these muscle groups and their impact on ACL injury risk.

Due to their attachment site, the hamstrings act as an agonist to the ACL by preventing excessive anterior translation of the tibia relative to the femur.[153] Conversely, the quadriceps serve as an antagonist to the ACL and in isolation can increase anterior shear force on the tibia.[153,174] As such, the co-contraction of the quadriceps and hamstring muscle groups has been extensively studied in the literature. A study by More et al. (1993) simulated quadriceps and hamstring loads in a cadaveric model during a squat exercise.[176] Results showed that anterior tibial translation was significantly attenuated when the hamstring muscle group was loaded. These results demonstrate that the hamstrings can have a protective effect against ACL injury by preventing anterior translation of the tibia. The concept of hamstring co-contraction as a protective mechanism to ACL injury is an enduring clinical concept since the hamstrings provide a posterior shear force that is to counter the anterior shear force applied by the quadriceps. However, research within the field has shown some resistance to these findings. Several research groups have shown the co-contraction of the hamstrings is not great enough to diminish ACL loading.[177–179]

In addition to agonist-antagonist interactions of the thigh muscles, medial-lateral imbalances in neuromuscular activity have also been examined in the literature. Due to their anatomy and insertion points, the medial and lateral musculature surrounding the knee has been shown to be capable of affecting knee biomechanics in the frontal and transverse planes.[175,180,181] Increases in medial quadriceps and lateral hamstrings have been observed in a pre-planned side-cut task, suggesting the co-contraction is to attenuate the knee abduction and internal rotation moments at the knee during this maneuver, however, inconsistent firing of the lateral hamstrings has been demonstrated in female athletes.[53] In addition, Myer et al. (2005) showed a decreased ratio of medial to lateral quadriceps recruitment.[182] This deficit in the quadriceps in conjunction with unbalanced hamstring activation may result in dynamic knee valgus during sport-specific tasks.

Moreover, it has been shown that low medial-lateral quadriceps activation in conjunction with increased lateral hamstring firing can alter the compressional force of the knee joint and decrease joint compression.[53,82] Decreased joint compression may limit passive resistance to dynamic valgus and anterior tibial translation, placing increased loads on the ACL when performing a sport-specific movement such as cutting or landing from a jump.[106,183]

The neuromuscular contribution to ACL injury risk is not solely limited to co-contraction and medial-lateral deficits. Both the magnitude and timing of muscle firing can play a role in the neuromuscular impact on ACL injury. EMG studies have reported sex differences in muscle activation during sport-specific movement,[53,182,184] including in a side-cut task.[86,175] Work by Besier et al. (2001) showed that external valgus and internal rotation moments increased in unanticipated cuts compared to pre-planned cutting tasks.[86] This increased loading during the unanticipated condition was postulated to be the result of differences in the timing of the neuromuscular system. In a later publication by the same group, neuromuscular activity was measured for both anticipated and unanticipated conditions.[175] The group measured two cutting tasks versus a run and compared the magnitude of muscle activation during three phases of the stride of the cut; pre-contact, mid-stance and peak push-off. Results from this investigation showed increases in muscle activation magnitude across all three phases for both cutting tasks relative to the run task. Additionally, during pre-contact the group noted that activation of the medial muscle group of the quadriceps (vastus medialis) increased by up to 33% in the pre-contact phase of the anticipated cut.[175] Likewise, the activation of the semi-membranosis was also increased, suggesting that activation patterns may be pre-programmed to prevent the knee joint from de-stabilizing. During unanticipated conditions, the average muscle activation increased between 10% and 25% relative to the preplanned task, however joint loads were estimated to increase by 70%.[175] It was hypothesized that the increase in muscle activation cannot attenuate the significant increase in joint loading, placing the knee at increased risk of injury during these dynamic unanticipated situations.

Sex differences during unanticipated side-cut tasks have also been identified in the literature.[5,73] In one specific study, females adopted a different motor unit recruitment strategy during both the pre-contact and initial contact phase, resulting in lower frequency components of the EMG signal in the lateral hamstring.[5] This strategy may play a role in explaining the sex bias in ACL injury rates through altered neuromuscular control strategies across sexes.

Fatigue

Fatigue has widely been reported to play a role in noncontact injury mechanisms, both at the knee and in other joints. The properties of fatigued muscles have been shown to resemble properties of weak or untrained muscles when examining the role of hip musculature on knee mechanics.[172] The notion of fatigue-altering knee mechanics has been supported elsewhere in the literature. A study by McLean et al. (2007) examined the impact of fatigue on sex-based landing strategies using NCAA Division I athletes. In this study, participants performed a 10 drop-jump protocol where they were instructed to fall off a 50 cm platform and explode to a max vertical jump upon landing. Participants then took part in a generalized lower-limb fatigue protocol before completing the same 10 drop-jump protocol. Results showed that fatigue increased peak knee adduction moments and internal rotation moments in both males and females, and increased knee abduction moments in females.[185] The additional findings in females are increasingly concerning as they may provide insight into the sex dimorphic rate of injury. The biggest limitation to this study was the sample size. Results were drawn from 10 male and 10 female athletes, across three sports, and thus may not be indicative of the entire athletic population. A similar study by Borotikar et al. (2008) examined fatigue and decision-making on NCAA female athletes performing single leg jump landings. The sample size (n=24) was determined using a power test based on the results reported by McLean et al. (2007), and three additional studies. Findings showed that fatigue caused significant increase in hip extension and internal rotation at initial contact, as well as increased knee abduction and internal rotation during stance.[186]

Fatigue was also shown to account for variations in kinematics and kinetics in a slightly larger (n=30) single-leg drop jump study of recreational athletes by Kernozek et al. (2007).[187] Both males and females demonstrated increased hip flexion at initial contact, however females presented upwards of 33 percent greater flexion than their male counterparts. At the knee, females showed less maximum knee flexion than males during pre-fatigue trials. Post-fatigue, however, knee flexion angles increased for males but did not change in females.[187] With knowledge of the sex dimorphic rate of injury, these results suggest that this additional knee flexion may be protective of ACL rupture in male athletes. The authors also reported that females exhibited greater anterior shear force post-fatigue, which may also be indicative of ACL rupture.[187]

Fatigue has also been shown to affect knee joint stability and neuromuscular coordination via electromechanical delay.[188] The results of this study showed that fatigue caused a greater time lag between muscle activation and muscle force production in the hamstrings of both male and female participants, suggesting that fatigue should be considered a risk factor for both sexes. Electromechanical delay was also found to be greater in younger individuals when compared to pubescent and post-pubescent athletes. These results suggest that greater electromechanical delay may compromise the neuromuscular control in muscles necessary for knee joint stabilization, causing excess loading in all three planes of the knee joint.[188] Additional research has shown that exercise causing neuromuscular fatigue is likely to affect knee joint stabilization at initial foot contact during ballistic athletic maneuvers. Results from this particular study showed greater anterior tibial translation of the knee joint during these maneuvers, which would increase load on the ACL.[189]

Knee joint laxity has also been linked with fatigue in the literature. While it is widely accepted that knee joint laxity increases with exercise, a study by Shultz et al. (2015) was the first to show how changes in knee joint laxity could be related to high risk landing biomechanics during prolonged fatiguing exercise.[190] The study used a 90-minute intermittent exercise protocol to simulate the physiological and biomechanical demands of a soccer match on 30 male and 29 female competitive soccer players. It was reported that fatigue had a more global effect on females than males, which resulted in more upright landing motions.[190] As anterior-posterior knee joint laxity increased, females showed an increase in knee internal rotation, while males showed increases in energy absorption and knee extensor loading. Females who exhibited medial-lateral and internal-external rotation laxity, demonstrated greater knee adduction and dorsiflexion. These results suggested that more energy was being absorbed by the knee joint and therefore may be putting the knee at greater risk of injury.[190]

Hormonal

The present study aims to make comparisons between young and old athletes and across sexes. Thus, the understanding of the menstrual cycle and the cyclic fluctuations of hormones during this cycle are vital to the understanding of risk factors that may predispose females to higher ACL injury incidence. Sex hormones underlie many of the sex-specific characteristics that emerge during puberty and also influence the sex-bias in ACL injury prevalence. To understand the basis of how sex hormones may influence ACL injury rate, we must first understand the hormonal changes at each phase of menses. The menstrual cycle can be divided into three distinct phases based on a mean cycle of 28 days. The follicular phase (days 1-9) is generally associated with low levels of progesterone and estrogen until the late follicular phase, at which time estrogen levels rise dramatically. Estrogen concentration continues to rise and peaks during the ovulatory phase (days 10-14) of the cycle. Finally, during the luteal phase (days 15-28), the concentration of both relaxin and progesterone are increased while concentrations of estrogen decrease.

A systematic review of the literature conceded that the cyclic nature of the menstrual cycle could play a significant role in ACL injury risk. Due to the variation in methodologies used within the literature, many of the reported results are largely controversial and contradictory. Hewett et al. (2006) stated that a major limitation within this research is the term "ovulatory phase," since ovulation is a single event and not a phase at all.[43] Thus, this group postulated that if the menstrual cycle were classified more simply as pre-ovulatory or post-ovulatory, that results within the literature would be more consistent. This terminology has been used elsewhere in the literature and has produced results consistent with the articles which are to be presented in this literature review.[43]

Early research examining a link between the menstrual cycle and traumatic injury reported that a higher incidence of injuries occurred during the luteal phase.[55,56,191] These results could suggest that increased levels of relaxin or decreased levels of estrogen are linked to ACL injury since estrogen concentrations are relatively low during this phase.

Alternatively, it could suggest there is no link between estrogen levels and ACL injury risk. However, more recent research tends to disagree with this analysis. Five of seven commonly cited pieces of literature support an increase in noncontact ACL injuries during either the late follicular phase or early ovulatory phase.[29,57,192–194] These are intriguing findings in that they display a distinguishable difference between the pre-ovulatory and post-ovulatory halves of the menstrual cycle. These results would also suggest a link between increased levels of estrogen and ACL injury risk, which has been generally hypothesized among researchers.

A 2007 systematic review by Hewett and colleagues, sided with the majority in that ACL injury risk is increased during the pre-ovulatory phase when estrogen levels are increased.[195] Hewett and colleagues do present a long list of limitations when discussing each of these articles individually, but more importantly they state a need for what they term "a common denominator" that will allow for consistent analysis between all studies.[195] This would include regulating the type of contraceptives ingested, the regimen of contraceptive use, injury exposure and exposure per phase, as well as using consistent language when defining the phases of the menstrual cycle.[43,195]

Despite research describing the link between menstrual phase, hormonal fluctuations and ACL injury incidence, less is known in regard to the mechanism in which the hormones affect the ACL. Cyclic fluctuations of estrogen, progesterone and relaxin in females continues to be an active area of ACL injury risk research, as it may contribute to the susceptibility of female athletes to ligamentous injury. Aside from having a direct effect on the structural properties of the ACL itself, [196–199] sex hormones may also account for varying ACL injury risk throughout the menstrual cycle by affecting the neuromuscular system. Studies dating back to the 1970s have identified a hormonal influence on the neuromuscular system.[200,201] More recently, the specific effects of estrogen have been demonstrated on the female neuromuscular system during handgrip strength and quadriceps maximum voluntary isometric contraction (MVIC) tasks. Sarwar et al. (1996) demonstrated changes in the function of skeletal muscles during the ovulatory phase of the menstrual cycle. More specifically their results indicated that females not taking oral contraceptives showed increased muscle strength, slower muscle relaxation and increased muscle fatiguability during the ovulatory phase of the cycle when estrogen concentrations are greatest.[202] The inability to contract and relax the quadriceps rapidly may suggest a protective role of muscles surrounding the knee joint during the ovulatory phase.

More recently, relaxin has been shown to alter the structural integrity of the ACL and increase injury risk. Relaxin, an insulin-family peptide, is secreted by the ovaries during the luteal phase of the menstrual cycle. Though the role of human relaxin is relatively unknown, relaxin has been shown to be responsible for physiological changes of interpubic ligament in both rat and guinea pig models, suggesting relaxin can alter the ligamentous structures in the body.[197,203] Relaxin receptors have been localized to the human ACL in females only [197] and expression of these receptors have been shown to be controlled by estrogen.[204] As such, relaxin levels peak in the luteal phase of the menstrual cycle. If the presence of relaxin is detrimental to ACL structure, then these findings support results from Wojtys et al. (2002) who found that ACL injury risk is highest during the luteal phase of the menstrual cycle.[56]

The association between varying hormone levels and ACL injury risk in females raises an interesting question surrounding oral contraception. Contraceptives work by inhibiting the production of follicle-stimulating and luteinizing hormones by introducing exogenous estrogen and progesterone to the user. Research surrounding the use of oral contraceptives is equivocally controversial. Moller-Nielson and Hammar (1989) were the first to suggest an association between oral contraceptives and traumatic sport injuries when their investigation revealed that fewer traumatic injuries occurred in those taking oral contraceptives.[55] These results have been reported in several more recent studies,[43,56,205,206] however, the mechanism in which oral contraceptives work to decrease ACL injury risk is still widely unknown. Anecdotal evidence based on unpublished research by Hewett (2000) showed that athletes taking oral contraceptives demonstrated lower impact forces and reduced torques at the knee, increased hamstring to quadriceps strength ratios, and decreased knee laxity as compared to those not taking oral contraceptives.[205] Despite the absence of published data, this review and claim by Hewett has been cited elsewhere in the literature to support findings that suggest oral contraceptives reduce the incidence of ACL injury.[43,56,206] However, substantially more research in this field is needed to validate the effect of oral contraception and the potential mechanisms by which oral contraceptives work to mitigate ACL injury risk. It should also be noted that the dose and ratio of hormones varies widely between oral contraceptives and future studies should look to standardize oral contraceptive use.[142]

2.4.2 Extrinsic Risk Factors

Extrinsic variables are those which occur outside, or external, of the regular anatomical or physiological differences between individuals. These factors have been considered controllable. However, research pertaining to extrinsic variables are relatively understudied compared to intrinsic variables, though they may all contribute to potential ACL injury risk. Extrinsic factors include footwear, playing surface, meteorological conditions and bracing.

Footwear

Shoe-surface interactions have been identified as risk factors in noncontact ACL injuries due to the coefficient of friction that occurs between the two mediums. While increasing friction may be advantageous to increase performance, the trade-off for footwear companies to consider when designing outsoles is that any increased friction may also inadvertently increase ligamentous injury.[38,207,208] These findings were strongly supported in a study by Olsen (2004) which observed female handball athletes and found that higher friction between the shoe and the playing surface increased the likelihood of ACL injury.[103]

Playing Surface

Playing surface has also been highlighted as an extrinsic ACL injury risk factor. Olsen et al. (2004) used longitudinal data over 11 seasons of European team handball and found that ACL injury on wooden floors occurred less frequently compared to those occurring on artificial floors.[103] A total of 53 injuries (9 males and 44 females) occurred over the duration of the study, with a total of 12 injuries (4 males and 8 females) occurring on wooden floors, versus 41 injuries (5 males and 36 females) on artificial floors. The group concluded that the increased incidence on artificial floors was due to a higher frictional force created by the surface. Additionally, a longitudinal study was conducted by the National Football League over four seasons which tracked the surface type, shoe type playing conditions and shoe spatting of each ACL injury. Sixty-one noncontact ACL injuries occurred over the course of the study at a 2:1 rate on natural grass.[209] However, it has also been found that there was a significant reduction in ACL injury risk on natural grass compared to turf.[210]

Meteorological Conditions

The effect of meteorological conditions on ACL injury risk factors is infrequently addressed in the literature. It is possible that the injury risk could be affected by altering the shoe-surface interaction. Orchard et al. (1999) reported that high evaporation rates were associated with greater incidence of noncontact ACL injuries in Australian football players.[211] Anecdotally, a higher rate of evaporation leaves the playing surface feeling drier, thus allowing for more friction and "bite" between the cleat and the surface. Orchard and colleagues later discovered that there was also an increase in ACL injury rates in open stadiums under warm conditions.[210] This is in agreeance with the group's earlier work and with a study published by Torg et al. (1996) which revealed increased pitch temperature can affect the shoe-surface interaction and thereby place the ACL at greater risk of injury.[212]

There is also evidence within the literature that ACL injury risk can be affected by association between temperature and neuromuscular recruitment patterns, where athletes can experience delayed or weaker contraction of the hamstring muscles. In contrast to Orchard and Powell (2003) who showed ACL injuries increase under warm conditions, Csapo et al. (2017) performed a study to show the likeliness of injury to the ACL under cold environmental conditions.[213] Researchers induced peripheral cooling at the knee joint by exposing participants to a cold environment for 30 minutes. Both knee extensor and flexor muscles were examined via electromyography activity, maximum voluntary contraction strength and rate of force development. Results showed the rate of force development was significantly reduced for the knee flexor muscles. The reduced capacity of cold knee flexor muscles to explosively generate force may limit the hamstrings' capability to counter the strong and fast contractions of the knee extensor muscles (quadriceps), which have been shown to produce anterior shear forces on the tibia and increase strain on the ACL.[213]

2.5 Asymmetry

Symmetry of lower limb biomechanics is often assumed in healthy individuals and deficiencies from perfect symmetry are believed to be the product of pathology. However, recent research has shown that limb asymmetry can be developed over time with repetitive use of – and thus a developed preference for – a specific limb to perform a given task.[214,215] Results from Rahnama et al. (2005) showed that knee flexors of the preferred kicking leg were weaker relative to the non-preferred leg in adult soccer athletes.[214] It was hypothesized in this study that during the kicking motion, the non-preferred leg acts as the support leg which actively engages the knee flexors to stabilize the knee joint while also supporting the body weight transfer during the maneuver. Conversely in the preferred kicking leg, flexor activation is minimized to allow the rapid extension of the knee during the kicking motion.[214]

These results compliment previous work by Hirasawa et al. (1979) who pioneered the thought that gait asymmetry can explained by functional task discrepancies.[215] The work by Hirasawa et al. (1979), showed asymmetry in the relative contribution of control and propulsion during healthy able-bodied walking performance.[215] Since then, Sadeghi and colleagues have shown that asymmetry in able-bodied gait can be explained in terms of the actions taken by the lower limbs to propel the body segments and to control forward progression.[97,216] This group suggest that these asymmetries may be the product of specific propulsion and control strategies that are related to each limb. Inherently then, healthy gait seems to be naturally asymmetrical and looks largely related to lateralization.

2.5.1 Asymmetry, Lateralization and the ACL

Lateralization and its relation to the lower limbs has been considered from a support (structural) or operational (functional) point of view. Early explanations of laterality of limbs extended from the idea that our bodies show limb preference for one side versus the other. The development of limb lateralization occurs at varying stages for both the upper and lower limbs.[217] While functional differences between tasks of the lower limbs have been studied less extensively from a motor development lens, the development of laterality has been reported since the 1960s.[217] Belmont and Birch (1963) showed that lateralization of the lower limbs or ‘footedness’ was clearly established by six years of age.[217] A more recent study by Sobera et al. (2011) demonstrated supporting results.[218] This investigation examined the are mapped by centre of pressure and noted that repeatability between left and right foot indices and centre of pressure reflect the development of lower limb laterality.[218] This study also concluded that lower limb lateralization was completed by six years of age.[218]

It is known that the purpose of locomotion is to move the body forward while supporting the body against gravitational forces. From the moment of heel strike and until midstance the body is under a loading phase where body weight (mass * gravity) is being supported by that limb. This phase is considered to be ended at the moment of midstance where the foot is flat and the opposing limb is in the mid swing phase. Beyond that moment of midstance and until toe-off, the body is in a propulsion phase, where it is thrusting the body in the direction of travel. The difference between the support and propulsion phases present a clear task discrepancy during the stance phase of each gait cycle. Asymmetry to human gait may therefore, at least in part, be attributed to functional task differences when examining the gait cycle. Hirasawa (1979) was the first to pioneer the idea that gait asymmetry in able-bodied individuals may be the result of the task association of the support and operational limb.[215] These results were soon supported in the literature by Vanden-Abeele who supported that humans may in fact display some level of footedness, similar to using a preferred hand to write.[219] Later, Sadeghi et al. (1997), defined these functional differences as being power absorbing in the case of loading, or power generating, in the case of propulsion. This group found that the interaction between limbs during able-bodied gait may reflect specific support and operational strategies that are related to each limb.[216]

Most recently, limb dominance has become a popularized method to explain the preferential use of one limb during specific tasks as opposed to limb lateralization. Limb dominance has often been studied among both healthy and injured athletes. Recent evidence shows limb dominance does not influence knee proprioception and single-leg postural control in healthy adults.[220] However, limb dominance has been shown to influence the biomechanics of the lower limbs.[118,119] The limb dominance theory states that there is an imbalance of muscular recruitment patterns and muscular strength between legs, which leads to differences in dynamic control.[118,119] This theory postulates that females have a greater tendency to be more one limb dominant compared to their male counterparts.[118] Research by Brophy et al. (2010) suggested limb dominance serves as an etiological factor regarding ACL rupture, whereby females are more likely to injure their non-dominant limb, and males their dominant limb. Brophy et al. (2010) suggested this discrepancy was a result of underlying sex-based anatomical difference or differences in neuromuscular coordination and biomechanical patterns during cutting maneuvers.[141]

The most prominent weakness of limb dominance and its use within the literature is how it is defined. The dominant limb is also used synonymously with the 'preferred limb.' Early evidence from Peters intuitively defined the preferred or dominant limb as the limb used to manipulate an object.[221] Conversely, this would label the non-dominant or non-preferred limb as the stabilizing limb.[221] Footedness questionnaires have also been used to help assess the self-determined dominant leg.[221–223] To our knowledge, however, only one such study has investigated the relationship of self-reported dominant limb and actual performance.[223] Hart and Gabbard (1998) noted that 98% of individuals correctly identified their dominant limb via responses from in a questionnaire as compared to the limb dominance expressed in various tasks.[223] It should be noted that the tasks performed in this study were done unilaterally from a seated position, and not in a dynamic environment. The definition of limb dominance becomes less intuitive for more dynamic maneuvers where there is bilateral mobilization such as kicking a ball. Is the preferred limb used for striking the ball, or is it used for postural support? According to Peters, in unilateral tasks only one leg is active, thus, the supporting limb would be labelled the dominant limb.[221] However, results from Hart and Gabbard (1998) refute their very own hypothesis that in bilateral tasks the dominant limb is the same limb identified in unilateral tasks. Results from their investigation showed between 44 and 62% of participants switched their standing leg between bilateral tasks indicating limb dominance may vary between tasks for each individual.

By definition however, limb dominance and limb lateralization are largely the same. Both concepts refer to the subconscious preference to use one limb versus the other to perform a specific task. For this investigation, self-reported limb dominance was not considered as a variable, since no distinct definition was prevalent in the literature.

Instead the concept of laterality will be used to discuss the preferential use of one limb versus the other based on task specific roles.

Given that its well understood that lateralization can lead to task-specific roles of the support and operational limb, it is conceivable that – similar to the limb dominance theory – developed bilateral strength differences and/or differences in muscle recruitment may lead to the unconscious preference toward greater loading of one limb versus the other. This in turn may result in asymmetry of lower extremity strength and muscular imbalances and ultimately asymmetry of lower limb biomechanics. When an individual exhibits limb laterality they too express an unconscious preference toward greater loading of one limb versus the other. An over-reliance on the dominant limb, during dynamic athletic maneuvers such as the side-cut, could place the supporting limb – and more specifically the joints of that limb – under greater stress, and thus increased risk of injury. All-the-while, the operational limb, used for propulsion, may not be placed under such stress and may be less prone to injury through decreased exposure to the task. Conversely, the preferential repetition of a side-cut task may account for developmental differences in the refined motor patterns and coordination to absorb the high forces that an individual may experience. Thus, a conflicting theory exists when the operational limb is placed under an instance of increased stress outside of the normal comfortable threshold. In this case, the operational limb may be more prone to injury.

In both cases, lower extremity limb laterality and subsequent asymmetry could lead to excess loading of one limb and could result in an ACL injury in cases where the load exceeds those tolerable by the ligament. Previous studies have identified asymmetry in various lower limb biomechanical measures that may serve as predictors of ACL injury[119,224,225] and re-injury.[226] Asymmetry of the knee valgus moment has been postulated as a risk factor for ACL injury risk in athletes.[224,225] Further research by Pappas et al. (2016) indicated greater asymmetry in hip flexion was associated with increased ACL injury risk.[119] Lastly, Paterno et al. (2010) showed that participants with greater asymmetry of internal knee extension moments were at greater risk of reinjury in a population of participants whom had already undergone ACL reconstruction.[226]

2.5.2 Asymmetry During Healthy Walking, Running and Cutting

Gait symmetry has been defined as perfect agreement between the actions of the lower limbs.[97] Lower limb symmetry is often assumed in biomechanical studies for simplicity in data collection and analysis. With this assumption, it is believed that any deviation from perfect symmetry is the result of pathology. Functional symmetry and muscle strength are commonly used in a clinical setting to define goals of rehabilitation and return to play protocols.[114–116] However, recent studies of healthy populations have shown that asymmetries of both kinematic and kinetic measures exist in the lower-limbs and are likely the result of natural functional differences between limbs.[95,227,228] These results are in agreement with a previous study by Sadeghi et al. (2000) who suggested that one lower limb is responsible for support and body weight transfer during walking, while the contralateral limb contributes to propulsion.[97]

A recent publication by Lathrop-Lambach et al. (2014) showed that 55-69 percent of their studied population experienced greater than 10% asymmetry[95] – the clinically relevant return to play criterion[114–116] – in adduction and flexion moments (external) at the knee and hip during healthy adult human gait. Results of that study compliment the previously presented work of Sadeghi et al. (2000).[97]

Research on lower limb asymmetry has, however, primarily been carried out during walking tasks as asymmetry is more commonly associated with pathology such as knee OA and less with injury prevention. More recently asymmetry has started to be recognized as a potential risk factor for injury in athletics. As such, recent studies have shown asymmetry between limbs increases with speed,[229] indicating potential for larger asymmetries during higher velocity tasks such as running and cutting. Though asymmetry measures during walking may be minimal, these deviations from perfect symmetry may become injurious during higher velocity activities due to increased ground reaction forces and kinetic demands placed on the musculoskeletal system.[230] A study evaluating asymmetry of kinetic and kinematic variables during rested and fatigued running showed that knee internal rotation and knee stiffness became more asymmetrical with fatigue.[227] These findings indicate that fatigue induced changes in gait may also induce change in lower limb asymmetries.

A similar study by Gilgen-Ammann et al. (2017) examined gait symmetry in well-trained runners during interval training sessions incorporating different distances.[231] Using an inertial sensor, this group measured ground contact time of each runner for every foot strike. The ground contact times were compared between the left and right foot strikes and asymmetry was reported as a percentage. It is important to note that some participants in this study did suffer from previous injury. It was reported that average gait asymmetry of all runners was 3.3%.[231] Within the previously injured cohort, asymmetry was found to be significantly greater than the non-injury group, however these asymmetries were only noted in short (400 m) but not at longer distances (600-1000 m).[231] Results from this study indicate that the method of training plays a role in detection of asymmetry. More specifically the results of this study suggest using high intensity runs over relatively short distances.

Finally, Rouissi et al. (2015) examined the time performance of two change of direction tasks at various angles (45, 90, 135 and 180 degrees) in adolescent elite level soccer players. Findings indicated that performance of a side-cut task was significantly greater with the dominant leg versus the non-dominant leg in all cutting directions.[232] Strength measures of the knee extensors/flexors and hip abductors of the dominant leg were also reported as being significantly greater than the non-dominant leg. These results suggest that cutting performance in young soccer players is improved when performing a side-cut off their dominant leg. It was postulated by the authors that this is because of the increase in muscle strength found in the dominant leg. Therefore, to reduce muscle imbalances and improve performance, it was suggested that athletes should promote the use of unilateral lower limb strengthening exercises in addition to bilateral exercises to limit deficits between dominant and non-dominant limbs during a side-cut task. Evidence of bilateral asymmetry was present in joint moments during both side-cut and jump landing tasks for a high school female athlete population.[117] As such, it is unclear if asymmetries and characteristics of limb dominance can be found for pre-pubescent and post-pubescent females, as well as in males of varying pubertal development.

2.5.3 Calculations of Asymmetry

At present there exists four main indices to evaluate asymmetry during human movement: Gait Asymmetry (GA), the Ratio index (RI), the Asymmetry Index (AI), and Symmetry Angle (SA). There exists a fifth asymmetry index, known as the Normalized Symmetry Index (NSI), however, the results from this study were under review and had yet to be published at the time this document was prepared. As such, the NSI will not be discussed in detail.

Despite the relative advantages and disadvantages on the four-primary asymmetry indices, the application of them in biomechanical studies, or clinical settings remains a challenge. Several studies have tested the efficacy of these indices, often noting the similarities between them, but also noting largely different results.[95,233–235] Specifically in the review by Blazkiewicz et al. (2014), a high correlation was noted between each of the four indices, which suggested they may be used interchangeably. However, in the presentation of NSI, authors Queen et al. (2020) note the prior four indices have shortcomings that may limit their utility in specific settings. Common shortcomings include potentially infinite, or unbounded asymmetry, or negative asymmetry. Ultimately both result in a loss of clinical significance.

The Gait Asymmetry index (GA) is a logarithmic transform of a standard ratio between measurements. GA is defined as: $GA = \ln\left(\frac{X1}{X2}\right) \bullet 100$, where 'X1' and 'X2' represents the measurements of both limbs. A measure of 0% would indicate complete symmetry between limbs, while GA >0% would indicate that percentage of asymmetry. A limitation to this calculation is that asymmetry can exceed 100% and is infinite. GA can also be negative where X1<X2. Due to the logarithmic nature of this formula, it is not intuitive how to use it when dealing with negative measurements as is common with gait data. Consider an example of 5 degrees of internal versus 5 degrees of external rotation. In this case it requires exclusion of that variable or using an absolute value which eliminates the clinical significance between the measures of each limb.

The Ratio Index uses the ratio of the values for the two limbs as an index of asymmetry. It is defined as: $RI = \left(1 - \frac{X1}{X2}\right) \bullet 100$, where 'X1' and 'X2' refer to the measurements of the two limbs respectively. Again, RI = 0% would indicate perfect symmetry between limbs, while RI > 0% would indicate asymmetry. >100% asymmetry is possible and is not bounded. A negative value for asymmetry is also possible if X1>X2.

The Symmetry Angle index captures asymmetry as an angular measurement. It is defined as: $SA = \frac{(45° - \arctan\left(\frac{X1}{X2}\right))}{90°} \bullet 100$, where 'X1' and 'X2' refer to the measurements of the two limbs respectively. Again, SA = 0% would indicate perfect symmetry, while SA>0% would indicate asymmetry. Asymmetry can exceed 100%, but this presents the first measure where asymmetry cannot be negative.

The Asymmetry Index is the most commonly used index in the literature and is closely related to the Ratio Index. The AI is defined as: $SI = \frac{(X1 - X2)}{0.5(|X1| + |X2|)} \bullet 100$, where 'X1' and 'X2' refer to the measurements of the two limbs respectively. Again, AI = 0% would indicate perfect symmetry, however asymmetry is again unbounded. Negative asymmetry can be calculated where X2>X1. However, in some cases in the literature, this formula has been adapted where a particular limb is used as reference, in which case this formula is simplified to look identical to the RI.[95] Lathrop-Lambach simplified this formula and defined limbs as greater or lesser depending on the magnitude of the variable.

Each of the methods previously share that a finding of 0% shows perfect agreement between limbs, or 0% asymmetry which occurs when X1=X2. However, each of the methods perform differently as values increase and present no upper bound on asymmetry, which raises the question of what maximal asymmetry is, and how are outrageously large asymmetry values relative clinically when measures from both limbs are within normal limits.

To combat these limitations the present study calculated asymmetry for each variable using the following formula:

$$\%\,Asymmetry = 100\ x\ (1 - \frac{lesser\ value}{greater\ value})$$

This formula, adapted from a previously defined limb Asymmetry Index,[95,236] indicates the relative difference between limbs for each variable. In this method, we ignore, the role of limb dominance but acknowledge there is potential for limb lateralization. In our calculation, if the variables are equal between limbs, it would result in zero asymmetry. Likewise, if the greater moment is twice that of the lesser moment there will be 50% asymmetry. However, by definition of our limb asymmetry index, a situation where the greater value (in terms of absolute value) was negative, it would force this larger negative value to the numerator regardless of its magnitude since any negative number is "lesser" than any positive number. Therefore, an absolute value was taken to eliminate negative asymmetry values.

Using absolute values presents an equally unique challenge, however, as it ignores the clinical difference in direction of joint kinematics or kinetics and treats flexion and extension as equal, for example. Our team concluded that this study would use a hybrid calculation that incorporated taking an absolute value to determine the rank of the numerator and denominator. However, despite the rank, the negative would not actually be removed from the calculation. Consider an example where the recorded knee adduction moments during a side-cut task were 0.1 and -0.2 respectively. In this case, an absolute value will only be used to determine the rank of the variable in question but will not be removed from the calculation itself. This would result in asymmetry ranging between 0-200% and providing some bound or measure of maximal asymmetry. This method would maintain clinical significant, as asymmetry values ranging from 0-100% asymmetry would indicate the direction of the knee adduction moments used as an example, in this case, were in the same direction, whilst, values of 100-200% asymmetry would indicate the direction of the knee adduction moment between limbs were in opposing direction.

2.5.4 A Novel Approach for Capturing Asymmetry in an Athletic Population.

Asymmetry has been measured in many ways through the literature when defining asymmetry of biomechanical data. Each method has its own merit, but are largely faced with the same issues of being unbounded and the ability to measure negative asymmetry. A novel approach will be adapted from a previously defined limb Asymmetry Index for the purpose of this study.

All the studies presented in this review of the literature have focused on only one age group or use a population of only one sex. Thus, there is a gap in the literature which looks at bilateral asymmetries in both male and female athletes at different ages (e.g. pre-pubescent and post-pubescent). A study, such as the one being presented, will be the first study to look for asymmetries in athletes of varying sex and age during a simple walking task, as well as in more sport specific tasks such as running, and the side-cut.

In addition, prevalence and magnitude of asymmetry was noted to be affected by laterality,[232] fatigue,[227] and speed of activity,[229] due to an increase in ground reaction forces and increased kinetic demands on the musculoskeletal system.[230] These findings are in agreement with ACL injury risk factors previously presented in this review of the literature indicating that asymmetry may, in itself, be a risk factor for ACL injury, and may be an explanation for the one-off nature of ACL injury.

2.6 Cutting Maneuvers

Understanding high-risk movement patterns is key to identifying mechanisms and the underlying risk factors of ACL injury. As previously discussed, the most common body position precluding a noncontact ACL injury involves a combination of dynamic knee valgus, rotation in the transverse plane and/or hyperextension that, in combination, results in unendurable anterior shear force on the ACL. These injurious biomechanics usually occur in the absence of contact during the deceleration phase of high-risk sport-specific maneuvers such as landing from a jump or cutting to change direction.

Cutting maneuvers are used to change direction in multidirectional sports such as basketball, soccer, football and rugby. Two separate cutting techniques have been defined in the literature and are known as the cross-cut and the side-cut.[237] The cross-cut maneuver is performed by rotating the torso and pelvis externally on the femur in the same direction of the plant foot, causing the cutting stride of the athlete to be across their body. For example, in a leftward crosscut, the pelvis and torso would rotate externally over the femur upon or just prior to initial contact of the left foot strike. The right foot would then cross the body and provide acceleration in the new direction, completing the cross-cut.[237] Conversely, the side-cut maneuver is performed by rotating the pelvis and torso away from the plant foot at or just following initial contact. To complete a leftward side-cut, the athlete would rotate the pelvis and torso internally over the femur upon or just prior to initial contact of the right foot. The left foot would then bound outward in the leftward direction opposite of the planted foot and provide acceleration in the new direction.[237]

Andrews et al. (1977) divided the cutting phase into three phases: i) preliminary deceleration, ii) plant and cut, and iii) takeoff. The aptly named preliminary deceleration phase refers to the phase where the player decreases their momentum as they head into the cut. The neuromuscular coordination of the quadriceps, hamstrings and gastrocnemii is important during this phase to stabilize the knee joint and provide the muscular power required to decelerate. The plant and cut phase refers to the point where the athlete experiences a change in momentum as they use the hip musculature to rotate in the transverse plane, turning the torso and pelvis toward the direction of the cut. Following contact of the plant leg, the momentum of the free leg allows the athlete to accelerate in the new direction. The takeoff phase refers to the push off of the plant leg, causing further acceleration in the new direction. This is accompanied by extension of the hip, knee and ankle joints, as well as a forward lean of the athlete to aid in acceleration in the new direction. Intuitively, muscle activation patterns differ between the two cutting techniques. Research comparing neuromuscular activation patterns have shown selective activation of the medial and lateral hamstring and gastrocnemii between side-cut and cross-cut tasks,[181] suggesting both muscle groups play a role in control of dynamic stability of the knee joint.

Retrospective video analysis by Olsen et al. (2004) described the noncontact ACL injury mechanisms in female team handball.[103] Of the twenty retrospectively observed ACL injuries, 60% occurred during a cutting maneuver. Olsen et al. (2004) added that in every case, this mechanism was associated with forceful valgus and internal-external rotation of the tibia with the knee near full extension. This research is in support of retrospective analyses by Boden et al. (2000).[2] This group reported the most common kinematic positions related to ACL injury during competitive play occurred while the knee was near full extension and the center of mass was outside the base of support.[2] Thus there is mounting evidence to suggest the side-cut may be the more hazardous of the two cut techniques since execution of the side-cut satisfies both near full extension of the knee and extension of the cutting limb away from the center of mass.

The side-cut maneuver involves a lateral foot plant to generate a medially directed ground reaction force that propels the athlete opposite of the plant leg. However, the ground reaction force can act lateral to the knee joint during this technique, creating potentially hazardous adduction moments at the knee. A study in 2007 noted that more lateral landing patterns resulted in increased peak knee adduction moments during 45 degree cuts in a male cohort.[238] Results from Sigward and Powers (2007) support the notion that lateral foot placement alters lower limb biomechanics. They demonstrated that a more lateral foot placement of the cutting limb (as noted by greater initial hip abduction angles) caused an increase in peak knee adduction moments in female soccer players performing a side-cut.[239] In combination, these studies suggest the most common noncontact mechanisms of injury occur with sudden deceleration in combination with a change of direction where the base of support is away from the canter of mass. Laboratory studies have extensively investigated side-cut techniques in isolation and results of these studies highlight the importance of continued research on the kinematics and kinetics of side-cut maneuvers.

2.7 Unanticipated Cutting Maneuvers

ACL injury rates in competition vastly exceed those in a practice setting. Scranton et al. (1997) investigated the rate of incidence of ACL tears in the National Football League and this report discovered that while only 47.5% of ACL injuries occurred during competition, athlete exposure to practice versus competition was nearly 5:1.[209] These results beg the question as to why ACL injury rates are so much higher during competition. One possible answer for this is the unanticipated nature of competition. Studies focusing on ACL risk factors have primarily been conducted in controlled laboratory settings. Unanticipated cutting has been incorporated into more recent studies in an attempt to better replicate sporting maneuvers that place the ACL at high risk of injury.[5,71–73,86,87,175,240] However, there remains a lack of sex and age comparative studies addressing at risk knee joint kinematics and kinetics during unanticipated cutting maneuvers.

Besier and colleagues analyzed how external loads at the knee joint[86] and neuromuscular activation strategies[175] differed between unanticipated side-cuts, anticipated side-cuts and running for 11 healthy male participants. These studies used light emitting diodes to randomly cue the participants to perform one of four maneuvers: i) straight run, ii) 30 degree side-cut, iii) 60 degree side-cut, and iv) 30 degree cross-cut. Participants also performed the tasks barefoot and at a pre-determined speed. Results from these studies demonstrated that flexion-extension moments were independent of cutting task, however, increases (of up to two times) were found in both varus-valgus and internal-external rotation moments during the unanticipated task versus the anticipated task.[86] The later publication found differences in neuromuscular activation comparable between cut styles and suggested that there was selective activation of the muscle groups during the anticipated condition. For the unanticipated maneuvers, muscles generally had 10-20% greater muscle activation and demonstrated generalized co-contraction strategies compared to the anticipated tasks among their male cohort.[175] An unanticipated cutting study by Pollard et al. (2004) showed that kinematics and kinetics of the knee and hip joint were not significantly different between sexes, with the exceptions of frontal plane hip motion for post-pubescent athletes. However, this study did not compare dominant and non-dominant limbs and did not make comparisons between athletes at different age groups. Together these findings suggest that the reduction of reaction time during the unanticipated task limits the athlete’s ability to make the appropriate postural adjustments that would be seen during a planned or anticipated task. The postural deficits thereby lead to greater lateral foot displacement, an increase in external adduction moments and increased activation of lower-limb musculature to stabilize the knee joint during these ballistic maneuvers.

To this authors knowledge, there exists only one published study investigated bilateral asymmetry of lower limb biomechanics during an unanticipated side-cut task.[241] While asymmetry was not compared directly, Greska et al. (2016) compared the biomechanics of dominant and non-dominant limbs during the unanticipated side-cut maneuver. This group used a MANOVA to determine the effect that limb dominance has on hip and knee mechanics for pre-contact, initial contact, peak knee adduction moment and peak ground reaction force periods. Using this analysis, results from twenty female soccer players showed no significant differences in any hip or knee biomechanics between dominant and non-dominant limbs.[241] However, the authors made note that the small cohort of athletes may not have been sufficient enough for statistical power, and recommend that future investigation be done with a larger sample size.

2.8 ACL Injury Prevention

The high prevalence of ACL injuries equates to $2 billion dollars in health care costs in the USA alone.[24,25] ACL injury prevention strategies pose as a pragmatic approach to reduce the incidence of primary ACL injuries, subsequently reduce health-care costs and further protect the overall health of athletes in reducing the incidence of re-injury and early onset osteoarthritis. To reduce rates of injury, movement assessment screenings to identify at-risk individuals and injury prevention programs have been proposed as solutions.

2.8.1 Injury Prevention Programs

Injury prevention programs aim to reduce ACL injury incidence through modifying aforementioned neuromuscular and biomechanical risk factors. There is significant support in the literature indicating that these programs are effective in provoking advantageous neuromuscular and biomechanical changes that help to reduce the incidence of ACL injuries.[242–247] It has been suggested in the literature that these programs have the capability to reduce ACL injuries by as much as 50% in all athletes, including up to 67% reduction in non-contact ACL injuries for females.[248] Recommendations include training that incorporates plyometrics, strengthening as well as decision making in unexpected situations, with focus on appropriate foot positioning to reinforce neuromuscular coordination, proprioception, and muscle activation.[142,248,249] A particular study on female basketball players consisted of a prevention program focused on preventing a ‘pivot and cut’ movement when changing direction, replacing it instead with an accelerated rounded turn to change direction. The rounded turn was to be executed with a flexed knee instead of being hyperextended and a three-step stop instead of a one-step stop. The purpose of this intervention was to reduce the quadriceps-cruciate interaction which loads the ACL with excessive forces during these key movements.[38] The training included a teaching tape demonstrating mechanisms of noncontact ACL injuries, as well as demonstrations of recommended drills specifically for young athletes to use in practice to produce skill modifications. Similarly, a study on the effect of softer jump landings and one versus two leg landings during stop jumps and side-cut maneuvers indicated that conscious efforts to land on the balls of both feet with greater knee flexion reduced ACL loading by limiting anterior tibial translation and ground reaction forces. Unfortunately, these mechanics were also shown to decrease performance through increased stance time, decreased jump height and decreased movement speed, thereby making it more difficult to implement these changes in a game competition where achieving maximum performance is of the utmost importance.[250] Other studies have also highlighted the importance of focusing on the musculature away from the knee, for example, strengthening at the hip and core to prevent injuries such as noncontact ACL injury.[38,172,247,251]

Notably, the FIFA 11+ injury prevention program has shown to be quite effective in significantly reducing injury, including ACL sprains of both male and female athletes of different ages and levels of competition.[252] The program, developed by F-MARC (FIFA Medical Assessment and Research Center) in 2003 and implemented in European and North American regions starting in 2004, consists of a complete warm up program aimed initially toward male and female soccer athletes greater than 14 years of age. The three-part warm-up first includes low speed running exercises in combination with active stretching and controlled contacts with a partner. The second part includes six exercises that incorporate strength, balance, and jumping exercises, all of which have three levels of increasing difficulty. The last stage consists of speed running combined with soccer specific movements with sudden changes in direction. The aim of the program in the beginning stages of implementation was to improve neuromuscular control and bodily awareness during standing, running, planting, cutting, jumping and landing maneuvers. Players were encouraged to concentrate specifically on core stability, hip control, and knee alignment to prevent excessive valgus motion during movement.[252] A 2008 randomized control trial by Soligard et al. (2008) implemented the FIFA 11+ warm up program as an injury prevention strategy in young female soccer players. The study consisted of 1892 female players between the ages of 13 and 17 years. 301 of the 1892 players sustained 376 injuries in total. Of that, 161 injuries were found in the intervention group and 215 in the control group. Of all injuries, 80% were classified as acute and 20% as overuse injuries.[253] Most notably, the incidence of knee injury in the intervention group was significantly lower compared to that of the control group, suggesting a major decrease in knee injury risk after the warm-up program had been implemented. These findings are significant as they suggest that the structured warm up can effectively reduce the risk of injury by as much as 33% and reduce severe injury as much as 50%. Warm up drills that involve the inclusion of instruction focused on proper neuromuscular control and movements specific to soccer are likely to result in a dramatic decrease in injury rate.[253]

Barriers to injury prevention program success or effectiveness include athlete and coaching staff compliance and specificity of the training program. One study in particular indicated that a reduction in ACL injuries in young female athletes was directly related to athlete compliance and completion.[254] Furthermore, supervision and emphasis on proper form is essential to the success of an injury prevention program.[254]

2.8.2 Screening

Field-based movement screening tests have potential to be a valuable and cost-effective way to identify 'at-risk' individuals by revealing deficits in movement coordination that if not addressed, may lead to an ACL injury. These tests would be advantageous due to minimal equipment, expertise, time and analysis involved, compared to a lab-based analysis.[255] However, in a cost effective analysis by Swart et al. (2014), injury screening was found not to be cost-effect due to low reported sensitivity and specificity.[256] In fact, it has been stated that injury prevention programs are more beneficial in reducing ACL injuries within a team as a whole than identifying individual athletes who are at risk through screening tests. More research is needed at this time to identify and develop tests that not only associate movements with injury risk, but that are also able to accurately predict injuries specific to different sports.[255–257]

2.9 Conclusion and Summary

Injury to the ACL is one of the most devastating injuries in sport, frequently requiring reconstructive surgery accompanied by secondary consequences such as missed playing time, and early onset of knee osteoarthritis.[13,14] Astonishingly, 70-80% of ACL injuries occur in the absence of contact and are usually the result of a combination of several factors when performing high-risk sporting maneuvers such as sudden deceleration, abrupt changes in direction and jump landing.[1–7] These maneuvers are shown to increased mechanical load on the ACL by increasing the amount of anterior shear force on the knee. This can force the lower-limb into a high risk injury position called the 'position of no return,' which is characterized by internal rotation at the hip, along with hip adduction, knee valgus, and external tibial rotation.[8] The most perplexing aspect of noncontact ACL injuries is the one-off nature of the injury, where the athlete sustains the injury during a maneuver they have performed safely countless times over their career.[9] Unfortunately for females, research shows that injury rates are sexually dimorphic in that females experience a higher injury incidence than their male counterparts. Prior to puberty, injury rates for males and females are negligible, however, following puberty, the literature has shown that females exhibit a 2-8 times greater likelihood of sustaining a noncontact ACL injury.[2,26–32,34,35]

The biomechanical mechanisms that contribute to increases in ACL load and injury risk are not well understood. It is widely accepted that ACL injuries occur through a mechanism that loads the knee joint across multiple planes.[82–85] However, a review of the literature demonstrates there are a multitude of both extrinsic and intrinsic risk factors that may put the knee at increased risk of injury by causing excessive knee joint loads. More specifically, biomechanical deficits of kinematic (joint angles) and kinetic (joint moments) variables have been shown to increase dynamic knee valgus, in addition to increase anterior shear forces and loading within the knee joint .[92,107,108] Intrinsically, females exhibit anatomical, biomechanical, neuromuscular and hormonal differences post-puberty that may help explain the increased incidence of injury in females.

Research by Brophy et al. (2010) suggested limb dominance may serve as an etiological factor regarding ACL rupture, whereby females are more likely to injure their non-dominant limb, and males their dominant limb. These results suggest that side-to-side differences exist between sexes, however, symmetry of lower limb biomechanics is often assumed in healthy individuals. Additionally, current return to play protocols use functional symmetry and symmetrical measures of strength as goals in rehabilitation for injured athletes. However, more recent findings from the literature demonstrate that greater than 10% asymmetry in joint mechanics is present in a large proportion of a healthy population during walking.[95] Furthermore, the literature has suggested that small biomechanical asymmetries of the lower-limb may become hazardous during activities with an increased velocity due to the increased ground reaction force associated with an increase in velocity.[229]

Retrospective video analysis has shown that 60% of noncontact ACL injuries occur during a cutting maneuver whereby the athlete exhibited a forceful valgus and internal-external rotation of the tibia with the knee near full extension.[103] Thus, cutting maneuvers have been widely examined in controlled laboratory studies to replicate the high-risk kinematics associated with game-like scenarios. More specifically the side-cut maneuver has been extensively studied since it appears the most hazardous of the two main cutting techniques (e.g. cross-cut versus side-cut). The side-cut maneuver involves a lateral foot plant to generate a medially directed ground reaction force that propels the athlete opposite of the plant leg. Posturally, this places the center of mass outside of the base of support and puts the knee near full extension, which has been identified as hazardous in a retrospective video analysis.[2] The ground reaction force could then act lateral to the knee joint during this technique and create potentially hazardous abduction of the knee joint. Results have shown increased knee abduction, or dynamic valgus, with greater lateral stride during a side-cut.[238]

There exists evidence to suggest ACL injuries occur five times more frequently during competition than in practice.[209] One explanation for this disparity is the unanticipated nature of competitions. Unanticipated cutting has been incorporated more frequently into studies in an attempt to better replicate sporting maneuvers that place the ACL at high risk of injury.[5,71–73,86,87,175,240] Results from the literature have shown altered neuromuscular activation strategies[175], and increases in both varus-valgus and internal-external rotation moments during the unanticipated task versus the anticipated task.[86] It has also been found that unanticipated cutting maneuvers generally produce 10-20 percent greater muscle activation and generalized co-contraction strategies relative to anticipated tasks, suggesting that a reduction in reaction time may be limiting the athlete's ability to make the necessary postural adjustments to safely stabilize the knee joint.[175]

Though our understanding of ACL injuries mechanisms and risk factors have increased significantly, there exists gaping holes within the literature to this date. To this author's knowledge, no studies exist that compare age and sex effects on bilateral symmetry in an athletic population. This will be the first study to examine these effects for a walking, running and unanticipated side-cut task. Additionally, there exists only one published study investigating bilateral symmetry of lower limb biomechanics during an unanticipated side-cut task,[241] however, it excludes the effects of sex and age. Additionally, there are no studies currently that address asymmetries in walking as well as more ballistic sport-specific tasks such as running or cutting. This study aims to address these gaps in the literature and to explore how asymmetry may play a role in the increased injury incidence in post-pubescent female athletes.

Chapter 3 - Methodology

3.1 Recruitment

Participants were recruited via laboratory demonstrations with groups of local high-performance athletes who were members of Acadia Performance Training or via emails distributed through Acadia University's varsity sports teams and local youth basketball and soccer club programs. Participants were scheduled for data collection sessions later through email or telephone conversation with the researcher. A consent form (Appendix A) outlining the purpose, procedure, and the risks and benefits of the research, was given to all athletes and/or their parent(s) or guardian(s) prior to testing.

3.2 Participants and Anthropometric Measurements

The Acadia University and Dalhousie University Research Ethics board approved this study (Appendix A & B). Testing took place in the John MacIntyre motion Laboratory of Applied Biomechanics (mLAB) between June of 2015 and August of 2018 as part of a longitudinal study investigating the change in biomechanical variables through pubertal development from age 8-25 years of age. After receiving written consent from the participant and/or their parent/guardian, the participant was asked to complete an initial participant information form (Appendix C) which included personal information such as phone number, contact email and address. This form was securely filed separately from all other forms for confidentiality. Next, the participant completed an eligibility questionnaire which included information regarding their demographics and sport involvement, as well as a second questionnaire regarding injury history (Appendix D) to determine their eligibility. A third questionnaire related to pubertal status was also filled out at this time to determine their puberty development score as defined by Carskadon & Acebo (1993) as part of the longitudinal study taking place in the laboratory (Appendix E).[258] Participants were advised prior to testing that they may withdraw their data within 30 days post-testing. Participants were also informed that their results were part of a longitudinal study in which they may be asked to participate in further testing in years to follow. Any questions were addressed verbally or through demonstration prior to testing if requested.

High performance court and field athletes (males n = 57, females n = 65) between the ages of 8 and 25 years of age participated in the present study. For this particular study, participants were categorized as pre/early pubescent (8-11 years old), and post-pubescent (17+), to obtain a large group sample sizes to satisfy the power calculations as opposed to using the results of the pubertal questionnaire which spread the subset of participants to thin over several pubertal groups.

Our current sample size was determined using power calculations based on a study by King et al. (2019) who demonstrated differences in internal knee abduction moments between healthy and ACLR participants in a change of direction task.[259] According to our power analysis, a sample size of 18 participants per group was required to produce an 80% chance of obtaining statistical significance at the alpha level of 0.05.

Anthropometrics for each group are summarized in Table 3.1.

Table 3.1. Summary of anthropometric data by sex and age. (Mean ± Standard Deviation)

	Pre-Pubertal		**Post-Pubertal**	
	Male	Female	Male	Female
Sample Size (N)	24	25	33	40
Age (years)	10.8±1.6	11.2±1.3	21.2±2.2	20.3±1.2
Height (cm)	145.7±11.4	148.3±8.2	181.5±6.6	168.5±7.0
Weight (kg)	38.3±9.6	39.1±8.0	83.9±9.1	69.6±11.2
BMI	17.7±2.4	17.6±2.4	25.5±2.1	24.5±3.5

Only high-performance athletes were selected out of a larger subset of participants to be included in the study. Based on a hierarchical approach, high-performance athletes were chosen if they played at the National, Provincial or Triple-A (the highest club level) in their respective court or field sport for one full season at the time of testing. Athletes were excluded if court or field sport was not listed as their primary sport on the eligibility questionnaire (Appendix D). For sports which may require various skill sets and body types at varying positions, such as football, athletes were only selected if they played a skilled position which required frequent and explosive cutting.

Participants also did not meet the inclusion criteria if they reported a history of major trauma or injury to the lower extremities or lower back. Participants who reported an ankle sprain were only included if the injury had been healed for a minimum of three months prior to testing and they had been cleared to fully return to their sport by a trained medical professional.

Participants were asked to bring appropriate indoor footwear that they would use in training for their respective sport. Participants were also asked to wear tight fitting compression shorts, and a tight-fitting compression shirt to accommodate accurate marker tracking and placement on bony landmarks. Compression clothing was provided by the laboratory in instances where the participant did not have their own. Participants' height and weight were measured using an electronic scale (Health-o-meter professional, McCook, IL, USA) with shoes removed. Participants were then asked to remain barefoot with feet shoulder width apart, and weight evenly distributed between their feet while maximum thigh and calf circumferences were measured for both legs using a tape measure. Foot width was measured using a Rosscraft caliper (Campbell, USA) from the head of the first metatarsal to the head of the fifth metatarsal.

3.3 Electromyography Setup and Instrumentation

Although not used for the present study, a wireless 16 channel EMG Delsys Trigno system (Delsys Inc., Natick, MA) was used to collect EMG data from eight lower extremity muscles, bilaterally (2000 Hz, preamp gain 1000 times, bandwidth 20-450 Hz). Wireless surface electrodes were placed on the surface of the skin on muscle bellies of the lateral gastrocnemius, medial gastrocnemius, vastus lateralis, vastus medialis, rectus femoris, lateral hamstring (biceps femoris), medial hamstring (semitendinosus / semimembranosus), and the gluteus medius of both the right and left legs based on previous measurement guidelines used by Landry et al. (2009) and found within the SENIAM online guidelines (accessed at http://www.seniam.org) (Table 3.2).[73] Electrode placement was slightly altered on some participants due to anatomical variability to ensure consistency of the electrode on the muscle belly.

After locating appropriate EMG electrode locations for each muscle, the skin was prepped to ensure quality EMG signals. Preparation involved shaving with a dry razor to remove hair and dead skin before being cleaned with rubbing alcohol swabs and dabbed with 3M medical tape to remove any remaining hair and dead skin that may attenuate or interfere with EMG signal. The 16 surface EMG electrodes were then placed on the prepared muscle bellies, and affixed with double-sided tape in the anatomical direction of the underlying muscle fibers. Signal quality of each surface EMG electrode was verified in a series of validation trials using manual muscle tests where the participant was asked to activate each muscle group by resisting plantarflexion, knee flexion, knee extension, hip extension and hip adduction. Inadequate signals required re-measurement, cleaning and replacement of an EMG electrode until all signals appeared adequate during validation. A bias trial was collected with the participant laying supine in a relaxed position over a one second period to measure the background noise detectable while the muscles were not being voluntarily activated. Surface electrodes were secured to the skin with Coverlet Cover-Roll Stretch Adhesive Bandages to ensure proper contact during movement and to prevent detachment over the duration of the testing period (Figure 3.1). Fabrifoam wraps were then used to cover the electrodes to further minimize movement during data collection (Figure 3.2).

Table 3.2. EMG electrode placement for the eight muscle sites on the lower limb.

Muscle site	Location
Lateral Gastrocnemius	30% of the distance from the knee's lateral joint line to the calcaneus
Medial Gastrocnemius	35% of the distance from the knee's medial joint line to the calcaneus
Vastus Lateralis	33% of the distance from the knee's lateral joint line to the anterior superior iliac spine
Vastus Medialis	20% of the distance from the knee's medial joint line to the anterior superior iliac spine
Rectus Femoris	50% of the distance from the superior border of the patella to the anterior superior iliac spine
Lateral Hamstring	50% of the distance from the knee's lateral joint line to the ischial tuberosity
Medial Hamstring	50% of the distance from the knee's medial joint line to the ischial tuberosity
Gluteus Medius	50% of the distance from the greater trochanter to the iliac crest

3.3.1 Maximum Voluntary Isometric Contractions (MVICs)

Maximum voluntary isometric contractions (MVIC) were performed to obtain strength measures and to normalize EMG muscle activation patterns during movement trials as a percentage of MVIC (% MVIC). Participants completed a total of 24 MVIC trials to obtain maximal EMG activations for each muscle bilaterally. Each MVIC exercise was performed twice, and the participant was instructed to give maximal effort for each trial. Many of the MVICs were performed on the Biodex System 3 dynamometer (Biodex Medical Systems Inc., New York, NY). The lever arm of each attachment was adjusted for each participant and recorded on the data collection sheet (Appendix F). Table 3.2 outlines the five MVIC exercises performed on the Biodex dynamometer, along with the resisted plantarflexion that involved a researcher pushing down on the participant's shoulders. To ensure maximal contractions and prevent the participant from using leverage while performing movements on the Biodex, the participant was instructed to place their hands across their chest or on their lap. Gravity correction trials were performed prior to each group of exercises with the participant at rest. Order of the MVIC trials (e.g. left or right) were randomized between participants such that any learned effect could be minimized within the population.

Figure 3.1. Electrode placement with Cover Roll on participant (Left: Posterior view; Right: Anterior view).

Figure 3.2. Electrodes covered with Fabrifoam wrap, in preparation for passive marker cluster placement.

Table 3.3. Exercises for strength measures and to normalize EMG data (% MVIC).

Muscle	Exercise
Vastus Medialis and Lateralis	Knee Extension (45˚)
Rectus Femoris	Knee Extension (45˚) + Hip Flexion (90˚)
Lateral and Medial Hamstrings	Knee Flexion (45˚)
Medial and Lateral Gastrocnemii	Seated Plantarflexion (0˚)
Medial and Lateral Gastrocnemii *	Standing Plantarflexion*
Gluteus Medius	Hip Abduction (Semi-prone)

* Performed without Biodex.

3.4 Determination of Self-Selected Walking Speed and Maximum Sprint Speed

The investigators took participants from the mLAB up to the indoor walking/running track within Acadia's Andrew H. McCain Arena where average walking speed and maximal effort sprint velocities were determined. Participants warmed up by completing a lap of the indoor track and were then instructed to complete any additional warm up exercises they deemed necessary to compete in competition.

The participants began with four walking trials at their self-selected walking speed. Participants were instructed to walk naturally through two Fusion Sport timing gates (Fusion Sport Inc., Coopers Plains, AUS), set two meters apart at a distance five meters from the participant's starting point. Using the preloaded Phosphate Decrement protocol, the time between gates was measured by the timing gate system. This protocol was repeated until four walking trials had been collected. The average of these trials was used to calculate the participant's self-selected walking speed, which was then used as a control in subsequent walking trials within the mLAB.

Participants also completed two maximum sprint trials in a similar manner along the indoor track. Using the preloaded 2-Gate Sprint protocol, participants started with a 10-meter approach and ran at a maximum speed between the two Fusion Sport timing gates. Participant were instructed to sprint completely through the second gate to encourage maximal effort sprints. Between trials the participant returned to the original start point and was given 30 seconds of rest. An average of the two sprint times determined the participants maximal sprint speed after 10-meters. This result was used to calculate the participant's approach speed for running and side-cut trails, which was 66.6% ± 10% (two-thirds) of their maximal effort sprint speed. This enabled consistent control of the participant's approach speed for each subsequent running and side-cut cutting trial within the mLAB. In addition, a participant's maximum speed was used to control the distance the timing gates were to be adjusted to give each participant an approximate half second of reaction time to respond to the illuminated gates for the unanticipated side-cut trials.

3.5 Motion Analysis Setup and Instrumentation

A full-body marker set of 14 mm retro-reflective markers were affixed to the participant to enable three-dimensional kinematic and kinetic data to be quantified during the motion trials. Individual markers were applied with double sided tape to bony landmarks on the participant's thorax, pelvis, as well as upper and lower limbs. Marker clusters, rigid plastic plates with markers fixed in each corner, were affixed on the shank and thigh over the Fabrifoam wrap using Velcro. Marker clusters were secured to the legs using athletic tape. Triad marker plates were positioned on each heel and secured using

Duct tape. Lastly, a headband containing 5 retro-reflective markers was placed snugly around the head of the participant. In total, 75 markers were placed on the body (Figure 3.3) (Appendix G).

3.5.1 Data Acquisition

Three-dimensional positions of the retro-reflective markers were captured using a 12-camera Qualisys (Qualisys AB, Sweden) motion capture system (Oqus 4) and one high speed video camera (Oqus 210c) at a sampling rate of 250 Hz. The ground reaction force data between the ground and the foot of the participant was captured at 2000 Hz by three AMTI (Advanced Mechanical TechnologyInc., Watertown, MA) strain gauge platforms embedded in the floor of the mLAB in an 'L' shape configuration (Figure 3.4). The Qualisys Tracker Manager (QTM) software (Qualisys AB, Sweden) simultaneously captured and synchronized all EMG data, force platform data, and Biodex Dynomometer torque data at 2000 Hz to the motion capture sampling rate of 250 Hz. The combination of the three-dimensional position data of each marker in conjunction with the force data output allowed for a full body three-dimensional inverse dynamics analysis to be performed in Visual3D biomechanical software (C-Motion Inc., Rockville, MD).

Figure 3.3. Retro-reflective marker placement on participant. Left: Anterior view; Right: Posterior view.

3.5.2 Calibration Trials

Initial calibration of the motion capture system took place prior to the participants arrival. A Qualisys calibration wand and L-shaped calibration frame was used to calibrate the testing volume where the trials took place within the laboratory space (Figure 3.4). The calibrated volume, which was visible within the software, was to contain no visible empty spaces and was to adequately cover the entire area where dynamic movements would occur. An overall volume residual of 1.5 mm or less was required for the calibration to be accepted. The calibration residual value was recorded, and the calibration file was saved to the data collection computer.

Participant calibration included a 5 second standing calibration in which the athlete was instructed to remain in the anatomical position while standing on two adjacent force platforms to establish joint centers, segment parameters and reference angles for the ankle, knee and hip joints. The participant then completed a moving calibration trial where they moved each of their extremities separately, along with rotation of the torso and head. Ten medial markers were then removed from the feet, ankles, knees and elbows, leaving 65 remaining markers to be tracked for the movement trials (Appendix G). The markers that were removed were used as virtual markers during the movement trials. An identical movement calibration was then completed without the virtual markers. Functional hip joint centers where identified using a 45 second trial where the participant placed their hands on their chest and while balancing on one foot, completed five repetitions of hip flexion and extension, adduction and abduction and circumduction for each leg. Participants were instructed to maintain balance and minimize pelvic movement as much as possible during the functional hip joint trials.

3.6 Dynamic Motion Trials

Participants' kinetic and kinematic data for the upper body, hip, knee, ankle and midfoot were recorded for a series of motion trials including walking and several athletic maneuvers: double leg drop jump, single leg drop jump, running and unanticipated side-cut. Each maneuver was demonstrated to the participant before they attempted the maneuver.

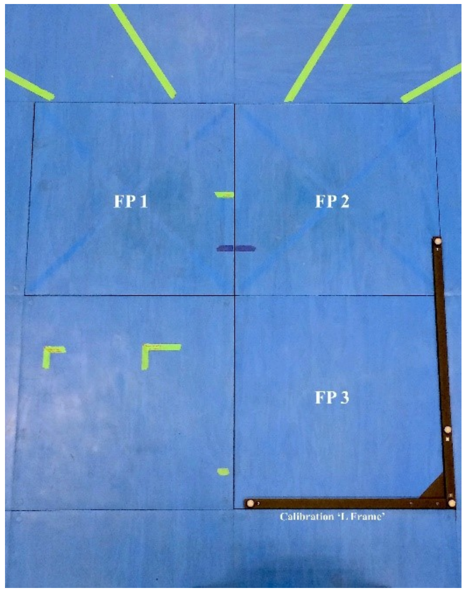

Figure 3.4. 'L-shaped' layout of embedded force platforms in laboratory floor including calibration 'L-frame'.

3.6.1 Double-Leg Drop Jump

Participants were instructed to stand atop a 30 cm box with the medial aspect of their shoes spaced 35 cm apart and toes at the edge of the box.[35,63] The box was centered between two adjacent force platforms at a distance of 15 cm from the edge of the platforms. Participants were instructed to drop directly off the box onto the force platforms such that their feet landed simultaneously and unilaterally on each platform, then immediately perform a maximal vertical jump, again landing unilaterally on each force platform. Participants were instructed specifically to "drop off the box without jumping" and "explode off the plate after the first drop landing".[150] After landing from the maximal vertical jump, participants were told to hold their landing position until the researchers concluded the trial. Trials were unsuccessful if the participant did not drop off the box with bilateral symmetry, if they jumped off the box rather than drop, if they did not land with both feet simultaneously, if full exertion was not demonstrated during the maximal vertical jump or if either jump landing was not entirely on each force platform. Participants were required to complete a total of four successful trials.

3.6.2 Single-Leg Drop Jump

Participants were instructed to stand unilaterally atop a 30 cm box such that their toes were at the foremost edge of the box. The box was placed 15 cm posterior from the edge of an embedded force platform. Participants were told to drop directly off the box onto the force platform such that their non-support leg did not touch the box. Participants were instructed to "drop off the box without jumping" and "explode off the plate after the first drop landing" using just the single leg. After landing from the maximal vertical jump, participants were told to hold their landing position until the researchers concluded the trial.[70] Trials were unsuccessful if the non-support leg contacted the force platform or the box at any point during the trial, if the participant jumped off the box rather than dropped, if the participant did not land within the bounds of the force platform, if full exertion was not demonstrated during the maximal vertical jump or if the participant performed a 'hop step' before executing the maximal vertical jump of the protocol. Participants completed a total of four successful trials on both the left and right limbs.

3.6.3 Walking Trials

Participants began walking at a distance of five meters from the first timing gate and in alignment with force platform 2 (FP2) and force platform 3 (FP3) (Figure 3.5). This allowed for two potential foot strikes on the embedded force platforms per trial. Fusion Sport timing gates were positioned both one meter prior and one meter after the midpoint of FP2 and FP3, creating a two-meter timing window which was used to calculate walking speed. Participants were instructed to walk straight forward at a natural pace, keeping their eyes up until they reached the far wall of the mLAB. Walking speed was controlled to be the participants previously calculated self-selected walking speed ± 10%. Immediate feedback was given via a scoreboard, and through "speed up" or "slow down" commands from the researchers. Four-to-eight successful walking trials were collected in total, including a minimum of four foot strikes of both the left and right limb. Trials were excluded if the participant intentionally targeted the force platforms, had notable altered gait patterns, did not meet the proper speed requirement, or did not have at least one full foot strike on either force platform.

3.6.4 Running Trials

Participants began at a distance of 10-meters from the first timing gate and in alignment with force platform 2 (FP2) and force platform 3 (FP3) (Figure 3.5). Fusion Sport timing gates were positioned both one meter prior and one meter after the midpoint of FP2 and FP3, creating a 2-meter timing window which was used to calculate run speed. Participants were instructed to run through the gates and only begin deceleration after passing through the second gate. Running speed was controlled to be 66.6% ± 10% of each participant's average maximum sprint speed, which was obtained earlier on the indoor track. Immediate feedback on approach speed was given via a scoreboard and through "speed up" or "slow down" commands from the researchers. Eight successful running trials were collected in total, including a minimum of four foot strikes for both the left and right limb. Trials were excluded if the participant intentionally targeted the force platforms, had notable altered gait patterns, did not meet the proper speed requirement, or did not have at least one full foot strike on either force platform.

3.6.5 Side-Cut Trials

Participants began at a distance of 10-meters from the center of the foremost force platforms (FP1/FP2), centrally aligned between FP1 and FP2. The timing gates were configured in a Y-shape configuration with the force platforms located in the middle of the runway (Figure 3.5). The first two timing gates allowed for controlling the participant's approach speed and were positioned at a variable distance from the center of the plates to allow 0.5 seconds of reaction time to cut either left or right. This distance was determined from the average maximum sprint speed calculated previously. Approach speed for the cuts was controlled to be 66.6 ± 10% (two-thirds) of each participant's average maximum effort sprint speed. Immediate feedback of approach speed was given via a scoreboard and through "speed up" or "slow down" commands from the researchers. Participants were instructed to run at the same speed as they had been for their running trials, with eyes focusing on the two timing reaction gates positioned after the force platforms and at the end of the mLAB. Upon breaking the second timing gate, either the left or right timing gate at the end of the mLAB would be randomly queued to activate. Participants were instructed to maintain forward momentum but were encouraged to decelerate or stutter step slightly, if needed, to perform a side-cut in the direction of the randomly queued gate, just as they might in a game-like situation. Participants were also instructed to cut and explosively sprint through the next timing gate such that their opposite foot made complete contact with the appropriate force platform. For leftward cuts, this meant a right footed cut off the right force platform (FP2); for rightward cuts, this meant a left footed cut off the left force platform (FP1). Tape was applied to the floor to guide the cutting angle of 45 ± 15° from the center of the two platforms as previously performed by Landry et al. (2009).[73] Eight successful cut trials were collected with a minimum of four in each direction. Trials were excluded if the cut was not in the proper direction, did not meet the speed requirement, was not performed explosively, or if full stance did not occur on the proper force platform. Trials were also excluded if forward momentum was deemed too slow or stopped, and if the cut was performed at an angle outside the designated range.

3.7 Data Processing

Data were post-processed within the Qualisys QTM software (Qualisys AB, Sweden) and exported to Visual3D (C-Motion Inc., Rockville, Maryland, USA) for further analysis. Both kinematic and kinetic data were filtered using a low-pass fourth-order recursive Butterworth filter and at a cutoff frequency of 12 Hz. This was consistent with previous side-cut studies.[75,260,261] Stance phase was defined for each trial in Visual3D by creating labelled events for initial contact and toe-off by determining the point at which the force platform surpassed a threshold of 10 N, and the point at which the force platform returned to a state under the 10 N threshold, respectively. These events were used to define the following phases for the various movement tasks: pre-contact (200 ms prior to initial contact), stance, post-contact (200 ms after initial contact) and stride (walking only).

Before exporting the kinematic and kinetic data, the stance phase for each trial was normalized to 101 data points. This allowed the joint angle and internal joint moment waveforms to be expressed as a percentage of the entire stance phase (0-100%). Ensemble averages were computed for each variable of interest, for each participant and for each task.

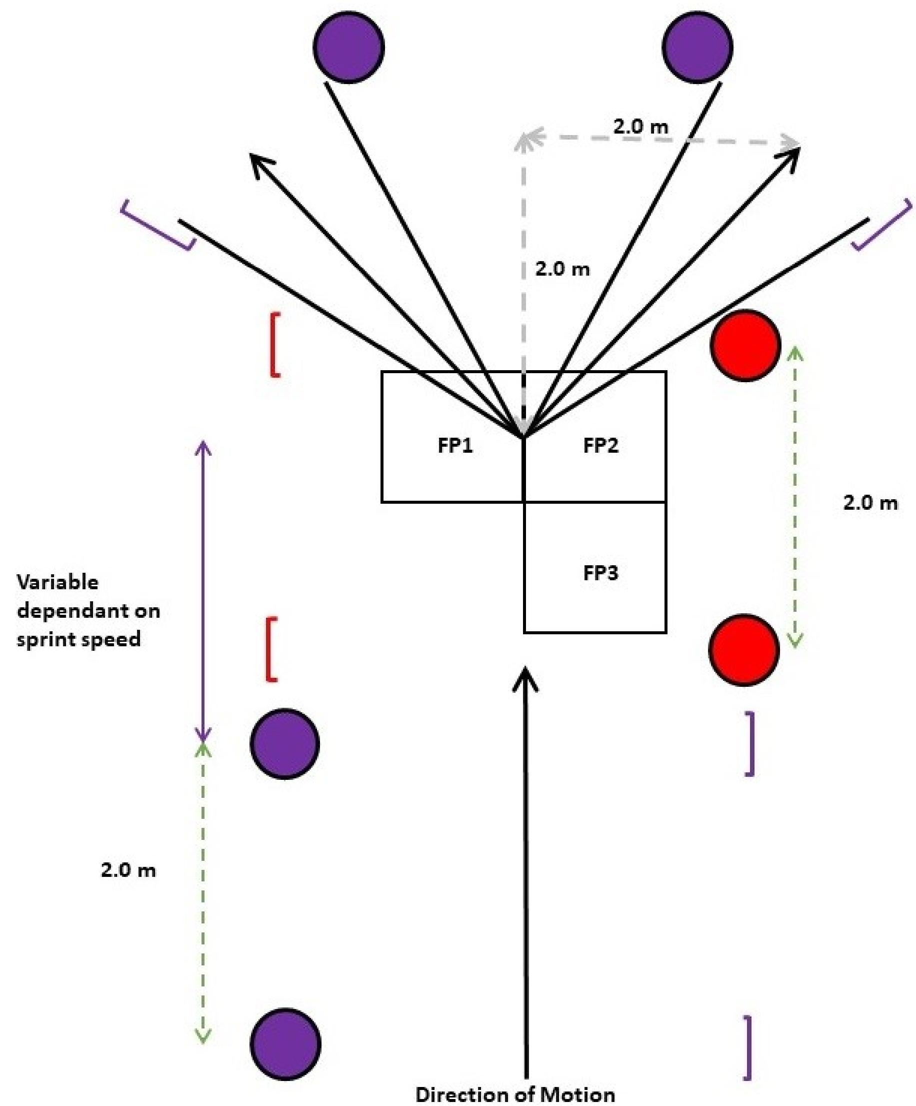

Figure 3.5. Y-shaped configuration of Fusion Sport SmartSpeed timing gates for unanticipated side-cut maneuvers (green) and walking and running motion analysis (purple).

3.7.1 Joint Kinematics

Using Visual3D (C-Motion Inc., Rockville, Maryland, USA), a biomechanical model was constructed by defining the nine lower limb segments (pelvis, thigh, shank, rearfoot, forefoot) as rigid bodies. Anatomical coordinate systems were constructed for the pelvis, thigh, shank, rearfoot and forefoot using retroreflective markers placed on each of the defined segments. The anatomical coordinate systems were used to quantify three-dimensional joint angles for the hip, knee and ankle based on the orientation of the relative proximal and distal segments with respect to one another. An X-Y-Z (flexion/extension, adduction/abduction, internal/external rotation) order of rotation was used to report hip, knee and ankle joint cardan angles on a floating joint coordinate system previously defined by Grood and Suntay.[262] All joint angles were reported in degrees (°) and referenced to a standing calibration trial. For convention, flexion, adduction and internal rotation were all positive joint angles at the hip and knee.

The calculated joint angles were described about the following axes. The flexion-extension axis at the hip was constructed by creating an axis which ran parallel to the line which passed through the right and left anterior superior iliac spines. The rotational axis was created about the distal long axis of the femur and an intermediate and orthogonal adduction-abduction axis was created by taking the cross product of the rotational and flexion-extension axes. The knee joint coordinate system was defined using the same convention. The flexion-extension axis at the knee was a bone embedded axis which passed though the lateral and medial femoral epicondyles. The knee's internal-external rotation axis was defined along the long axis of the shank (tibia) from the lateral femoral epicondyle to the lateral malleolus, while the adduction-abduction axis was defined using the cross product of the two pre-defined axes. The dorsiflexion-plantarflexion axis at the ankle was defined in the shank (tibia) and the eversion-inversion axis was defined about the long axis of the foot. The toe-in (adduction) – toe out (abduction) axis was orthogonal to the two axes mention above for the ankle. Each of these axes were expressed for the hip, knee and ankle joint centers where the calculated joint angles were expressed about these joint coordinate systems. The hip joint center was defined using a previously published regression equation which used the relative distance between the anterior superior iliac spine markers.[263,264] The knee joint center was defined as the midpoint of the two femoral epicondyles and the ankle was, likewise, defined as the midpoint of the two malleoli. All joint angle waveforms were normalized according to the stance phase of each maneuver and represented by 101 data points ranging from 0% (initial contact) to 100% (foot off) in 1% increments

3.7.2 Joint Kinetics

Using both the kinematic and ground reaction force data, inverse dynamics techniques within the Visual3D software were used to calculate three-dimensional internal hip, knee and ankle joint moments for the stance phase of all motion trials (walking, running, cutting and jump-landing). Internal joint moments were calculated by first calculating the joint force generated at the segment most distal to the joint of interest. For the ankle, the resultant force at the ankle required to balance both the ground reaction forces and the effects of gravity and accelerations on the foot we calculated. The resultant internal ankle moment was then calculated based on the moment created by the forces applied to the foot and the moment generated through contact with the ground. After the joint forces and moments at the ankle were computed, they were then used with the inertial properties and kinematics of the shank to calculate the resultant forces and moments about the knee joint. Similarly, hip joint forces and moments were calculated using the resultant joint forces and moments at the knee joint in combination with the inertial properties and kinematics of the thigh. The net internal resultant moments for the hip, knee and ankle were expressed via the same axes used to express or calculate the corresponding joint angles (floating joint coordinate system). The moment waveforms were similarly time normalized to 101 data points ranging from 0% to 100% of stance in 1% increments. Joint moments were normalized to the participant's body mass (N*m/kg). Positive internal moments for the three joints were the same as those described for the joint angles at each joint.

3.7.3 Biomechanical Variables of Study

The biomechanical variables of interest for this study included percent asymmetry for both peak sagittal plane joint angles at the hip and knee as well as angles at initial contact (IC), again, for the hip and knee. Peak internal joint moments in both the sagittal and frontal planes were also examined for this study (Table 3.4). Frontal plane joint angles as well as transverse plane kinematics and kinetics were excluded due to the sensitivity of our asymmetry calculation to small measures. It was also a goal of this study to build on the findings of Lathrop-Lambach et al. (2014).[95] Thus, it was important to include both peak flexion and adduction moments for both the hip and knee. To build on that study, it was thought that the inclusion of peak sagittal plane joint angles and sagittal plane joint angles at initial contact for the hip and knee joints of the lower limb would provide additional insight into the role of asymmetry as an ACL injury risk factor. These variables were not solely selected based on just one previous study, however. Numerous studies have sought to investigate sagittal plane kinematics,[92,102,265–267] and kinetics,[110,268,269] as well as, frontal plane kinetics[79,82,108,110,112,187] of the side-cut maneuvers.

Studies of sagittal plane kinematics have reported that females exhibit greater knee extension at initial contact than their male counterparts.[92,265,270] This is particularly relevant for ACL injury since non-contact injury has been reported to occur in higher frequency when the knee is more extended.[2,103]As such, it is believed that reduced flexion at initial contact may increase the risk of ACL injury for females. In addition, females have been reported to exhibit smaller peak hip extension moments and greater peak knee extension moments relative to males.[110] This is evidence to suggest females and males display different strategies in the execution of the side-cut, and merit additional research into the role sex and age may play on asymmetry of sagittal plane joint kinematics and kinetics.

Likewise, research on frontal plane joint moments have reported that females exhibit greater internal knee abduction moments compared to males.[108,110,187,271] Knee valgus moments have also been shown to increased mechanical load on the ACL, which would potentially place it at higher risk for rupture.[79,82] Additionally, asymmetry of frontal plane joint moments have also been previously linked to ACL injury. Hewett et al. (2005) was the first to demonstrate that asymmetry of the knee valgus moment is a predictor of future ACL injury in female athletes.[150] This study also showed that asymmetry of the external knee flexion moment can be used to predict re-injury among a population of athletes who had previously underwent ACL reconstruction. This provides motive for investigation of frontal plane joint moments in addition to the sagittal plane joint angles and joint moments.

A custom written MATLAB (Mathworks, Natick, Massachusetts, USA) script was applied to the Visual3D exported data (angle and internal moment ensemble averaged waveforms) to isolate the peak and instant of contact variables during the stance phase of the ensemble averaged waveforms for walking, running and side-cut. Values were determined for limbs for each variable of the three movements and used to calculate percentage of asymmetry.

Table 3.4. Biomechanical variables of interest for this study

Joint Moments*	Joint Angles
Hip	***Hip***
Peak Hip Extension Moment	Peak Hip Flexion
Peak Hip Abduction Moment	Hip Flexion at IC
Knee	***Knee***
Peak Knee extension moment	Peak Knee Flexion
Peak Knee Abduction Moment	Knee Flexion at IC

* All moments are reported as internal joint moments

3.7.4 Asymmetry in a Healthy Athletic Population

An additional custom written MATLAB (Mathworks, Natick, Massachusetts, USA) script was applied separately to the data to determine asymmetry in the healthy athletic populations during the walking, running and side-cut tasks. An asymmetry measure was calculated for each variable and for each task using the following formula:

$$\%\,Asymmetry = 100\ x\ \left(1 - \frac{lesser\ value}{greater\ value}\right)$$

This formula, adapted from a previously defined limb asymmetry index,[95,236] indicates the relative difference between limbs for each variable. Using this method, the greater value is forced as the denominator to avoid any misrepresentation of asymmetry within the population by averaging what may otherwise be positive and negative values. Descriptive statistics of the percentage of asymmetry (% asymmetry) were calculated for each variable of interest in each age and sex group.

Based on a previously clinically accepted level of symmetry, 10% was chosen as a threshold for determining asymmetrical joint angles and moments for this study. This value is in accordance with other clinically relevant differences in muscle strength and performance based testing.[20,116,236] This method has only once been used to define symmetry of joint moments during a walking task in a healthy population but has yet to be used to define asymmetry in an athletic cohort.[95] In addition, this study is the first to use this clinically relevant definition of asymmetry for sport-specific ballistic athletic maneuvers such as running and a side-cut.

3.8 Statistical Analysis

Percent Asymmetry for the biomechanical variables of interest (Table 3.4) were submitted to statistical treatment using SPSS (version 22.0; SPSS Inc, Chicago IL).

3.8.1 Effects of Age and Sex on Asymmetry in an Athletic Population

A Chi-Square test with alpha of 0.05 was used to find the proportion of subjects with greater than 10% asymmetry across all age and sex combinations (pre-pubertal male, pre-pubertal female, post-pubertal male, post-pubertal female) and to detect if there were differences in the proportion of subjects with greater than 10% asymmetry between groups for each kinematic and kinetic variable of interest.

Differences in mean % asymmetry between the four populations across age and sex were tested using a 2-way factorial analysis of variance (ANOVA). The ANOVA was performed using the % asymmetry data to assess the interaction and main effects of the independent variables, age and sex. The level of significance (alpha) was established at 0.05. Tukey pairwise correction tests were performed post hoc to establish specific group differences after a significant 2-way interaction to determine where those differences lay.

The assumptions for outliers, normality and homogeneity of variance were tested individually prior to the completion of the ANOVA (Appendix K). Outliers for each population were determined for each variable using box and whisker plots. Outliers remained unchanged and were kept in the data set as these measurements for % asymmetry were deemed to be real data points and were not artifacts of the protocol. Removal or adjustment of these outliers was also deemed unethical by the research team and could not be justified. Normality was tested using a Shapiro-Wilk test (Appendix K). Data was considered non-normal if results of this test were less than the established p-value ($p<0.05$). Normality is often sensitive to the presence of outliers; thus it was expected that data would be non-normal for study populations which presented outliers. This expectation was validated with results of the Shapiro-Wilk test (Appendix K). Percent asymmetry was largely non-normal across all groups and for all variables (Appendix K). Though the ANOVA is generally a robust analysis, and not particularly sensitive to outliers, and violations of the normality assumption, a transformation of $\sqrt{X}$ was applied to non-normal data sets. The transformed data set met the assumption of normality for 93 of 96 variables included in the analysis (Appendix K). Full transformed and untransformed data sets were then subject to Levene's Test of homogeneity to test the final assumption of homogeneity of variance (Appendix K). The assumption of homogeneity was violated if $p<0.05$. Untransformed data did not meet the assumption of homogeneity of variance. The transformed data set did, however, meet the assumption of homogeneity across 21 of 24 variables (Appendix K). Thus, the transformed data set was subject to the 2-way ANOVA as reported.

Chapter 4 – Kinematic and Kinetic Asymmetry Present in Athletic Populations Across Sex and Age for Walking, Running and Side-Cut Tasks

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Scott C. Landry

John Kozey

4.1 Introduction

Anterior cruciate ligament (ACL) injury is one of the most common and traumatic sports-related injuries to the knee joint. ACL injury results in an increased risk of reinjury to both the affected an unaffected limb[10–12], while increasing the likelihood for development of early onset osteoarthritis.[13–17] ACL injury is also associated with decreased participation in sport, and a rigorous rehabilitation following an invasive surgical procedure.[18–20] A recent study showed that ACL reconstruction rates have increased by 37 percent between 1994 and 2006.[23] At present, it is estimated that more than 400,000 ACL injuries occur annually in North America alone,[21,22] resulting in a burden on the healthcare system in excess of two-billion dollars annually.[24,25]

Approximately 70-80 percent of ACL injuries occur via a noncontact mechanism during the deceleration phase of a ballistic athletic maneuver such a cutting or landing. [1–7] Female athletes exhibit a 2-8 times greater likelihood of sustaining an acute non-contact injury of the ACL, with the highest frequency of these injuries occurring in females between 15 and 25 years of age.[2,26,35,27–34,38] Collectively, studies show little evidence for a sex disparity in ACL injury rates in pre-pubescent athletes[26,36,37], which is in harsh contrast to pubescent[30,32,34] and collegiate level or post-pubescent athletes.[29,31]

Over the past two decades, researchers have attempted to identify the biomechanical mechanisms involved to elicit noncontact ACL injury. This search has yielded several proposed risk factors which are broadly categorized as intrinsic or extrinsic. Extrinsic risk factors are deemed controllable and are those which occur outside of regular anatomical or physiological differences between individuals such as playing surface, footwear type or bracing. The bulk of the research, however, addresses intrinsic risk factors as they are largely considered uncontrollable. Intrinsic risk factors include anthropometry, as well as hormonal, neuromuscular and biomechanical influences. Research to this point has identified that ballistic athletic maneuvers place increased mechanical load on the ACL and thus increase the potential for ACL injury.[76–81] Researchers have also noted that the mechanism for ACL injury is multi-planar and dependent upon both joint positioning and loading.[82–85] Both the hip and knee have been associated with influencing the strain on the ACL, with frontal and sagittal joint angles and moments noted as injury risk factors.[230]

While a complete picture of ACL injury has not yet been resolved, recent epidemiological evidence suggests lateralization as a potential etiological factor in ACL injury.[141,272] The limb dominance theory states that there is an imbalance of muscular recruitment patterns and muscular strength between legs, which leads to differences in dynamic control.[118,119] Similarly, the concept of lateralization has been previously used to define differences between the operational and support limb. Given that its well understood that lateralization can lead to task-specific roles of the support and operational limb, it is conceivable that developed bilateral strength differences and/or differences in muscle recruitment may lead to the unconscious preference toward greater loading of one limb versus the other, thus resulting in asymmetry of lower extremity biomechanics. While development of asymmetry is often associated with the presence of pathology, healthy adults have been shown to be asymmetrical in both frontal and sagittal plane moments at the hip and knee joints during a normal walking task.[95] Asymmetry between limbs have been shown to increase with speed,[229] indicating potential for larger asymmetries during higher velocity tasks such as running and cutting tasks.

Hewett et al. (2005) was the first to demonstrate that asymmetry of the knee valgus moment is a predictor of future ACL injury in female athletes.[150] Likewise, asymmetry of the knee flexion moment has been shown to predict re-injury among a population of athletes who had previously underwent ACL reconstruction.[226] Studies using a jump-cut protocol found no differences between legs in both male and female collegiate athletes.[186] However, a more recent study using the same jump-cut protocol found differences between limbs in collegiate athletes. This study showed asymmetry in dynamic knee valgus angle at initial contact, whereby the self-reported non-dominant limb displayed the greater angle.[273] During bilateral drop jumps, Ford et al. (2003) reported side-to-side differences in knee abduction angle for females.[183] A more recent investigation by Pappas and Carpes (2012) supported this finding as they also reported greater knee abduction asymmetry in females as opposed to their male counterparts.[225] In contrast, others have reported limb dominance does not yield asymmetry in kinematic or kinetic measures.[185]

All of the aforementioned studies use a jump-cut or bilateral jump landing protocol, which only accounts for one athletic maneuver that commonly elicits non-contact ACL injury. To our knowledge, there exists only one published study investigating bilateral symmetry of lower limb biomechanics during an unanticipated side-cut task.[241] While asymmetry was not compared directly, Greska et al. (2016) compared the biomechanics of dominant and non-dominant limbs during the unanticipated side-cut maneuver to determine the effect of limb dominance on hip and knee mechanics for pre-contact, initial contact, peak knee adduction moment and peak stance periods (defined by the author as the first 50% of stance). Results from twenty female soccer players showed no significant differences in any hip or knee biomechanics between dominant and non-dominant limbs.[241] However, the authors make note that the small cohort of athletes may not have been sufficient enough for statistical power, and recommend that future investigation be done with a larger sample size.

Thus, there exists a gap in the literature in which there are no studies directly investigating bilateral asymmetry of the kinematics and kinetics of the lower limbs during a side-cut task. Moreover, there exists further gaps in the literature in comparing bilateral asymmetry of the kinematics and kinetics of the lower limbs across sex and age, for all tasks including walk, run and side-cut tasks. As a result, it puts an emphasis on understanding the role of asymmetry as a potential risk factor for ACL injury as it could lead to improvement of current intervention techniques and return to play guidelines. It seems logical then to construct a study for comparison of asymmetry for sagittal and frontal plane joint moments and angles across sex and age. This robust study design should provide evidence for or against asymmetry as a possible etiological factor for ACL injury given the inconsistency that currently exists in the literature. Therefore, the underlying goal of this study was to perform a comprehensive analysis of lower limb asymmetry of peak sagittal and frontal plane hip and knee internal joint moments, as well as peak sagittal plane hip and knee joint angles for healthy high performance athletes across sex and age during walk, run and side-cut tasks. Sagittal plane joint angles were also analyzed at initial contact. The purpose was to i) identify if asymmetry existed beyond a clinically accepted 10% threshold for lower limb biomechanical variables of interest in an athletic population of varying age and sex during a walk, run and side-cut task, ii) identify the proportion of each sub population (pre/post-pubescent males, pre/post-pubescent females) that experiences greater than 10% asymmetry for each of the biomechanical variables of interest, and iii) identify if differences in asymmetry exist across age and sex within an athletic population, to further the understanding of the sex bias of ACL injury as well as recognize how asymmetry may function as an etiological risk factor for ACL injury.

4.2 Methods

The majority of the methods employed in this study are comprehensively outlined in Chapter 3. The following methods are an abbreviated version specific to this chapter.

4.2.1 Participants

Based on a hierarchical approach, 122 high performance athletes were selected from a larger subset of participants if they played at the Professional, Provincial or Triple-A level (the highest club level) in their respective court or field sport for one full season at the time of testing. Participants were categorized as pre/early-pubescent (8-11 years old), or post-pubescent (17+). Anthropometrics for each group are summarized in Table 3.1.

Athletes did not meet inclusion criteria if they reported any sport other than a court of field sport as their primary sport of play. For sports which may require various skill sets and body types at varying positions (such as football), athletes were only selected if they played a skilled position which required frequent and explosive cutting. Participants also did not meet inclusion criteria if they reported a history of major trauma or injury to the lower extremities or lower back. Participants who reported an ankle sprain were only included if the injury had been healed for a minimum of three months prior to testing and they had been cleared to fully return to their sport by a trained medical professional. Prior to testing, each participant or their respective guardian read and signed the informed consent form that was approved by the Acadia University and Dalhousie University Research Ethics Board.

Participants wore their own shoes for testing and were asked to wear tight-fitting compression clothing to accommodate accurate retroreflective marker placement and tracking. Compression clothing was provided by the laboratory in instances where the participant did not provide their own.

4.2.2 Determination of Self-Selected Walking Speed and Maximum Sprint Speed

The participants began with four walking trials at their self-selected walking speed. Participants were instructed to walk naturally through two Fusion Sport timing gates (Fusion Sport Inc., Coopers Plains, AUS), set two meters apart at a distance five meters from the participant's starting point. This protocol was repeated until four walking trials had been collected. The average of these trials was used to calculate the participant's self-selected walking speed, which was then used as a control in subsequent walking trials within the mLAB.

Participants also completed two maximum sprint trials using the same gate setup but with a longer approach. Participants started with a 10-meter approach and ran at a maximum speed between the two Fusion Sport timing gates. Participant were instructed to sprint completely through the second gate to encourage maximal effort sprints. An average of the two sprint times determined the participant's maximal sprint speed. The approach speed for running and side-cut asks was determined for each participant by taking 66.6% ± 10% (two-thirds) of their own maximum effort sprint speed. In addition, a participant's maximum speed was used to control the distance the timing gates were to be adjusted in order to give each participant an approximate half second of reaction time to respond to the illuminated gates for the unanticipated side-cut trials.

4.2.3 Instrumentation

A full-body marker set of 14 mm retro-reflective markers were affixed to the participant with double sided tape to bony landmarks of the participant's thorax, and pelvis, as well as upper and lower limbs for the tracking of segment coordinate systems during the walk, run and cut trials. These markers were used to define and track a three-dimensional model that included the head, arms, trunk, pelvis, thighs, shanks, and feet, though only the biomechanical data of the lower extremity was analyzed in this study. Marker clusters, rigid plastic plates with markers fixed in each corner, were affixed on the shank and thigh over Fabrifoam wraps using Velcro. Marker clusters were secured to the legs using athletic tape and triad marker plates were positioned on each heel and secured using Duct tape. Lastly, a headband containing five retro-reflective markers was placed snugly around the head of the participant. In total, 75 markers were placed on the body (Figure 3.3).

A 13-camera Qualisys (Qualisys AB, Sweden) motion capture system (12 Oqus 4 cameras and one Oqus 210C camera) was used to collect kinematic data at a sampling rate of 250 hertz (Hz). Ground reaction force data was captured at 2000 Hz using three floor-embedded AMTI (Advanced Mechanical Technology Inc., Watertown, MA) strain gauge platforms in an L-shape configuration (Figure 3.4). Kinematic and kinetic data were simultaneously captured and synchronized using Qualisys tracker Manager (QTM) software (Qualisys, AB, Sweden).

4.2.4 Experimental Protocol

4.2.4.1 Calibration

Data collection began with the capture of a static and dynamic calibration trial to establish the anatomical coordinate systems and to calculate static reference joint angles. Ten medial retroreflective markers used solely for anatomical definition were then removed leaving 65 retroreflective markers during the motion trials. Markers that were removed were recreated as virtual markers during the movement trials using the marker cluster or individual markers on the same rigid segment.

4.2.4.2 Walking and Running Trials

In the walking trials, participants began walking at a distance of five meters from the first timing gate and in alignment with two adjacent force platforms (FP2/FP3) to allow for two sequential foot strikes on the embedded force platforms per trial (Figure 3.5). Fusion Sport timing gates were positioned both one meter prior and one meter after the midpoint of the adjacent force platforms, creating a two-meter timing window which was used to calculate walking speed. Participants were instructed to walk straight forward at a natural pace, keeping their eyes up until they reached the far wall of the laboratory. Walking speed was controlled to be the participants previously calculated self-selected walking speed ± 10%. Four foot strikes of both the left and right foot were collected.

For running trials, participants began running from a distance of 10 meters from the first timing gate and in the same alignment as in the walking trials. Participants were instructed to run through the gates and only begin deceleration after passing through the second gate. Running speed was controlled to be 66.6% ± 10% (two thirds) of each participant's average maximum sprint speed. Four foot strikes for both the left and right limb were collected. In either case, trials were excluded if the participant intentionally targeted the force platforms, had notable altered gait patterns, did not meet the proper speed requirement, or did not have at least one full foot strike on either force platform.

4.2.4.2 Cutting Trials

Participants began at a distance of 10 -from the center of the foremost force platforms (FP1/FP2)(Figure 3.5). The timing gates were configured in a Y-shape configuration with the force platforms located in the middle of the runway (Figure 3.5). The first two timing gates allowed the research team to control the participant's approach speed and were positioned at a variable distance from the center of the plates to allow 0.5 seconds of reaction time to cut either to the left or right. This distance was determined from the average maximum sprint speed calculated previously. Approach speed for the cuts was controlled to be 66.6 ± 10% (two-thirds) of each participant's average maximum effort sprint speed. Participants were instructed to run at the same speed as they had been for their running trials, with eyes focusing on the two-timing reaction gates positioned after the force platforms and at the end of the laboratory. Upon breaking the second timing gate, either the left or right timing gate at the end of the laboratory would be randomly queued to activate. Participants were instructed to maintain forward momentum but were encouraged to stutter step slightly to adequately perform a side-cut in the direction of the randomly queued gate, just as they may in a game-like situation. Participants were also instructed to cut and explosively sprint through the next timing gate such that their opposite foot made complete contact with the appropriate force platform. For leftward cuts, this meant cutting off the right foot planted on the right force platform (FP2); for rightward cuts, this meant cutting off the left foot planted on the left force platform (FP1). Tape was applied to the floor to guide the cutting angle of 45 ± 15° from the center of the two platforms as previously outlined in the literature.[73] Eight successful cut trials were collected with a minimum of four in each direction. Trials were excluded if the cut was not in the proper direction, did not meet the speed requirements, was not performed explosively, or if full stance did not occur on the proper force platform. Trials were also excluded if forward momentum was deemed too slow or stopped entirely, and if the cut was performed at an angle outside the designated range.

4.2.5 Calculation of Dependent Variables

Data were post-processed within the Visual3D (C-Motion Inc., Rockville, Maryland, USA) software to construct a biomechanical model composed of nine lower limb segments (pelvis, thigh, shank, rearfoot, forefoot) which were used to determine three-dimensional joint angles for the hip and knee. Low-pass fourth-order recursive Butterworth filters were applied to both the kinematic and kinetic data at a cutoff frequency of 12 Hz. This was consistent with previous side-cut studies.[75,260,261] Joint angles were calculated using an X-Y-Z (flexion/extension, adduction/abduction, internal/external rotation) order of rotation and measured in degrees with reference to the static calibration trial. Inverse dynamics techniques using ground reaction force data, kinematic data and segment inertial properties were utilized to calculate three-dimensional internal joint moments that occurred at each joint center during stance phase of each trial. Internal joint moments were normalized to the participant's body mass (N*m/kg) and reported about the X-Y-Z joint coordinate system.

Kinematic variables of interest included peak hip and knee flexion angles (pHFA, pKFA respectively), along with hip and knee flexion angles at initial contact (HFA_IC, KFC_IC respectively). Kinetic variables of interest included peak hip extension and abduction (pHEM, pHAM respectively), peak knee extension and abduction (pKEM, pKAM respectively) net internal joint moments. Stance phase was normalized for each trial to 101 data points and the individual trials were ensemble averaged for each task for both left and right legs of each participant.

Using a custom written MATLAB script, peak internal joint moments, peak joint angles and joint angles at initial contact were selected from the ensemble averaged waveforms for both the left and right limbs and were reclassified as "greater" or "lesser" based on their magnitude of the variable of interest to avoid underrepresentation of absolute asymmetry due to potentially taking an average of positive and negative values when calculating group means.

Asymmetry was calculated for each variable for each subject using the following formula:

$$\% \, Asymmetry = 100 \, x \, (1 - \frac{lesser \ value}{greater \ value})$$

This formula, adapted from a previously defined limb asymmetry index,[95,236] indicates the relative difference between limbs for each variable. In this earlier method, if moments are equal, it would result in zero asymmetry. Likewise, if the greater moment is twice that of the lesser moment there will be 50% asymmetry. Descriptive statistics of the percentage of asymmetry (% asymmetry) were calculated for each variable of interest in each age and sex group.

A clinically relevant value of asymmetry has not yet been established for joint angles or joint moments. However, 10% was chosen as a threshold for determining asymmetrical joint angles and moments for this study. This threshold is in accordance with other clinically relevant asymmetry indices for measuring differences in muscle strength and performance based testing, which are especially common in rehabilitation from injury, including injury to the ACL.[20,116,236]

4.2.6 Statistical Analysis

Percent asymmetry data for each biomechanical variable during the three tasks were imported into SPSS (Version 25, IBM, Armonk, New York, USA) and analyzed for normality using a Shapiro-Wilks Test. It should be noted that no data sets met the normality assumption, and outliers were present for the majority of biomechanical variables. Outliers were deemed to be real data points and not artifact of data collection or methodology, thus they were not corrected for or removed. A transformation was applied to the data of $\sqrt{x}$, where x was the value of percent asymmetry and both the transformed and untransformed data was analyzed for homogeneity of variance using the Levene's Test. Variables of the untransformed data set did not meet the assumption of homogeneity, while the transformed data did meet the assumption. Transformed and untransformed data sets were then subject to a two-way analysis of variance (ANOVA) with Tukey post-hoc comparisons to determine the main effects of sex and age for all dependent variables and among all tasks. It should be noted that results of the ANOVA for untransformed and transformed data were largely similar and thus, this study will present the p-values of the transformed data set while reporting the untransformed means since they are more easily understood from a clinical perspective.

A Chi-square test was used to determine differences between the proportion of participants with greater than 10% asymmetry within each group as well as to test if there was a statistically significant difference in the proportion of participants exhibiting greater than 10% asymmetry between each of the groups for each biomechanical variable and task.

4.3 Results

Data from 122 subjects were included in the analysis. No subjects were removed from the original cohort. Descriptive statistics of each population can be found in Table 3.1. The mean and standard deviation of percent asymmetry between lower limbs was calculated for each group and for the total athletic population for all kinematic and kinetic variables reported in Table 4.1 and Table 4.3 respectively. Likewise, the calculated proportion of each group and of the total population that exhibited greater than 10% (>10%) asymmetry is reported for calculated joint angles in Table 4.2 and internal joint moments in Table 4.4. Plots of main effects (age *or* sex) as well as the interaction effects (of age *and* sex) for percent asymmetry were plotted (Figure 4.2 / Figure 4.4) and p-values are reported in Table 4.1 and Table 4.3 for joint angles and internal joint moments respectively. Statistical significance was determined at $p<0.05$ and is noted by a highlight in the corresponding table. Chi-Square results, which detected statistically significant differences ($p<0.05$) between proportions of participants that exhibited greater than 10% asymmetry within each group are also reported in Table 4.1 and Table 4.4. The 95% confidence as well as Figure 4.1 and Figure 4.3. Graphically, 95% confidence intervals for the proportion of subjects in each population with greater than 10% asymmetry for peak hip and knee joint extension and abduction moments are shown in Figure 4.1 and Figure 4.3. From these intervals, we can be 95% certain that the percentage of each population exhibiting greater than 10% asymmetry will fall within the confidence intervals shown.

4.3.1 Kinematics

4.3.1.1 Hip

Statistically significant differences between groups were found using a 2x2 analysis of variance (ANOVA). Main effects of age were found for HFA_IC in all three tasks (Table 4.1 & Figure 4.2). Additionally, an age effect was found for pHFA during walk and run tasks. In all cases across the two measures of hip flexion, the pre-pubescent cohorts displayed greater mean percent asymmetry values than their post-pubescent cohort. No main effects of sex, or interaction effects were found using the ANOVA (Table 4.1 & Figure 4.2). The chi-square test did not find any statistically significant differences among the proportion of subjects in each group that exceeded 10% asymmetry for HFA_IC or pHFA in walk, run or cutting tasks (Table 4.2).

Although they not deemed to be statistically significant, mean values of percent asymmetry (% asymmetry) exceeded 10 percent for pHFA for pre-pubescent males and females across all tasks with the exception of pre-pubescent females during the running task (Table 4.1). Conversely mean asymmetry values of both post-pubescent male and female populations did not exceed 10 percent in pHFA across walking, running and cutting tasks. Likewise, the proportion of participants experiencing greater than 10% asymmetry was greatest for pre-pubescent populations for pHFA (Table 4.2). In summation, 36.9% of the total population exceeded 10% asymmetry for pHFA in the walking task, 29.5% during the running task, and 46.7% in the cutting task (Table 4.2).

This trend continued at mean asymmetry at initial contact. Mean HFA_IC exceeded 10% for both male and female pre-pubescent populations with the exception of pre-pubescent females in the running task (Table 4.1). Again, mean asymmetry values were greater for pre-pubescent populations than their post-pubescent counterparts across all tasks and in both sexes. Similarly, pre-pubescent populations experienced a greater proportion of athletes demonstrating greater than 10% asymmetry for all walk, run and cutting tasks (Table 4.2). Of the total population, 36.1% of athletes experienced greater than 10% asymmetry for both walk and run tasks, while 45.9% exceeded the 10% threshold for the cutting task (Table 4.2).

4.3.1.2 Knee

There existed only two main effects (sex or age) at the knee that were deemed significant among the four cohort subsets based on results of the 2x2 ANOVA. A (main) sex effect was found for KFA_IC (p=0.003) during the walking task where males displayed greater mean asymmetry than their female counterparts. A statistically significant age effect was found for pKFA (p=0.017) during the cutting task (Table 4.1 & Figure 4.2). Again, in this case pre-pubescent populations displayed greater percent asymmetry values. Likewise, differences in proportion of individuals with greater than 10% asymmetry were noted between groups using a chi-square test (Table 4.2). Post-pubescent males had a significantly greater proportion (78.8%) of individuals exhibiting greater than 10% asymmetry for pKFA in the walk task (p=0.045), while pre-pubescent males showed a significantly greater proportion exhibiting greater than 10% in pKFA during the cutting task (Table 4.2).

As for non-statiticially significant trends, asymmetry of both peak knee flexion angles and flexion angles at initial contact displayed a pattern which was not evident among any other biomechanical variables – either kinematic or kinetic – in this study, whereby asymmetry of knee flexion was found to be greatest during the walking task and least during the cut task (Table 4.1). The proportion of participants displaying greater than 10% asymmetry also supported this pattern (Table 4.2).

Mean percent asymmetry exceeded 10% for pKFA for all groups during the walking task (Table 4.1). Post-pubescent males displayed the greatest mean pKFA at 20.04% while pre-pubescent males displayed the lowest mean asymmetry, 12.94%, during the walking task (Table 4.1). 59.8% of athletes experienced greater than 10% asymmetry for the walking task with post-pubescent males exhibiting the greatest proportion among the groups at 78.8% (Table 4.2). No mean asymmetry values exceeded 10% for pKFA during the running task. In this case pre-pubescent males displayed the greatest mean asymmetry at 8.46% (Table 4.1). Both pre-pubescent and post-pubescent males exhibited greater proportions of athletes experiencing greater than 10% asymmetry than their female counterparts at 33.3% and 36.4% respectively. For the side cut task, pre-pubescent males were the only group whose mean asymmetry value exceed 10 percent (11.43%). Likewise, pre-pubescent males experienced the greatest proportion of athletes with greater than 10% asymmetry at 50.0% (Table 4.2). Both pre-pubescent males and females displayed greater mean asymmetry than their post-pubescent counterparts of the same sex, however, these differences were not statistically significant (Table 4.1).

KFA_IC exhibited the most asymmetry of any of the kinematic variables across all groups, and for all tasks. Mean percent asymmetry for KFA_IC during the walking task exceeded 49.42% for all groups with the greatest percentages being both pre and post-pubescent males at 82.71 and 78.01% respectively (Table 4.1). However, the proportion of athletes who experienced greater than 10% asymmetry ranged between 87.5% for post-pubescent females and 100% for post-pubescent males (Table 4.2). In total 93.4% of participants exceeded the 10% threshold for KFA_IC during the walking task (Table 4.2). During the running task mean asymmetry values decreased in magnitude and ranged from 22.88 in pre-pubescent females to 31.15 in post-pubescent males (Table 4.1). The proportion of athletes exhibiting greater than 10% asymmetry also decreased. Again, post-pubescent females exhibited the lowest proportion (72.5%) while post pubescent males exhibited the greatest (84.5%). The proportion of athletes exceeding the 10% threshold for KFA_IC again decreased for the cutting task; however, pre-pubescent males exhibited the lowest proportion, at 66.7% while post-pubescent males exhibited the greatest proportion, still, at 81.8% (Table 4.2).

4.3.2 Kinetics

4.3.2.1 Hip

Asymmetry measures were not statistically different among the four cohorts based on the results of the 2x2 ANOVA for walk, run or cut tasks. In each case, no main effects of age, or sex were detected, nor was an interaction effect of sex and age identified (Table 4.3 & Figure 4.4). No group experienced a statistically significant difference in the proportion of athletes experiencing greater than 10% asymmetry based on the results of the Chi-Square test (Table 4.4).

Statistically non-significant findings included that mean values of percent asymmetry exceeded 10 percent for pHEM across all groups and tasks, with the exception of pre-pubescent females during the walking task (Table 4.3). This would indicate that on average each group exceeded our clinically relevant threshold for asymmetry for each of the respective tasks for pHEM. The greatest mean pHEM % asymmetry for walking and running tasks occurs in the pre-pubescent male group, while pre-pubescent females demonstrated the greatest mean asymmetry for pHEM during the cut task (Table 4.3).

The proportion of individuals who experienced greater than 10% asymmetry exceeded 40.0% for all groups and across all tasks. Overall 45.9% of the population exceeded 10% asymmetry for pHEM in the walking task, 48.4% for running task, while 63.1% of participants exceeded 10% asymmetry for the cutting task (Table 4.4).

The mean percent asymmetry of all internal hip abduction moments (pHAM) for all groups exceeded 10 percent during each of the walking, running and cutting tasks (Table 4.3). Again, pre-pubescent males displayed the greatest mean asymmetry values for walking and running tasks, while post-pubescent males displayed the greatest asymmetry for the cutting task at 37.13% (Table 4.3). Though task was not compared in this analysis, the mean calculated percentage of asymmetry for the cutting task was between 2-3 times greater than either the walk or run task for both sexes when examining the pHAM. In each task, greater than 57.5% of the population exceeded 10% asymmetry for pHAM. Overall, 61.5% of the entire athletic population exceeded the 10% threshold for the walking task, 59.8% for the running task, and 83.6% for the cutting task.

4.3.2.2 Knee

Asymmetry measures were statistically significant among the four group cohorts based on results of the 2x2 ANOVA. A (main) sex effect was found for pKEM ($p=0.024$) during the running task where males were significantly greater than their respective female cohorts (Table 4.3 & Figure 4.4). These results were supported by the Chi-Square test which showed a strong trend toward a higher or lower proportion of participants exhibiting greater than 10% asymmetry during the run task for pKEM (Table 4.4). Additionally, a statistically significantly age (main) effect was found for pKAM during the walking task (Table 4.3) where pre-pubescent athletes displayed greater asymmetry values than the post-pubescent cohorts. No population had a statistically significant proportion of individuals exhibiting greater than 10% asymmetry based on the Chi-Square test (Table 4.4).

In statistically non-significant findings included that % asymmetry of pKEM did not increase with task difficulty as the hip did. Mean asymmetry for pKEM exceeded 18.97% for all groups during the walking task with the greatest asymmetry of any group occurring for pre-pubescent females at 22.52%. However, during running tasks the mean percent asymmetry for each group decreased (Table 4.3) such that the greatest mean % asymmetry was 13.13% and it occurred in pre-pubescent males. Pre-pubescent and post-pubescent female groups did not exceed 10 percent for mean % asymmetry at 8.63% and 7.72% respectively (Table 4.3). Pre-pubescent boys displayed the greatest proportion of the population exceeding 10% asymmetry for both walking and running tasks, while pre-pubescent females displayed the greatest proportion of asymmetry (80.0%) in pKEM during the cutting task (Table 4.4).

pKAM showed the greatest proportion of total participants experiencing greater than 10% asymmetry for walk (70.5%), run (77.0%) and cut (96.7%) tasks (Table 4.4). For all tasks, pre-pubescent populations displayed a greater proportion of participants above the clinically accepted 10% threshold than their post pubescent counterparts. 100% of pre-pubescent males displayed greater than 10% asymmetry, while the least prevalent group were post-pubescent males, although this group still experienced 93.9% of participants above the 10 percent clinical threshold during the side-cut task (Table 4.4). Similarly, mean values of pKAM exceeded 67.96% for all groups, with pre-pubescent males experiencing the greatest mean pKAM value at 81.68% and post-pubescent females experiencing the least at 67.96% (Table 4.3).

Table 4.1. Group mean and standard deviation values of percent asymmetry for hip and knee flexion angles during walk, run and side-cut tasks. Main and interaction effects for 2-way ANOVA are listed for each variable and task.

		Male		Female		Age	Sex	Age*Sex
Variable	Task	Pre	Post	Pre	Post	p-Value	p-Value	p-Value
pHFA	Walk	11.94 ± 9.18	8.86 ± 6.16	10.51 ± 7.24	6.89 ± 5.83	0.013*	0.158	0.769
	Run	10.75 ± 8.14	6.95 ± 6.64	8.76 ± 6.57	7.22 ± 6.12	0.029*	0.606	0.412
	Cut	13.67 ± 9.09	9.31 ± 7.56	11.86 ± 10.41	9.69 ± 6.33	0.105	0.662	0.370
HFA_IC	Walk	11.01 ± 9.88	8.9 ± 6.95	11.11 ± 7.21	7.06 ± 5.4	0.028*	0.759	0.561
	Run	12.05 ± 7.79	7.57 ± 7.2	9.77 ± 6.34	8.61 ± 6.87	0.020*	0.832	0.207
	Cut	14.53 ± 10.57	9.14 ± 7.77	15.86 ± 13.43	8.98 ± 6.44	0.002*	0.702	0.874
pKFA	Walk	12.94 ± 8.06	20.04 ± 23.93	13.77 ± 14.28	13.75 ± 13.22	0.227	0.129	0.318
	Run	8.46 ± 7.33	8.08 ± 5.21	7.28 ± 5.74	6.04 ± 4.41	0.682	0.163	0.538
	Cut	11.43 ± 10.67	6.73 ± 5.05	8.41 ± 5.31	6.61 ± 4.37	0.017*	0.451	0.405
KFA_IC	Walk	82.71 ± 55.14	78.01 ± 46.89	49.42 ± 32.98	58.9 ± 49.46	0.798	0.003*	0.679
	Run	27.01 ± 22.07	31.15 ± 20.22	22.88 ± 15.76	24.23 ± 22.21	0.473	0.142	0.492
	Cut	21.19 ± 14.81	18.89 ± 10.72	24.64 ± 20.74	15.71 ± 9.85	0.145	0.791	0.276

Note: Age*Sex denotes an interaction effect of sex and age.
*** Indicates a statistically significant difference ($p < 0.05$).**

Table 4.2. Percentage of the population exhibiting >10% asymmetry for hip and knee flexion angles during walk, run and side-cut tasks. Chi-square results showing statistical differences between groups in the proportions of athletes experiencing greater than 10% asymmetry.

		Male		**Female**			**Chi-Square**
Variable	Task	Pre	Post	Pre	Post	Total	p-value
pHFA	Walk	45.8%	36.4%	48.0%	25.0%	36.9%	0.205
	Run	37.5%	18.2%	44.0%	25.0%	29.5%	0.128
	Cut	58.3%	39.4%	52.0%	42.5%	46.7%	0.461
HFA_IC	Walk	41.7%	39.4%	52.0%	20.0%	36.1%	0.052
	Run	45.8%	27.3%	40.0%	35.0%	36.1%	0.515
	Cut	66.7%	39.4%	52.0%	35.0%	45.9%	0.071
pKFA	Walk	62.5%	78.8%	48.0%	50.0%	59.8%	0.045*
	Run	33.3%	36.4%	28.0%	22.5%	29.5%	0.596
	Cut	50.0%	18.2%	32.0%	17.5%	27.0%	0.02*
KFA_IC	Walk	95.8%	100.0%	92.0%	87.5%	93.4%	0.177
	Run	79.2%	84.5%	76.0%	72.5%	77.9%	0.642
	Cut	66.7%	81.8%	72.0%	70.0%	73.0%	0.576

*** Indicates a statistically significant difference ($p < 0.05$).**

Table 4.3. Group mean and standard deviation values of percent asymmetry for sagittal and frontal plane hip and knee internal joint moments during walk, run and side-cut tasks. Main and interaction effects for 2-way ANOVA are listed for each variable and task.

		Male		**Female**		**Age**	**Sex**	**Age *Sex**
Variable	Task	Pre	Post	Pre	Post	p-Value	p-Value	p-Value
pHEM	Walk	10.99 ± 8.71	10.75 ± 8.68	8.97 ± 6.97	10.17 ± 7.54	0.877	0.634	0.595
	Run	12.19 ± 9.71	10.97 ± 9.04	11.50 ± 8.75	12.02 ± 10.93	0.834	0.917	0.636
	Cut	18.88 ± 16.14	13.58 ± 10.30	19.06 ± 18.12	17.17 ± 11.89	0.299	0.509	0.362
pHAM	Walk	17.23 ± 13.01	12.61 ± 8.21	16.98 ± 11.46	14.99 ± 9.07	0.165	0.526	0.454
	Run	17.34 ± 13.17	15.56 ± 11.42	13.00 ± 8.51	13.89 ± 11.15	0.648	0.204	0.877
	Cut	35.90 ± 19.11	37.13 ± 23.01	29.48 ± 18.83	33.18 ± 27.14	0.812	0.119	0.814
pKEM	Walk	19.11 ± 13.72	18.97 ± 13.10	22.52 ± 18.00	21.38 ± 19.24	0.727	0.731	0.766
	Run	13.13 ± 10.88	10.21 ± 6.85	8.63 ± 5.76	7.72 ± 5.40	0.233	0.024*	0.575
	Cut	19.16 ± 20.33	14.91 ± 12.02	15.79 ± 8.00	12.78 ± 7.83	0.094	0.774	0.913
pKAM	Walk	23.57 ± 14.42	15.81 ± 11.60	23.83 ± 14.87	19.08 ± 13.81	0.015*	0.626	0.565
	Run	31.03 ± 21.53	25.99 ± 16.98	28.36 ± 19.14	26.88 ± 18.67	0.475	0.716	0.606
	Cut	81.68 ± 50.44	71.12 ± 46.16	79.64 ± 48.97	67.96 ± 34.33	0.264	0.931	0.945

Note: Age*Sex denotes an interaction effect of sex and age.
*** Indicates a statistically significant difference ($p < 0.05$).**

Table 4.4. Percentage of the population exhibiting >10% asymmetry for sagittal and frontal plane hip and knee joint moments during walk, run and side-cut tasks. Chi-square results showing statistical differences between groups in the proportions of athletes experiencing greater than 10% asymmetry

		Male		**Female**			**Chi- Square**
Variable	Task	Pre	Post	Pre	Post	Total	p-value
	Walk	54.2%	48.5%	40.0%	42.5%	45.9%	0.732
pHEM	Run	54.2%	42.4%	56.0%	45.0%	48.4%	0.67
	Cut	62.5%	60.6%	60.0%	67.5%	63.1%	0.913
	Walk	58.3%	57.6%	64.0%	65.0%	61.5%	0.899
pHAM	Run	70.8%	57.6%	56.0%	57.5%	59.8%	0.677
	Cut	91.7%	93.9%	76.0%	75.0%	83.6%	0.074
	Walk	75.0%	66.7%	68.0%	65.0%	68.0%	0.865
pKEM	Run	54.2%	48.5%	36.0%	25.0%	39.3%	0.074
	Cut	58.3%	60.6%	80.0%	57.5%	63.1%	0.269
	Walk	75.0%	63.6%	84.0%	65.0%	70.5%	0.289
pKAM	Run	87.5%	75.8%	76.0%	72.5%	77.0%	0.573
	Cut	100.0%	93.9%	96.0%	97.5%	96.7%	0.629

*** Indicates a statistically significant difference ($p < 0.05$).**

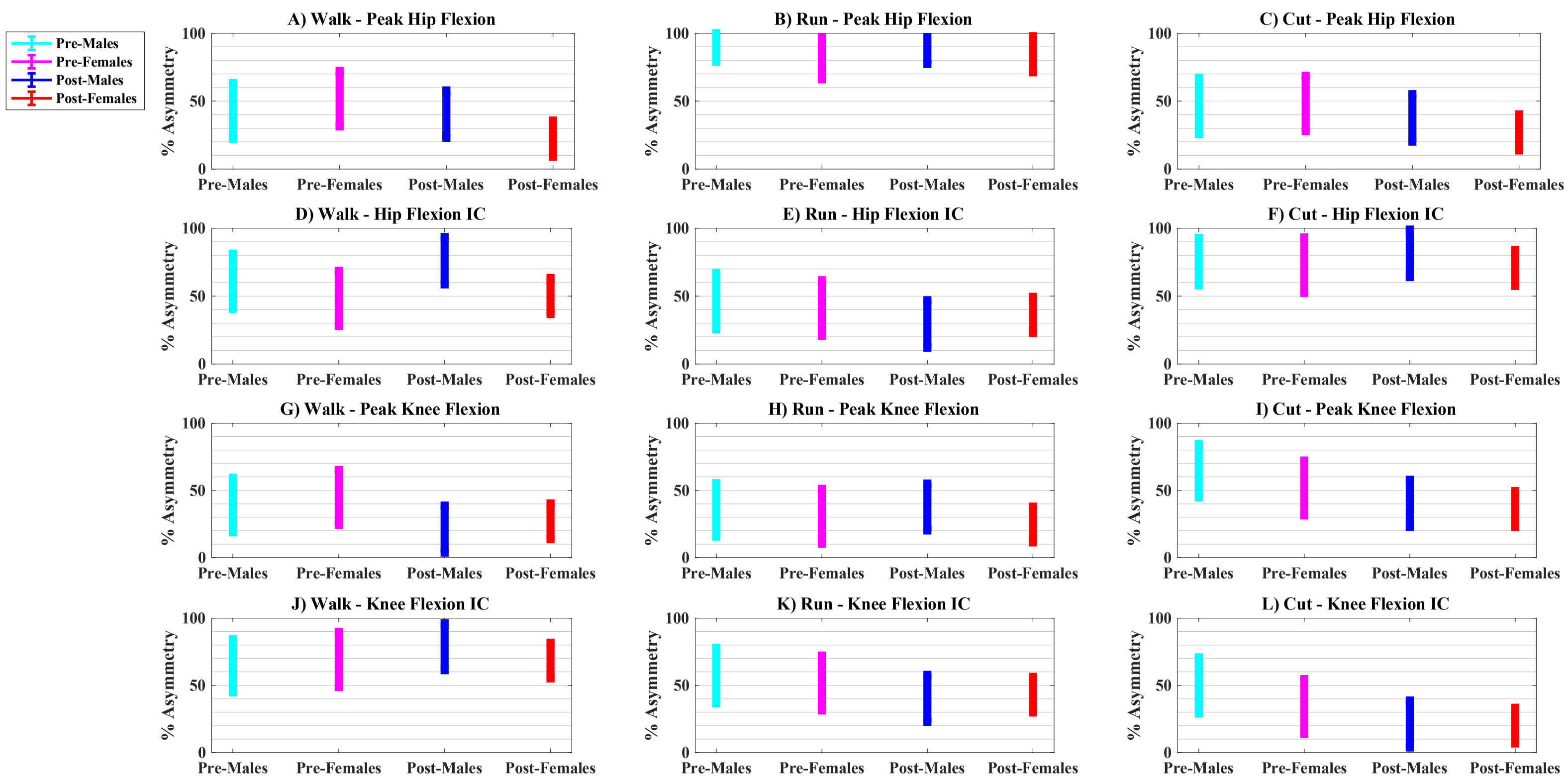

Figure 4.1. Confidence intervals for the proportion of subjects in each population with greater than 10% asymmetry for hip and knee joint flexion angles at peak flexion and instant of contact.

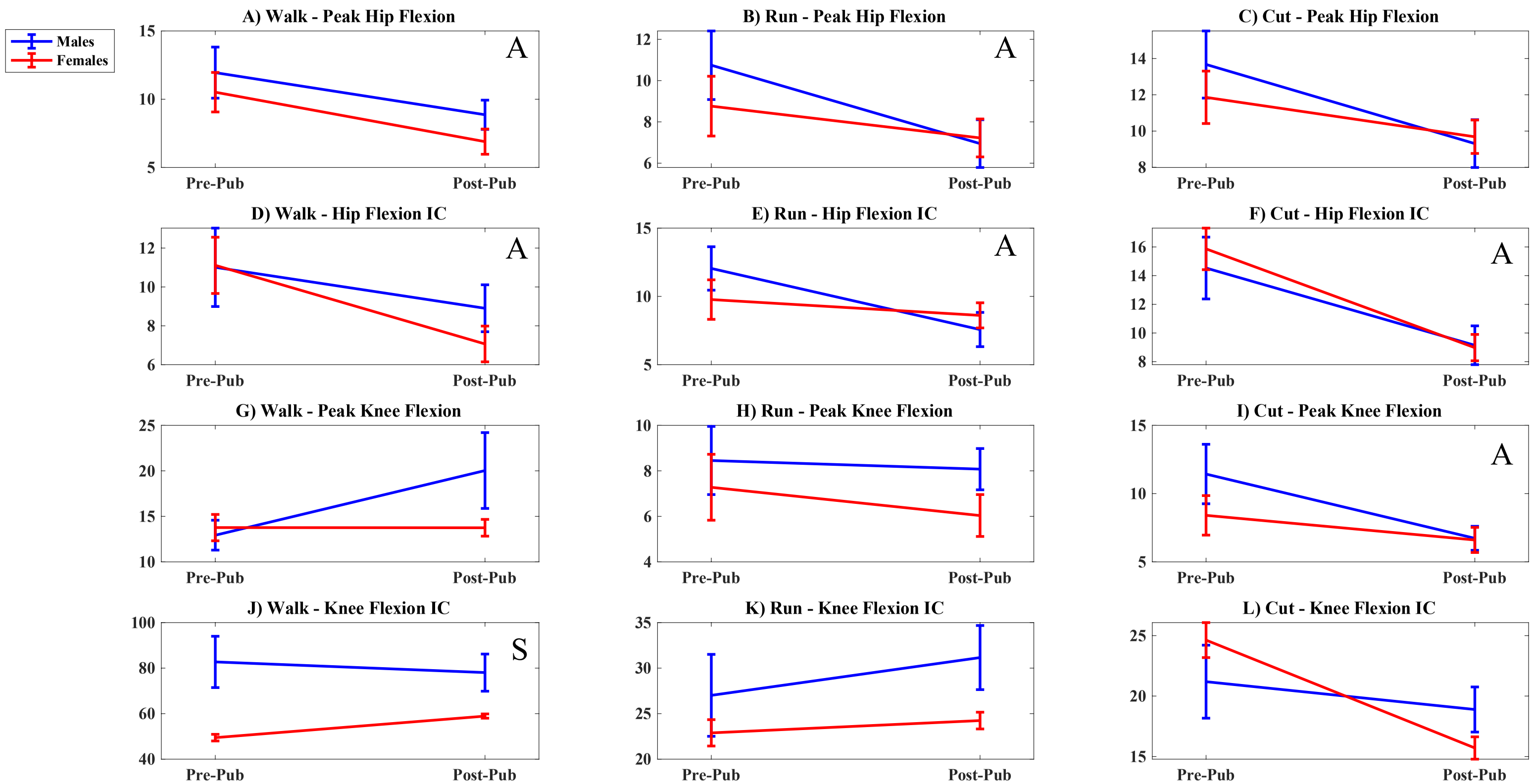

Figure 4.2. Average joint asymmetry for hip and knee flexion angles with standard error at peak flexion and instant of contact (IC) for walk, run and cut tasks. 'A' refers to a statistically significant main effect of age, while 'S' refers to a statistically significant main effect of sex.

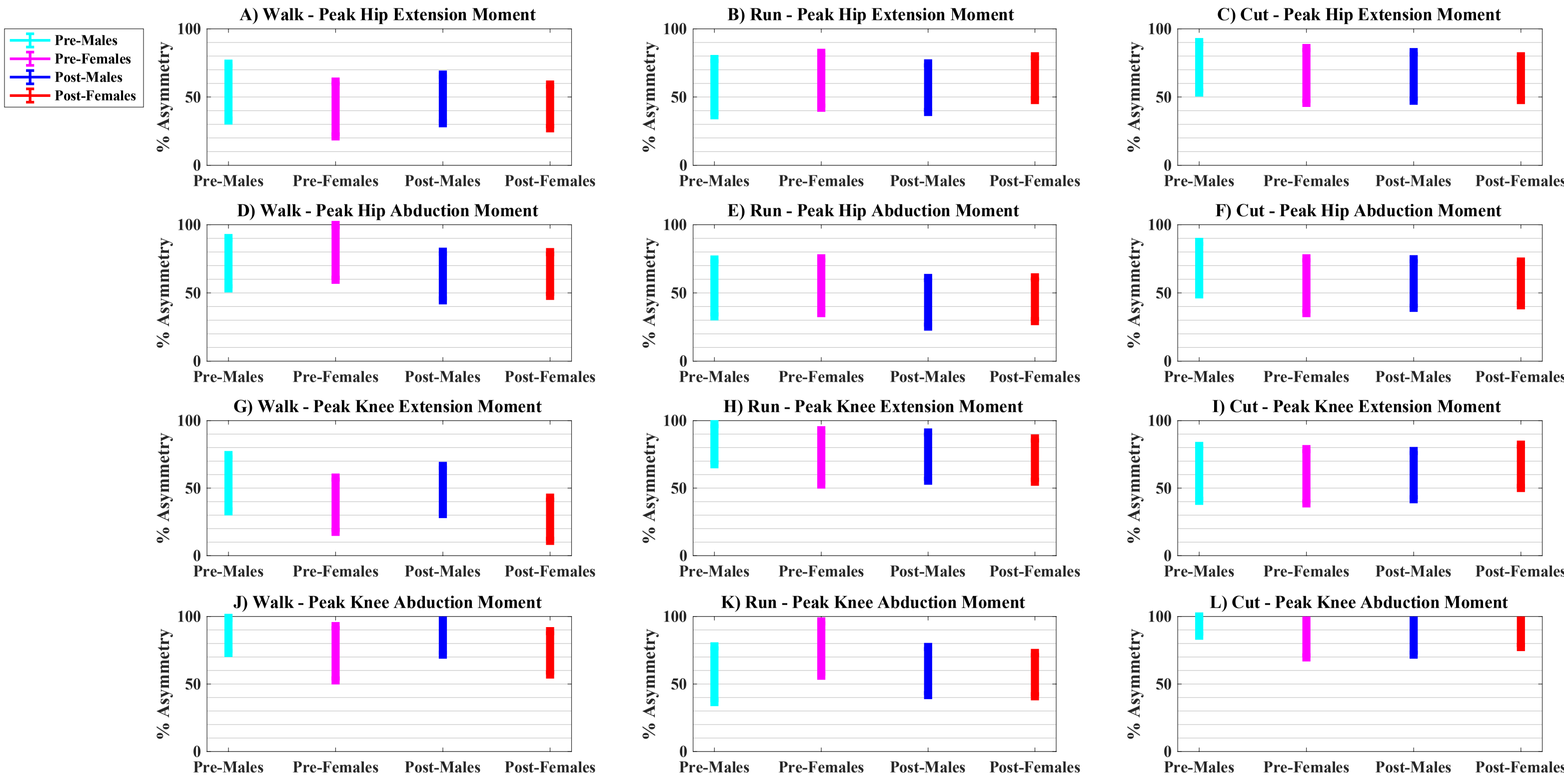

Figure 4.3. Confidence intervals for the proportion of subjects in each population with greater than 10% asymmetry for peak hip and knee joint flexion and adduction moments. From these intervals, we can be 95% certain that the percentage of each population exhibiting greater than 10% asymmetry will fall within the confidence intervals shown.

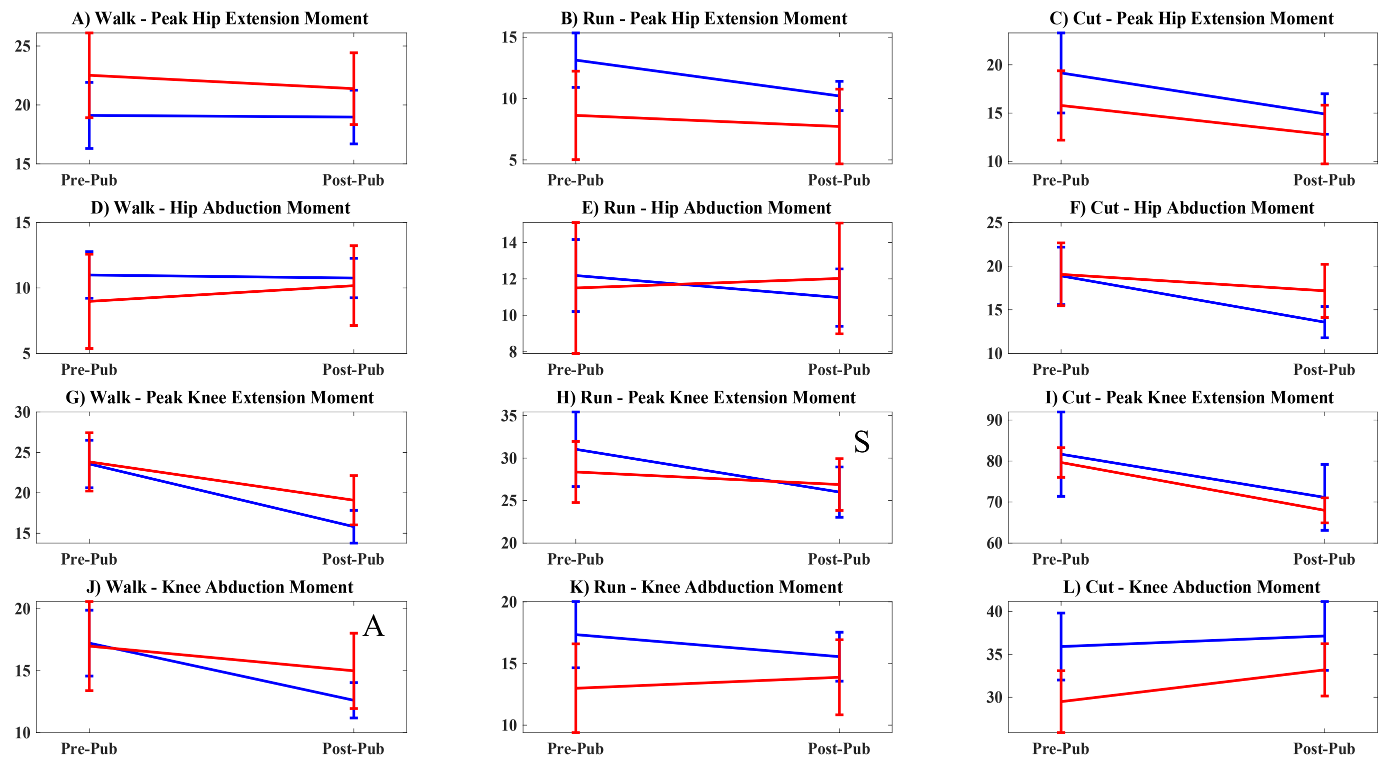

Figure 4.4. Average joint asymmetry of peak hip and knee flexion and peak hip and knee abduction moments with standard error for walk, run and cut tasks. ‘A’ refers to a statistically significant main effect of age, while ‘S’ refers to a statistically significant main effect of sex

4.4 Discussion

The objectives of the study were threefold: i) identify if asymmetry existed beyond a clinically accepted 10% threshold for lower limb biomechanical variables of interest in an athletic population of varying age and sex, ii) identify the proportion of each sub population (pre/post-pubescent males, pre/post-pubescent females) that experienced greater than 10% asymmetry for each of the biomechanical variables of interest, and iii) to identify if differences in asymmetry existed across age and sex within an athletic population to further the understanding of the sex bias of ACL injury as well as recognize how asymmetry may function as an etiological risk factor for ACL injury. This study will be the first to use a clinically relevant definition of symmetry to investigate lower limb biomechanics for an entire population of healthy high-performance athletes during gait. In addition, this study will be the first study to use that same clinically relevant definition to investigate asymmetry of lower limb biomechanics for an athletic population during sport-specific athletic tasks such as running, and a side-cut. Finally, this study will be the first to assess asymmetry of the lower limbs for walk, run and side-cut tasks across age and sex.

4.4.1 Existence and Etiology of Asymmetry

The primary finding of this study was that the percentage of asymmetry peak sagittal and frontal plane joint moments and sagittal plane joint angles at peak flexion and instant of contact during walk, run and side-cut tasks were greater than expected in all of our populations, for all variables. More than half of our overall population exhibited greater than 10% asymmetry during all tasks (walk, run and side-cut) for KFA_IC, pHAM, pKEM, and pKAM (Table 4.2 & Table 4.4). Greater than half the population also exceeded 10% asymmetry for pKFA during the walk task, and pHEM during the side-cut task (Table 4.2 & Table 4.4). Of the variables and groups investigated in this study, the lowest proportion of the entire athletic cohort exceeding the clinically relevant 10% threshold occurred for pKFA during the cut task (Table 4.2).

The results of the present investigation are consistent with previous research that demonstrates asymmetry in lower extremity joint moments during gait.[95,101] Lathrop-Lambach et al. (2014) used a population of 182 healthy adults (across four diverse groups) to detect asymmetry of lower limb external joint flexion and adduction moments at the hip and knee joint. Similar to this study, more than 50% of the overall population exhibited greater than 10% asymmetry for peak external hip and knee flexion and adduction moments for healthy young, healthy old, healthy obese and healthy university football linemen.[95]

However, the results appear to contradict research elsewhere in the literature that suggests symmetry of the lower limbs,[99,241] including the only other study to investigate asymmetry during the side-cut task.[229] This study, by Greska et al. (2016), used only a female collegiate, athlete population but found no differences in internal joint moments between limbs. The author notes that the lack of differences in neuromuscular timing, kinematic and kinetic parameters may be a factor of athletic experience or expertise, however, that does not appear to be a factor in the present study. Similarly, Teichtahl et al. (2009) found no difference in asymmetry in external peak knee adduction moments between of the dominant versus non-dominant limb of healthy individuals.[99] However, this study labelled the dominant and non-dominant limb prior to the calculation of asymmetry. This resulted in both positive and negative results of asymmetry between limbs within each group. As such, these positive and negative asymmetry values may have led to the underestimation of the absolute asymmetry between limbs. By grouping our joint moment, and joint angle data by magnitude (greater/lesser) our calculated asymmetries were always positive giving a more accurate result of absolute asymmetry between limbs for group analyses. The asymmetry index used in this study also took into consideration the clinical significance of lower limb biomechanics that occur in opposing directions. While the greater and less labelling of limbs may calculate a clear absolute asymmetry and eliminate the issue of negative asymmetry, it loses clinical significance in doing so by treating the same magnitude of motion in opposing motions (flexion versus extension) as equal and symmetric. The asymmetry calculation in this study used an absolute value to determine the rank of greater or lesser variables but did not factor absolutes into the actual calculation of asymmetry. For limbs moving in the same direction – flexion for example – the calculation would return a standard score of asymmetry ranging between 0-100%. However, for limbs in opposing directions – flexion versus extension – asymmetry scores ranged from 100-200% which retained clinical significance. Further discussion of the asymmetry index calculation is highlighted in section 5.3.3 of this document and in the appendices of this study (Appendix I).

The vast prevalence of lower limb asymmetry among all four athletic populations in this investigation may be due to the repetitive use of a particular limb to perform a sport-specific task such as kicking. This has previously been noted in the literature by Rahnama, Lees and Bambaecichi (2005).[214] This group reported that knee flexors of the operational kicking leg of adult soccer athletes were weaker than the non-preferred supporting leg.[214] Intuitively, from a strength perspective, the supporting leg would act as the load bearing leg during the kicking motion and thus the knee flexors would be responsible for helping to stabilize the joint while supporting the transfer of weight and energy throughout the kicking motion. Conversely, knee extensors would be activated in the operational limb to extend the leg and pull the shank through the kicking motion. All athletes who took part in the present study were considered to be high-performance athletes, thus the amount of repetition of their respective sport-specific tasks may increase lower limb asymmetry.[214]

The concept of limb dominance has also been proposed as an explanation for sex differences in ACL injury incidence. The limb dominance theory states that there is an imbalance of muscular recruitment patterns and muscular strength between legs, which leads to differences in dynamic control between limbs.[118,119] Similarly, the concept of lateralization has been previously used to define differences between the operational and support limb. Given that it is well understood that lateralization can lead to task-specific roles of the support and operational limbs, it is conceivable that developed bilateral strength differences and/or differences in muscle recruitment may lead to limb dominance. The expression of limb lateralization is evidence for the unconscious preference toward greater loading of one limb versus the other, thus resulting in asymmetry of lower extremity biomechanics. Previous research has supported this theory, as post-pubescent females have been shown to exhibit greater interlimb asymmetries compared to their male counterparts.[183,225] Thus, the current study hypothesized that post-pubescent females would exhibit the greatest percentage asymmetry for the biomechanical variables of interest during the side-cut task. However, this hypothesis was not supported by the calculated mean values of asymmetry. Post-pubescent females did, however, display the greatest proportion of asymmetry for pHEM as 67.5% of the post-pubescent female population exceeded 10% for the pHEM during the cut task. This is in agreement with Brown et al. (2009) which found post-pubescent females demonstrated substantially different hip postures compared to males indicating altered hip strategies. These altered biomechanics have been previously associated with potential high-risk knee joint loading.[273] However the findings of the present study were determined to be non-significant between age and sex. The proportion of post-pubescent females exhibiting greater than 10% asymmetry was also shown to be non-significant relative to the other age-sex groupings via a chi-square test.

4.4.2 Differences Among Age and Sex

One of the strengths of this investigation is our identification of similar levels of asymmetry across all age and sex groups. This is the first study to present asymmetry data across participants of pre-pubescent and post-pubescent populations. In addition, it is the first study to investigate the effects of sex and age on asymmetry of the lower-limbs during walk, run or side-cut tasks. This is important to highlight as no statistically significant interaction effects of sex and age were present in our investigation of % asymmetry for internal joint extension and abduction moments (Table 4.2 & Figure 4.4), peak flexion angles, or flexion angles at initial contact at the hip or knee joint ($p>0.05$, Table 4.1 & Figure 4.2). Thus, no combination of age and sex, (pre-pubescent males/females, post-pubescent males/females) differed from one another. This is likely a result of a high variance among percent asymmetry values within all participants as noted by the large standard deviations across both kinetic and kinematic variables (Table 4.1 & Table 4.3). While it may be logical to assume that variance increased with task difficulty (from walk to run and from run to cut), the standard deviation values did not always increase with task. This suggests that perhaps the measure for percent asymmetry should potentially be normalized in a manner to reduce the overall variance in the data set. One such solution has recently been provided in the literature as the Normalized Symmetry Index (NSI), whereby the absolute difference between the ensemble averaged variable of interest is divided by the absolute difference between the maximum and minimum values over the all trials (Appendix I).[233] This method could potentially decrease the overall variance of each population sample, and lead to increased findings of both main (sex and age) effects or interaction effects of sex and age. This would present an additional solution to simply increasing the sample size, as the relative amount of variance would lead to the need for an extremely large sample which may be unrealistic. This is an important finding for future research studies using measure of asymmetry as an outcome.

However, a sex effect in percent asymmetry was detected for two variables, KFA_IC during the walk task (p=0.003) and pKEM during the running task (p=0.024). In both cases, males showed greater asymmetry than their female counterparts at both pre-pubescent and post-pubescent age levels. While these results may not align with ACL injury rates, it does not exclude that these movement patterns may be hazardous and may still threaten the ACL. This result highlights that differences in knee joint kinematics and kinetics exist between sex for an athletic population. It is reasonable to hypothesize that post-pubescent males may have more developed movement patterns or learned coping strategies to mitigate the effects of the asymmetries within the knee joint outside of the laboratory setting. Thus, when females are subject to sudden, abnormal asymmetric loading they may not have developed the same coping mechanisms, making them more prone to ACL rupture.

Investigations of ACL injury rates show post-pubescent females experience a 2-8 times greater rate of non-contact ACL injury than their male counterparts.[2,26,35,27–34] Additionally, bilateral asymmetry and limb dominance have been shown as a potential etiological risk factor for ACL injury.[119,224,225] Thus, if we attribute greater percent asymmetry with higher rate of injury, the present study logically hypothesized that post-pubescent females would exhibit the greatest percent asymmetry across any combination of age and sex groupings. This hypothesis, however, was not supported by the results of this study. A simple investigation of the magnitude of mean percent asymmetry values showed that post pubescent females did not exhibit the greatest mean percent asymmetry for any of the biomechanical variables of interest, either kinematic or kinetic. Post-pubescent females did exhibit the greatest proportion of the population with greater than 10% asymmetry for the pHEM during the cut task, however, indicating the greatest median value for percent asymmetry of any of the groups. The magnitude of post-pubescent female pHEM was found to be non-significant relative to the other groups as was the proportion of post-pubescent female participants with greater than 10% asymmetry relative to the other groups. Contrary to our hypothesis, post-pubescent females showed the smallest magnitude of mean percent asymmetry for HFA_IC, pHFA and pHAM during the walking task; KFA_IC, pKFA, and pKEM during the run task; and HFA_IC, KFA_IC and pKEM in the cutting task. These results contradict previous work by Hewitt et al. (2004) who had post-pubescent females demonstrated greater bilateral asymmetry compared to pre-pubescent and early pubescent females. However, the findings of this investigation examined frontal plane joint angles which are not included in the present study.[63]

The post-pubescent female cohort included in this study was largely composed of collegiate athletes. Being that post-pubescent females have been shown in the literature to be more likely to sustain a non-contact ACL injury[2,26,35,27–34] the majority of current prevention programs are targeted toward addressing strength deficiencies and possible hazardous movement patterns for female athletes.[248,274,275] Given the mean age of our collegiate athlete population, it is conceivable to think that this population has been subject to training programs from a younger age, and thus may have developed more symmetrical joint kinematics and kinetics through these prevention and training programs. Evidence of prior prevention or training programs was not considered a co-variate for this study.

Currently there appears to be no study in the literature that compares bilateral asymmetry between pre-pubescent male and female athletes. Since these two groups experience a 1:1 ratio of injury incidence, it was hypothesized that pre-pubescent males and pre-pubescent females would exhibit similar levels of asymmetry. There were no sex effects to indicate that pre-pubescent males differed from pre-pubescent females in percent asymmetry for walking or running tasks. With relevance to ACL injury, there were also no sex differences between pre-pubescent males and females for kinematic or kinetic variables during the side-cut task specifically. This would indicate that the relative magnitudes of percent asymmetry were similar for all biomechanical variables (excluding KFA_IC and pKEM as previously discussed) between the pre-pubescent populations.

Since increased percent asymmetry has been suggested to be linked with increased ACL injury risk, it was hypothesized that pre-pubescent athletes would exhibit less bilateral asymmetry than post-pubescent athletes. Age differences in percent asymmetry were present across all tasks for various internal joint moments and for peak joint angles as well as joint angles, at initial contact. However, contrary to our hypothesis, pre-pubescent athletes experienced greater bilateral asymmetry than their post-pubescent counterparts in every case (Table 4.1 / Table 4.4) for both sexes. Pre-pubescent participants may have exhibited significantly greater asymmetries to their post-pubescent counterparts due to lack of experience and refined motor coordination in their respective sport. The post-pubescent cohort, both males and females, were comprised of primarily collegiate level athletes. These collegiate participants may have far greater experience in their respective sport than our pre-pubescent population. As such, they may be privy to consistent bilateral repetition where both limbs undergo moments of being the support or operational limb. Pre-pubescent athletes, in contrast, are still building the motor skills to perform sport-specific maneuvers for their preferred limb as opposed to training the non-preferred limb. If this is put into the context of kicking a soccer ball, our post-pubescent population would undergo much more bilateral repetition of sport specific tasks such as passing the ball, whereas pre-pubescent athletes may consistently use their preferred, operational limb. This factor of experience and bilateral repetition could therefore decrease overall limb asymmetry and result in our contradictory findings.

4.5 Perspective

The purpose of the current study was to identify biomechanical asymmetry of the lower limbs during walk, run and side-cut tasks in four healthy high-performance athletic populations across sex and age. It was hypothesized that asymmetry would be detected in all groups, and that pre-pubescent athletes would show the lowest magnitude of bilateral biomechanical asymmetry and lowest proportion of the population exceeding 10% asymmetry. It was also hypothesized that post-pubescent female athletes would show the greatest magnitude of bilateral biomechanical asymmetry and the greatest proportion of the population exceeding 10% asymmetry. Together these hypotheses parallel ACL injury risk as noted in the literature, and thus would allow us to infer that lower limb asymmetry is a risk factor for non-contact ACL injury. No significant interaction effects of sex and age were present in percent asymmetry across all biomechanical variables, both kinematic and kinetic. Post-pubescent females largely displayed the lowest percent asymmetry and proportion of athletes exhibiting greater than 10% asymmetry across the four groups (pre-pubescent males, pre-pubescent females, post-pubescent males, post-pubescent females), although these results were not statistically significant. Age effects were noted for pHFA, across all tasks, HFA_IC in walk and run rasks, as well as, KFA_IC and pKFA in only the side-cut task.

Contrary to expected results, pre-pubescent athletes displayed the greatest percent asymmetry. These findings suggest that increased biomechanical asymmetry may not adversely influence non-contact ACL injury rate and thus may not be an etiological risk factor for ACL injury. These contrary findings may be the result of developed differences through increased experience and bilateral exposure to the side-cut task or through the prior or current participation in ACL prevention programs. The results may also indicate that bilateral asymmetry of the lower limbs may function as a protective mechanism against non-contact ACL injury since all findings are in opposition to expected outcomes from previously established epidemiological figures for non-contact ACL injury. However, this would require extensive investigation. Setting aside biomechanical asymmetry, lower limb dominance or lateralization may still exist for the variables investigated in this study and may still influence non-contact ACL injury risk. With a lack of evidence in the literature regarding the links between biomechanical asymmetry and limb dominance during the side-cut task, future studies should examine these elements in unison and their role in non-contact ACL injury.

In summation, the findings of this study may have important implications on gait evaluations, particularly in clinical and research settings where asymmetry is used as an outcome. Although a large portion of subjects in this study exhibited greater than 10% asymmetry, it is important to note that the clinical relevance of our 10% threshold is not well established for lower limb biomechanics. Rather, the 10% threshold is most often used when defining measures of strength and range of motion or functional movement.[20,116,236] The results of our 95% confidence intervals for pKAM would suggest 79.7% to 99.9% of athletes experience greater than 10% asymmetry during the ballistic side-cut task meant to replicate the ACL injury mechanism. However, overall ACL injury occurs in under one percent for all female athletes including high-level collegiate sport athletes.[276] Therefore, these findings would suggest that a more appropriate level of asymmetry be established as a threshold when examining biomechanics. As it stands, asymmetry exceeding 10% does not appear to be an indicator or precursor of ACL injury.

Chapter 5 – Conclusion

5.1 Summary

The findings of this study support previous studies, such as Lathrop-Lambach, to show that healthy populations exhibit a great magnitude of lower limb asymmetry. Moreover, this is the first study to present such findings in an athletic population, and the first to do so across both sex and age simultaneously. Though the mean value of asymmetry and proportion of athletes exhibiting greater than 10% asymmetry were vastly greater than expected, no interaction effects of sex and age were found indicating asymmetry is not specific to a single age and sex combination. All sex effects found in this study favored males showing greater asymmetry than females, which is contrary to what was expected given ACL injury epidemiology. Likewise, all age effects noted in this study favored pre-pubescent athletes exhibiting greater asymmetry than post-pubescent athletes. This again was in contrast to our expectation given ACL injury rates. Results from this study would indicate that asymmetry is attenuated with age. This could, at least in part, be attributed to bilateral use of limb in sport. Post-pubescent athletes would have additional training for their non-preferred kicking limb, for example, whereas pre-pubescent athletes would still be developing this motor skill for their preferred limb.

With relevance to ACL injury, sex differences during the cutting task were not present for percent asymmetry of internal joint moments of the sagittal or frontal plane. Nor were they present for percent asymmetry of sagittal plane joint angles at initial contact or peak sagittal plane joint angles. Age effects were found for HFA_IC, pHFA, KFA_IC, and pKFA for cut tasks. In all instances of a statistically significant difference for age, pre-pubescent athletes were found to have greater asymmetry than post-pubescent athletes. These results highlight that there are differences between males and females at the knee joint when performing ballistic maneuvers. However, results were contrary, to the expected outcomes as they did not parallel ACL injury rate data.

In summation, the findings of this study may have important implications on gait evaluations, particularly in clinical and research settings where asymmetry is used as an outcome. The high proportion of the healthy population exhibiting greater than 10% asymmetry suggests additional research is required to determine acceptable levels of lower limb kinematic and kinetic asymmetry in a healthy population as well as for return to play criteria. Additional research should also be conducted to further investigate a suitable threshold for biomechanical asymmetry. Lastly, more research is needed to address if lower limb asymmetry is an etiological risk factor for ACL injury.

5.2 Clinical Implications

The proportion of participants exhibiting greater than 10% asymmetry was greater than expected across all variables and for all groups and tasks. At least 27% of our overall population exhibited greater than 10% asymmetry for all biomechanical variables of interest across all tasks. These values suggest that either bilateral asymmetry is far more prevalent than expected in an athletic population, or, it suggests 10% may not be an appropriate threshold when comparing what is symmetrical versus asymmetrical. While this 10% threshold is found elsewhere in the literature, it is generally reported in accordance with other clinically relevant differences in muscle strength and performance based testing[20,116,236] and not biomechanical asymmetry. Future research should be dedicated to determining a more appropriate threshold for biomechanical asymmetry.

In the context of ACL injury, the 10% asymmetry threshold is largely used as a return to play criteria for muscular strength in the affected versus non-affected limb. In the population of athletes used in this study, previous lower extremity injury to the hip, knee or ankle was deemed as exclusion criteria. Therefore, all athletes that took part in this study were ‘healthy’ at the time of testing. If then we assume that these athletes are healthy and have not experienced major injury to the hip, knee or ankle in the past, and experience a great deal of asymmetry, it questions the validity of the 10% threshold for muscular strength for return to play following ACL injury. If otherwise healthy individuals are asymmetrical, do athletes need to be within the 10% threshold to return to play? However, it is also important to note that current return to play protocols infrequently use biomechanical analysis to assess asymmetry before the release of an athlete. Instead a visual examination conducted by the physiotherapist is often used to assess limb symmetry.

Finally, the results of this study largely contradict the results found elsewhere in the literature. Though asymmetry has been linked to the limb dominance theory, which has subsequently been identified as an etiological risk factor for non-contact ACL injury, this study found no evidence to support this claim. Post-pubescent females, whom experience the highest rate of non-contact ACL injury, displayed the lowest degree of asymmetry for many of the biomechanical variables in this study. Pre-pubescent athletes are said to be the least prone to non-contact ACL injury, yet this study showed pre-pubescent athletes to be more asymmetrical than post-pubescent athletes. The combination of these findings would therefore suggest that percent asymmetry and limb dominance should not be considered an etiological risk factor for non-contact ACL injury.

5.3 Limitations

There are limitations to the findings of this study that highlight the need for continued experimentation to validate the reported results. There exist limitations that suggest future additions to the current protocol to ensure for a more complete assessment of ACL injury risk in an athletic population, as well as limitations pertaining to the data used in this study.

5.3.1 Future Additions to Current Laboratory Protocol

Future studies occurring in the John MacIntyre motion Laboratory of Applied Biomechanics (mLAB) should implement recorded visual and verbal instruction to participants for each task. Implementation of this could clarify if the findings of this study truly represent differences in the athletic population or are temporary phenomena observed only after the instructions are given. Though there is a wide variation in the execution of the cutting maneuver between sex and age, demonstrating a need for additional research in this field, regulation of verbal and visual instruction could attenuate variations in deceleration prior to the execution of the cutting maneuvers. It was noted that participants, primarily in the pre-pubescent groups, tended to stutter-step far more often in reaction to the light stimuli and cut direction. This strategy may reflect actual game-situation side-cuts for these groups, or it could be an artifact of additional instructions given to young athletes. In future studies it is recommended that a time to complete requirement (speed) is regulated for participants in the execution phase of the cut rather than just the approach time prior to the cut.

The inclusion criteria for the current study, and many other biomechanical studies for that matter, does not consider the hormonal fluxions associated with the menstrual cycle of females. Given the literature and concern surrounding varying knee joint biomechanics with expression of female sex hormones, more rigorous questioning should be given to female participants. Questions pertaining to contraceptive use, time since last menses, and menses frequency and normalcy should be included in future participant questionnaires (Appendix C). With a large database of participants, this could allow future investigations to examine the effects of hormonal fluxions within female participants. While this data would not include urine assays, it would provide criteria that could make inclusion for the current study more strict or provide investigators with a potential explanation of variation within the post-pubescent female data set.

Additionally, this particular study used only a subset of a much larger experimental protocol as noted in Chapter 3. Participation in the study consisted of a two-to-four hour time commitment that included maximal sprints, a minimum of 30 MVICs, as well as a comprehensive motion analysis protocol that included walk, run, double and single leg drop jumps and unanticipated side-cut trials. Though the unanticipated side-cut trials occurred before the double and single leg drop jump trials, the volume of MVICs and protocol length and inclusion criteria may have induced physical and mental fatigue. In some cases criteria for trial inclusion forced participants to replicate several motion trials, with unanticipated side-cuts being the most commonly excluded trial type. In the most extreme cases, participants did greater than 40 repetitions of the side-cut task to record 8 total trials that met inclusion criteria. Since the effects of physical fatigue on lower limb biomechanics are discussed in detail in Chapter 2, it is worth noting fatigue as a potential limitation or factor in the data collection.

The physical space of the laboratory did not allow for any maximal effort trials. While approach speed was regulated as a percentage of maximal sprint speed, execution of cutting tasks at different velocities would provide insight into whether the findings in this study are phenomena based on solely the task or are based on speed. Thus, we can only speculate how joint kinematics and kinetics would change with increasing speed. It is also unclear if 67% (two-thirds) of maximal sprint speed is representative of the speed at which ligamentous injury occurs in sport. Unfortunately, to the best of this author's knowledge, no literature exists around the speed in which these ligamentous injuries occur most frequently.

5.3.2 Data Limitations

The present study does not use a control (or pre-planned side-cut) when considering the unanticipated side-cut maneuver. Though running is used to assess change in asymmetries in joint kinematics and kinetics, the degree of ‘unanticipation’ can be questioned. Participants may develop strategies during the protocol to more efficiently perform each task. This may decrease the true unanticipated nature of the cut. As such, it is recommended that joint kinematics and kinetics from anticipated cuts be used in the future to ensure participants are not intuitively creating methods of cheating the unanticipated side-cut protocol.

In addition, no neuromuscular data was presented in this present study, thus, the author can only hypothesize how the joint kinematics and kinetics affect dynamic knee stability using knowledge of previously presented neuromuscular activation data. There is opportunity here for future studies to investigate the changes in asymmetry of peak neuromuscular contributions to ballistic sport-specific maneuvers such as side-cut. Results comparing asymmetry for both the maximal neuromuscular activation and joint kinematics and kinetics would paint a more complete picture of potentially injurious knee biomechanics and explain how dynamic knee stability is achieved at the knee joint during ballistic maneuvers.

Furthermore, when comparing asymmetry, there will often be the question of limb dominance. While the concept of limb dominance was important to describe in detail as it serves as an etiological risk factor for ACL injury, it did not factor limb dominance into its calculation of the dependant variables, nor did it include limb dominance as a co-variate in the analysis. The decision to do this was based upon the idea not to directly compare side-to-side differences, but rather asymmetry which is a measure of the ratio of side-to-side difference. This is the first study to examine asymmetry across sex and age during athletic maneuvers, thus, this research team felt it was not imperative to know which limb displayed greater discrete variables, but rather just if asymmetry existed between limbs. Additionally, there remains ambiguity for leg dominance within the literature as it pertains to athletics (Chapter 2.5). The dominant limb is often used synonymously with the ‘preferred’ limb. Early evidence from Peters labelled the dominant limb as the limb used to manipulate an object, while the non-dominant limb was conversely labelled as the non-preferred stabilizing limb.[221] While dominance may be easily described in unilateral tasks it becomes less clear for dynamic tasks where there exists bilateral mobilization, such as kicking a ball. In this case, should the dominant limb be classified as the limb used to strike the ball of the limb used for postural support?

In either case, the concept of lateralization and distinction of operational (kicking) and support (stabilizing) limbs is parallels the concept of limb dominance. For the broader longitudinal study, self-reported limb lateralization was assessed by asking for the kicking leg (Appendix D) which had been used elsewhere in the literature.[141,214] In future studies, and investigation of limb lateralization may provide further understanding on the wide degree of asymmetry of biomechanical variables during sport-specific ballistic maneuvers such as a side-cut.

Lastly, the current study also showed a great deal of variability between participants during execution of walk, run and side-cut tasks as noted by the standard deviations reported in Table 4.1 and Table 4.3. The variability was visually noted by investigators during data collection. Several participants were forced to stutter-step prior to executing a cut or used different strategies to perform the cuts. The research team did exercise quality to address this issue, and control for variance in the cutting maneuver. Trials were removed if participants slowed their forward momentum drastically with stutter-steps and by monitoring approach speed. However, reliability of biomechanical measures is largely under-researched and not commonly cited in the literature.

Our results were in alignment with results from Sankey et al. (2015) which investigated the inter-trial, inter-session and inter-observer reliability of knee loading during side-cut maneuvers.[277] Results of this study suggested that discrete metrics may not be reliable due to high inter trial variability.[277] This group postulated that variations in technique (foot placement and postural control) between participants may affect horizontal forces and thus could elicit lower limb kinetics with high variability.[277] However, they did note that differences in speed of approach did not impact variability.[277] Increased variability is troublesome in that it reduces the ability to detect significant differences between groups when performing statistical tests on group means. Conversely, a more recent publication by Mok et al (2017), investigated within-session and between-session reliability of lower limb discrete biomechanics variables during a side-cut task.[278] Results of this study showed that all discrete variables, including those in the present investigation, showed fair to excellent between-session and within-session reliability implying that these measurements can reliably reproduce the ranking of individuals if tested repeatedly.[278] Therefore, interpretation of these two studies, has led this research team to believe discrete metrics are reliable within a participant, both within-session and between-session but may show high variability when comparing between individuals. This was supported by the current investigation. This presents an interesting challenge for future research which aims discrete variables as high variance may force investigators to alter study design to include an adequate number of participants to detect statistically significant differences between groups or highly variable data, as it most definitely presented itself as a limitation in the current analysis.

Regarding the analysis of discrete biomechanical parameters, gait data is often reported as temporal waveforms which represent specific joint measures throughout the entirety of the gait cycle. As such, description of a single waveform can involve a very large amount of data. A commonly used method for analyzing gait data, and the one chosen for this study, is the definition and extraction of discrete variables which corresponds to specific temporal values (maximums, minimums, ranges etc.).[279] Thus, detection of significant differences between groups was reduced to comparing between subject group averages of these discrete variables. This approach, however, is subjective, and neglects the temporal information which may be a crucial when investigating etiological risk factors for ACL injury. In the case of this study, discrete parameters were extracted from peak values, as well as at the moment of initial contact. While we know the values at initial contact occur at the same moment, temporally, peak values likely occur at different points temporally within the gait cycle. While this study did try to minimize this variance by extracting peak values from the first 50% of stance, they still are likely temporally misaligned.

A commonly used technique in the literature is principal component analysis, a multivariate approach which maintains the temporal and spatial characteristics of the gait pattern.[279,280] A 2005 study by Wrigley et al. (2005) showed that analysis of peak variables was unable to detect differences in low back pain experienced by industrial workers, whereas variables derived from principal component analysis were able to identify biomechanical differences in lifting technique prior to the development of low back pain.[281] Therefore, based on the ability of principal component analysis to identify both temporal and spatial features across gait waveforms that may differ between groups, it should be considered for any future studies.

5.3.3 Limitation of the Asymmetry Calculation

The short comings of the asymmetry calculation used in this study were threefold. The asymmetry calculation was required to: 1) yield only positive integers of asymmetry, and 2) yield values of asymmetry that were not infinite in nature, and 3) provide some clinical significance in its measure. Asymmetry values were required to be positive as that would ensure that asymmetry values would avoid any misrepresentation in post-processing statistical tests. Asymmetry was required to be non-infinite such that extreme outliers were not present in the data set. Finally, clinical relevance was necessary so that equivalent but opposite measures of biomechanical variables were not misrepresented as being equivalent (ex. Adduction and Abduction).

The asymmetry calculation used in this study was adapted from a previously defined limb symmetry index.[95,236]

$$\%\,Asymmetry = 100\ x\ (1 - \frac{lesser\ value}{greater\ value})$$

In this method, the greater average peak or mean value of the dependent variable is forced as the denominator to avoid any misrepresentation of asymmetry within the population by averaging what may otherwise be positive and negative values. While this is intuitive for values of the same sign (positive or negative), the formula experiences shortcomings when values straddle zero. These situations are best explained in the following examples:

Shortcomings

Example 1 – A situation where the biomechanical variable of interest is represented by a negative value for the both limbs.

By definition of our limb symmetry index formula, this situation would force the larger negative value to the numerator regardless of its magnitude. Consider an example of a knee abduction moment during the side-cut task where the value for the right leg is -0.1 but the value of the left limb is -0.15.

$$100\ x\ \left(1-\frac{-0.15}{-0.10}\right) = -50\%\ Asymmetry$$

This presents the first limitation to this formula. If this example is represented as -50% asymmetry it does not overcome the requirement of providing only positive asymmetry values. This method also does not account for the infinite asymmetry requirement. Thus, these shortcomings would suggest using absolute values to eliminate the issue of negative symmetry and infinite asymmetry.

Example 2 – Using absolute values for each biomechanical variable.

Consider the same example from above; a knee abduction moment during the side-cut task where the value for the right leg is -0.1 but the value of the left limb is -0.15.

$$100\ x\ \left(1-\frac{-0.15}{-0.10}\right) = -50\%\ Asymmetry$$

Now consider how this will change if absolute values are present.

$$100\ x\ \left(1-\frac{|0.10|}{|0.15|}\right) = 33\%\ Asymmetry$$

This would force all asymmetry values to range from 0-100%, a rather intuitive result. However, while this method also eliminates the problem of negative asymmetry for post processing statistical tests, it presents an issue of its own. By taking the absolute values, it removes the clinical significance from the value itself. If we consider angles, 10 degrees of knee adduction is significantly different than 15 degrees of knee abduction when we are speaking about knee joint asymmetry. Taking the absolute in this case treats adduction and abduction as one.

Example 3 – Factoring out the negative value and not altering the rank of the greater and lesser variable.

In this method we could factor out the negative sign and insert it on the final asymmetry result. Consider an example where one limb has a knee abduction moment of 0.1 and the contralateral limb has a moment of -0.3 during a side-cut task.

$$100\ x\ \left(1-\frac{-0.30}{0.10}\right)=-200\%\ Asymmetry$$

Now consider if the negative were factored out of the equation.

$$-100\ x\left(1-\frac{0.30}{0.10}\right)=\ 200\%\ Asymmetry$$

In this case, this method seems to address the problem of negative asymmetry, however, it does not hold if the magnitude of negative value is less than the magnitude of the positive value. Consider the same example but switching the signs of the variables (-0.1, 0.3).

$$100\ x\ \left(1-\frac{-0.10}{0.30}\right)=-133\%\ Asymmetry$$

Now consider if the negative were factored out of the equation.

$$-100\ x\left(1-\frac{0.10}{0.30}\right)=\ -167\%\ Asymmetry$$

In both cases, now the requirement of having only positive asymmetry values is not met. It also does not address the requirement of eliminating infinite asymmetry.

Example 4 – The Hybrid Method

It was concluded that this study would use a hybrid calculation that incorporated taking an absolute value, as shown in example two, to determine the rank of the numerator and denominator. The difference in this case is that the negative value is not removed from the calculation. Consider an example where the recorded knee abduction moments during a side-cut task were 0.1 and -0.2 respectively. In this case, *if* the absolute of one limb is greater than the absolute of the other limb, then the greater absolute value is forced as the denominator. In this example that would force -0.2 to the denominator, and 0.1 would become the numerator.

$$100\ x\ \left(1 - \frac{-0.20}{0.10}\right) = \ would\ become\ \ldots$$

$$100\ x\ \left(1 - \frac{0.10}{-0.20}\right) = \ 150\%\ Subject\ Asymmetry$$

In this case, results of asymmetry would range between 0-200%, thus provide only positive asymmetry values within some defined range, making them non-infinite. This method also maintains clinical significance as values ranging from 0-100% asymmetry would indicate the direction of the knee abduction moments used as an example, in this case, were in the same direction, whilst, values of 100-200% asymmetry would indicate the direction of the knee adduction moment between limbs were in opposing direction.

Conclusion:

The hybrid method in example 4 maintained all of the requirements of the research team and thus was selected for use in this study. The hybrid method does, however, create a larger variance within each data set as it allows for a greater range of asymmetry scores. This was particularly relevant in this study for KFA_IC in the walking task, and pKAM in the cutting task. This resulted in greater mean asymmetry values, which may have influenced group statistics when being subject to the ANOVA. A potential method to solve this problem in future investigations would be to normalize the asymmetry value using the 'normalized symmetry index' (Appendix I).[233]

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Appendices

Appendix A: Informed Consent

Date of ethics approval: 2017-05-29

The effect of sex, age and leg-dominance on lower limb biomechanics during athletic maneuvers: Relevance to preventing Anterior Cruciate Ligament (ACL) injuries

The Researchers

You are being invited to participate in a research project being conducted by Dr. Scott Landry and his research students (Jessica Lohnes, Sam Fioretti and Nick DeAdder) at Acadia University's School of Kinesiology. The purpose of this study is to perform a biomechanical comparison between pre- and post-pubescent athletes of their dominant and non-dominant legs during pre-planned and unanticipated athletic maneuvers within the John MacIntyre mLAB (motion Laboratory of Applied Biomechanics). The information obtained from this study may help to reduce future incidences of Anterior Cruciate Ligament (ACL) injuries in the knee. This study is longitudinal in nature, in that you, or your child may be contacted and asked to participate in subsequent years for follow-up testing to analyse changes in biomechanics with age. Participation in follow-up testing is not mandatory. If you have any questions or concerns about the research, the principal investigators' contact information is provided below. If there are any questions or concerns regarding the nature of the project, the contact information of the research ethics board is also provided.

Jessica Lohnes – Principal Investigator
BKin Hons Candidate – Acadia University
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(902) 585-1937 (mLAB)

Dr. Scott Landry – Supervisor
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Nick DeAdder – Principal Investigator
MSc Candidate – Dalhousie University
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(902) 585-1937 (mLAB)

Dr. Stephen Maitzen
Chair, Acadia Research Ethics Board
smaitzen@acadiau.ca
(902) 585-1407

Research Involvement

The questionnaire will have questions concerning your body measurements (weight and height), pubertal status, limb dominance, prior injuries and duration of sport activity. It should take approximately 5-10 minutes to complete and if you meet the inclusion criteria based on your responses to the questionnaire, then you will be asked to complete a series of exercises and athletic maneuvers in the Acadia mLAB (motion Laboratory of Applied Biomechanics). You will be excluded if you have sustained a major injury to your lower extremity which has not fully healed at the date of testing.

This testing protocol will involve walking, running, jumping, and side-cutting maneuvers guided in a pre-planned and an unanticipated manner using a series of timing lights. To track body movement and muscle activity during these athletic maneuvers, you will have small removable muscle activity sensors and reflective motion capture markers taped on your skin over the arms, back, pelvis, thighs, shins, and feet. The muscle sensors will measure muscle activity (on and off), and the reflective markers will allow for a 3D representation of your movements to be recorded by a video system. Force platform sensors in the ground and in the treadmill will also record the forces occurring between the feet and surface during the athletic maneuvers. During the testing session, your leg strength will also be measured using a dynamometer, which is a specialized strength measuring machine.

You will spend a maximum of 2.5 hours at the mLAB for testing, and this will include time for warm-up, instruction and practice of the maneuvers, set up and attachment of the monitoring equipment and execution of the strength measures, along with the walking, running, jumping and side-cutting athletic maneuvers.

Potential Harms

There are minimal risks associated with performing this study, as the side-cutting and jumping maneuvers are movements you are very familiar with as an athlete. Risk of injury is also very minimal, as you will be completing the study in a controlled, indoor setting. You will also be asked to complete a warm up and practice trials to decrease the risk of injury and familiarize yourself with the maneuver and equipment. A slight discomfort may be experienced or redness may become apparent when the tape and sensors are removed due to the shaving of small areas of your skin and removal of sensors. By consenting to participate in this study, you do not waive your right to legal recourse in the event of any research-related harm.

Though our testing does not require participants to exceed regular sporting requirements, the risk of injury is still present. All injuries can be treated on site as one of our researchers has been trained as a student athletic therapist and she will be present for the data collections. We also have full access to the physiotherapy clinic at Acadia University and they are located down the hall from the mLAB.

Potential Benefits

There are no direct benefits to you as a participant, however, your contribution in this research will provide researchers, coaches, and athletes with a better understanding of ACL injury risk factors and injury preventative training programs could be enhanced based on findings from this study.

Confidentiality

As a participant, you will be allotted a number code, which will be used to identify your data after it has been collected. The completed questionnaire, which will be entered into a database on the lab computer, and your data will only be identified by a number code. The questionnaires will be stored in a locked filing cabinet in the security alarmed mLAB. The researchers will not have access to individual participant results in the database until after they have completed trials to remove any experimental bias.

The data will be stored on a password-protected computer, which will remain inside the locked and highly secured Acadia mLAB. Only Dr. Scott Landry and his research students will have access to your individually collected data and any relevant personal information obtained from you in the questionnaire. If the information from the study is published, your individual data will not be identifiable.

Publications
Research students will use the results of this study for the completion of their theses. The results will be presented as group data and may be published in a scholarly peer reviewed journal and presented at several conferences.

Compensation
As a participant, you will receive no financial compensation. If requested by you or your parent and/or guardian, the general findings and conclusions from the study will be made available upon the study's completion via your contact information stored on the password-protected computer. To encourage participation, you will be provided upon arrival with an Acadia Athletics T-shirt and a varsity sport game pass. This compensation will not be withheld if you decide to withdraw from the study at any time prior to completion.

Participation
Your participation in this study is completely voluntary and you may withdraw from the study at any point in time without any penalty or negative repercussions. You may also withdraw your data from the study within 30 days from testing, at which point the data will be used in the primary researchers' honours or masters theses. You are free to ask questions or address any concerns at any point in time either before, during, or after the study.

Consent
By signing the informed consent, I confirm that I have read and understood the nature of my participation. I verify that I have not had any major lower extremity (hip, knee or ankle) injuries in the past six months or previous surgeries to the lower back or lower extremities. I also acknowledge that I am able to withdraw or ask questions at any time during the study without penalty. I am also aware that if I wish to withdraw from the study, I may do so at any time, but to withdraw my previously collected data I must do so within 30 days of data collection.

Photo Release
For photographs and images taken of me by the John MacIntyre mLAB, I grant to the mLAB:
copyright and/or use of photographic representations of myself in various forms of media that will be used by researchers for future theses, posters and other scientific presentations produced by mLAB researchers.

I hereby realize and accept that I am participating on a voluntary basis and will not receive financial compensation from photographs taken by the mLAB researchers. Please check one of the following:

☐ I have read, understand and agree to the above photo release guidelines.

☐ I have read and understand, however, I do not agree to any photos being shared by the researchers in the mLAB.

Participant:

Name: ______________________ Date: ________________
(please print)

Signature: _____________________________

By signing the informed consent I consent to my child's participation in this study and have fully read and understood the nature of their participation. I also confirm that my child has not had any significant lower extremity (back, hip, knee or ankle) injuries in the last six months or previous surgeries to the lower back or lower extremities. I am aware that my child and I are free to ask questions and withdraw at any time during the study without penalty. I am also aware that if my child or I wish to withdraw the data from the study, we must do so within 30 days of data collection.

Guardian (if above participant is under 18 years of age):

Name: ________________Relationship: _____________Date: _______________
(please print)

Signature: _____________________________

Witness: _______________________________

Appendix B: Ethics Approval

DALHOUSIE UNIVERSITY
Research Services

Health Sciences Research Ethics Board
Annual Renewal - Letter of Approval

June 16, 2017

Mr Nicholas DeAdder
Health Professions\Health & Human Performance

Dear Nick,

REB #: 2016-3801
Project Title: The effect of age and leg-dominance on lower limb biomechanics during unanticipated athletic maneuvers: Relevance to preventing Anterior Cruciate Ligament (ACL) injuries

Expiry Date: May 17, 2018

The Health Sciences Research Ethics Board has reviewed your annual report and has approved continuing approval of this project up to the expiry date (above).

REB approval is only effective for up to 12 months (as per TCPS article 6.14) after which the research requires additional review and approval for a subsequent period of up to 12 months. Prior to the expiry of this approval, you are responsible for submitting an annual report to further renew REB approval. When your project is complete and no longer requires REB approval, please complete a Final Report to close your file in good standing. Forms are available on the Research Ethics website.

I am also including a reminder (below) of your other on-going research ethics responsibilities with respect to this research.

Sincerely,

Dr. Tannis Jurgens, Chair

Appendix C: Participant and Collection Information

Collection Information:

*Participant's Name:
*File Extension:
*Collection Date: YY MM DD
*Collector(s):
*St. Dev of Wand Length (Lab):
*Max Residual & Cam #:
*St. Dev of Wand Length (Treadmill):
*Max Residual & Cam #:

*Mass (kg):
*Height (cm):
*Thigh Circumference (cm)
Left: Right:
*Calf Circumference (cm)
Left: Right:
*Foot Width (cm)
Left: Right:

Additional Comments:

*Necessary Paperwork Collected: ☐

Appendix D: Eligibility Questionnaire

Participant Information: **FILE EXTENSION:___________**

Sex: Male	Female				Participant's Mailing Address
Dominant Hand (Writing):	L	/	R		Street or Box #:
Dominant Leg (Kicking):	L	/	R		Town/City:
Dominant Hand (Throwing):	L	/	R		Postal Code:
Date of Birth: YY	MM		DD		Participant's Email:
Age:					Participant's Phone:

Questionnaire:

Would you be interested in being contacted in the future for follow-up testing. Note, future testing will be of a similar nature as your session today (Please Circle):
YES / NO

Playing Experience:

Please list the main sports you participate in, with the sports you participate more frequently listed first.

Sport	Total Years Played	Most Common Position Played (Guard, Midfield, Wing, Receiver)
1.________________________	_______	______________________________
2.________________________	_______	______________________________
3.________________________	_______	______________________________
4.		

What single sport do you consider your **primary sport**? ______________________

Please answer the following questions for your **primary sport** listed above.

Current level of play: __________ (e.g. U10, U12 Tier 2A, U14 Tier 1, Senior Premier, University)

Highest level of play: __________ (e.g. Club, Provincial Team, Canada Games, Regional Training)

If you play university level, what is your current year of eligibility (circle): 1 2 3 4 5

For the current season, answer the following for the primary sport listed above:
Average number of games played per week:
Fall/Winter_________ Spring/Summer _________
Average number of training sessions per week:
Fall/Winter_________ Spring/Summer _________
Average length of training session in hours:
Fall/Winter_________ Spring/Summer _________

Other sports currently playing (this year): _________________________
Average number of games played per week:
Fall/Winter_________ Spring/Summer _________
Average number of training sessions per week:
Fall/Winter_________ Spring/Summer _________
Average length of training session in hours:
Fall/Winter_________ Spring/Summer _________

Other sports currently playing (this year): _________________________
Average number of games played per week:
Fall/Winter_________ Spring/Summer _________
Average number of training sessions per week:
Fall/Winter_________ Spring/Summer _________
Average length of training session in hours:
Fall/Winter_________ Spring/Summer _________

Injury History:
Have you previously had any type of surgery on your lower extremities (e.g. hip, knee or ankle/foot) or lower back? Yes / No

If yes to the above question on surgery, when and what type of surgery was performed (Year/Month)?

1. ____

2. ____

Have you had any significant injuries in the past 6 months to the lower extremities (e.g. hip, knee or ankle/foot) or lower back? Yes / No

If yes to the above question, explain the diagnosis of the injury and date that the injury occurred (Year/ Month).

1. ____

2. ____

3. ____

If yes to the above question on injuries, indicate what health professional diagnosed the injury (e.g. doctor, physiotherapist, athletic therapist, etc.)

1. ____

2. ____

3. ____

If yes to the above question on injuries, explain how many weeks you were away from activity.

1. ____

2. ____

3. ____

Are you currently experiencing any injuries to the lower extremity or back?
Yes / No
Are you currently experiencing any lower extremity or back pain?
Yes / No

If yes to either of the above two questions, explain the diagnosis of the injury or pain, the date that it occurred (Year/Month) and if it is preventing you from participation (games/practice)?

1. ____

2. ____

3. ____

Have you, or are you currently practicing an injury prevention program (e.g. FIFA 11+) ?
Yes / No
If yes, please state the programs name to the best of your ability.
____
If not practicing currently, when did you practice this injury prevention program?
____

Appendix E: Pubertal Questionnaire

The next questions are about changes that may be happening to your body. These changes normally happen to different young people at different ages. Please do your best to answer carefully. If you do not understand a question, please ask one of the researchers or one of your parents/guardians.

Question 1
Would you say that you have experienced a growth in height:

- ☐ Not yet started (1 point)
- ☐ Barely started (2 point)
- ☐ Definitely started (3 point)
- ☐ Seems complete (4 point)
- ☐ I don't know

Question 2
What would you say about the growth of your body hair? ("Body hair" means hair in places other than your head, such as under your arms)

- ☐ Not yet started (1 point)
- ☐ Barely started (2 point)
- ☐ Definitely started (3 point)
- ☐ Seems complete (4 point)
- ☐ I don't know

Question 3
Have you noticed any skin changes, especially pimples?

- ☐ Not yet started (1 point)
- ☐ Barely started (2 point)
- ☐ Definitely started (3 point)
- ☐ Seems complete (4 point)
- ☐ I don't know

FOR BOYS ONLY

Question 4
Have you noticed a deepening of your voice?

- ☐ Not yet started (1 point)
- ☐ Barely started (2 point)
- ☐ Definitely started (3 point)
- ☐ Seems complete (4 point)
- ☐ I don't know

PAGE 1 of 2
Go to page 2 →

Question 5
Have you begun to grow hair on your face?

- ☐ Not yet started (1 point)
- ☐ Barely started (2 point)
- ☐ Definitely started (3 point)
- ☐ Seems complete (4 point)
- ☐ I don't know

FOR GIRLS ONLY

Question 6
Have you noticed that your breasts have begun to grow?

- ☐ Not yet started (1 point)
- ☐ Barely started (2 point)
- ☐ Definitely started (3 point)
- ☐ Seems complete (4 point)
- ☐ I don't know

Question 7a
Have you begun to menstruate (started to have your period)?

- ☐ Yes
- ☐ No

Question 7b
If yes, how old were you when you started to menstruate?
__________ years old
****
FOR RESEARCHERS USE ONLY
MALES Total Sum of Questions **2, 4 & 5**: _______________
FEMALES Total Sum of Questions **2 & 6**: ______________
Pubertal Stage: ____________________

Stage	Male	Category #	Female	Category #
Prepubertal	3	0	3	5
Early pubertal	4 or 5 (No 3-point responses)	1	3 and no menarche	6
Midpubertal	6 - 8 (no 4-point responses)	2	>3 and no menarche	7
Late Pubertal	9-11	3	≤ 7 and menarche	8
Postpubertal	12	4	8 and menarche	9

Appendix F: Data Collection Sheet

MVCs (Biodex): *** Note: all trials are 3 seconds with at least 30 seconds rest between trials**

Torque Scaling: __________ **EMGBias Trial: ______________________**

Trial Name	**Trial Ext.**	**Leg**	**Arm Length**	**Muscle Group**	**Exercise**
MVC	0001	Right		Quadriceps	Gravity Correct
MVC	0002				Extension (45°)
MVC	0003				
MVC	0004	Right		Quad (Rec Fem)	Knee Ext & Hip Flex (45°)
MVC	0005				
MVC	0006	Right		Hamstrings	Gravity Correct
MVC	0007				Flexion (45°)
MVC	0008				
MVC	0009	Right		Gastrocnemii	Plantarflexion (90°)
MVC	0010				
MVC	0011	Left		Gastrocnemii	Plantarflexion (90°)
MVC	0012				
MVC	0013	Left		Quadriceps	Gravity Correct
MVC	0014				Extension (45°)
MVC	0015				
MVC	0016	Left		Quad (Rec Fem)	Knee Ext & Hip Flex (45°)
MVC	0017				
MVC	0018	Left		Hamstrings	Gravity Correct
MVC	0019				Flexion (45°)
MVC	0020				
MVC	0021	Right		Gluteus Medius	Gravity Correct
MVC	0022				Laying Adduction
MVC	0023				
MVC	0024	Left		Gluteus Medius	Gravity Correct
MVC	0025				Laying Abduction
MVC	0026				

MVC (Bed): **Note: all trials are 3 seconds with at least 30 seconds rest between trials**

Trial Name	**Trial Ext.**	**Leg**	**Muscle Group**	**Exercise**
MVC	0027	Right	Gastrocnemeus	Standing Plantarflexion
MVC	0028			
MVC	0029	Left	Gastrocnemeus	Standing Plantarflexion
MVC	0030			

Walk Times: _________/_________/_________/_________

Walk Window: _________-_________

Sprint Times: _________/_________

Sprint Window:_________-______ **Smartspeed Gate Position -_________**

Calibration and Fundamental Movement Protocol:

	Trial Name			Foot on FP		Notes
5s	StandCal					
60s	MoveCal					
90s	HJC					
20s	S_Balance_R			R		
	S_Balance_L			L		
20s	D_Squat			Both		
20s	Lunge_R			R		
20s	Lunge_L			L		

Walking/Running Protocol: **Walk: ________________ Run: __________________**

	Trial Name	Counter	Split 1	Foot on FP	Notes
10s	Walk (1)	00_ _			
	Walk (2)	00_ _			
	Walk (3)	00_ _			
	Walk (4)	00_ _			
	Walk (5)	00_ _			
	Walk (6)	00_ _			
	Walk (7)	00_ _			Avg Walk Vel (L)
	Walk (8)	00_ _			Avg Walk Vel (R)
10s	Run (1)	00_ _			
	Run (2)	00_ _			
	Run (3)	00_ _			
	Run (4)	00_ _			
	Run (5)	00_ _			
	Run (6)	00_ _			
	Run (7)	00_ _			Avg Run Vel (R)
	Run (8)	00_ _			Avg Run Vel (L)

Cutting Protocol:

	Trial Name	Counter	Split 1	Foot on FP	Split 2	Notes
10s	PPC_R	00_ _				
	PPC_L	00_ _				
	PPC_R	00_ _				
	PPC_L	00_ _				
	PPC_R	00_ _				
	PPC_L	00_ _				
	PPC_R	00_ _				Avg PPCut R Vel
	PPC_L	00_ _				Avg PPCut L Vel
10s	Cut (1)	00_ _				
	Cut (2)	00_ _				
	Cut (3)	00_ _				
	Cut (4)	00_ _				
	Cut (5)	00_ _				
	Cut (6)	00_ _				
	Cut (7)	00_ _				Avg Cut R Vel
	Cut (8)	00_ _				Avg Cut L Vel

Jumping Protocol:

	Trial Name	Counter		Foot on FP		Notes
10s	D_Jump (1)	00_ _		Both		
	D_Jump (2)	00_ _		Both		
	D_Jump (3)	00_ _		Both		
	D_Jump (4)	00_ _		Both		
10s	S_Jump R	00_ _		R		
	S_Jump_L	00_ _		L		
	S_Jump_R	00_ _		R		
	S_Jump_L	00_ _		L		
	S_Jump_R	00_ _		R		
	S_Jump_L	00_ _		L		
	S_Jump_R	00_ _		R		
	S_Jump_L	00_ _		L		

Performance Protocol:

	Trial Name	Counter	Split 1	Foot on FP	Split 2	Total Time	Notes
15s	Start_R	0001					
15s	Start_L	0001					
15s	Plant_R	0001		R			
15s	Plant_L	0001		L			

Appendix G: Retro-reflective Marker List and EMG Setup

75 including virtual markers; 65 for motion trials.

Trunk and Pelvis (12 total)
Anterior Superior Iliac Spine (ASIS) (2)
Posterior Superior Iliac Spine (PSIS) (2)
Iliac Crest (2)
Sternum - Jugular Notch (1)
Sternum - Xiphoid Process (1)
Acromion Process (2)
T2 Spinous Process (1)
Inferior Angle of Scapula (Midline) (1)

Arms (12 total)
Medial Epicondyle of Humerus (2)
Lateral Epicondyle of Humerus (2)
Ulnar Styloid Process (2)
Radial Styloid Process (2)
Posterior triceps (2)
50% of the radial lateral aspect of forearm (2)

Thighs (6 total)
Medial Epicondyle of Femur (2)
Lateral Epicondyle of Femur (2)
Greater Trochanter (2)

Shank (4 total)
Tibial Tuberosity (2)
Fibular Head (2)

Feet (14 total)
1st Distal Hallux (2)
1st Metatarsal Head (2)
2nd Metatarsal Head (2)
5th Metatarsal Head (2)
Medial Calcaneus (2)
Lateral Malleolus (2)
Medial Malleolus (2)

Clusters (27 total)
Headband (5 markers)
Thigh Cluster (4 markers) x 2
Shank Cluster (4 markers) x 2
Heel Triad (3 markers) x 2

Removable Markers (Following Standing Calibration) (10 total)
Medial Malleolus (2)
Medial Epicondyle of Femur (2)
Medial Epicondyle of Humerus (2)
2nd Metatarsal Head (2)
Medial Calcaneus (2)

Electrode Placements:

Lateral Gastroc (LG) – 30% of the distance from the knee’s lateral joint line to the calcaneus.
Medial Gastroc (MG) – 35% of the distance from the knee’s medial joint line to the calcaneus.
Lateral Hamstring (LH) – 50% of the distance from the knee’s lateral joint line to the ischial tuberosity.
Medial Hamstring (MH) – 50% of the distance from the knee’s medial joint line to the ischial tuberosity.
Vastus Lateralis (VL) – 33% of the distance from the knee’s lateral joint line to the ASIS.
Vastus Medialis (VM) – 20% of the distance from the knee’s medial joint line to the ASIS.
Rectus Femoris (RF) – 50% of the distance between the superior part of the patella to the ASIS.
Gluteus Medius (GM) – 50% of the distance from the iliac spine to the greater trochanter.

Appendix H: Selection of Participants from Database

Initial participant pool: 203 participants

Initial Participant Pool
N = 203

↓

Subtraction of all duplicate participants (13)

↓

Remaining Participant Pool
N = 190

↓

Subtraction of all injury exclusions (6)

↓

Remaining Participant Pool
N = 184

↓

Subtraction of all participants ages 13-16 (43)

↓

Remaining Participant Pool
N = 141

↓

Subtraction of participants with incomplete data sets (19)

↓

Remaining Participant Pool
N = 122

Appendix I: Calculation of % Asymmetry

Example: KFA_IC (Walking Task):

Ensemble Average Values:
Limb 1: -0.94235802
Limb 2: 0.35104200

Max/Min values across all individual trials (4):
Max: 1.4645
Min : 0.3510

Lathrop-Lambach et al. (2014): Absolute asymmetry[95]

$$100\ x\ \left(1-\frac{|0.35104200|}{|0.94235802|}\right)=\mathbf{62.75\%\ \mathit{Asymmetry}}$$

DeAdder et al. (2020): Absolute asymmetry with clinical relevance

$$100\ x\ \left(1-\frac{0.35104200}{-0.94235802}\right)=\ \mathbf{137.25\%\ \mathit{Asymmetry}}$$

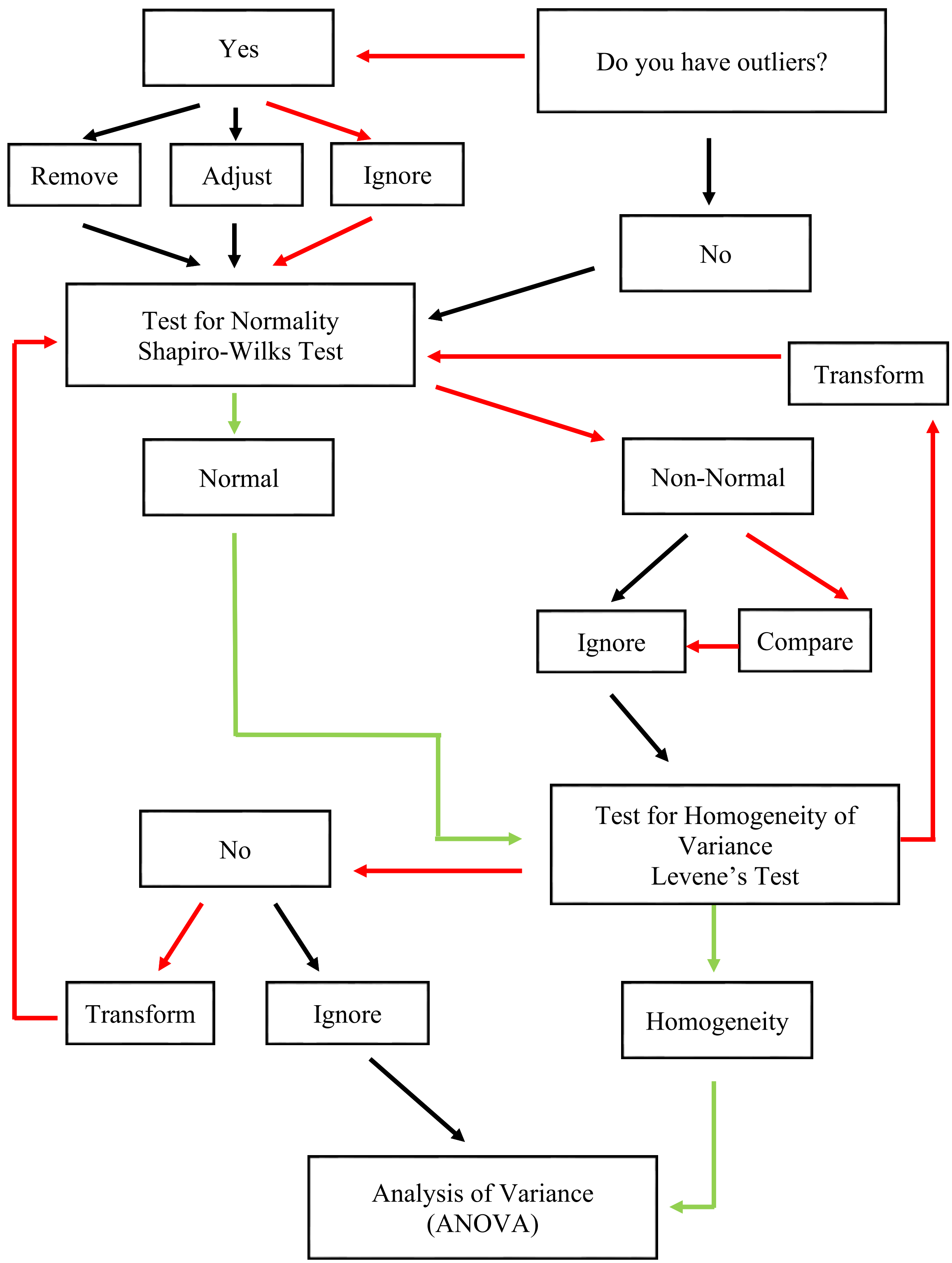
Queen et al. (2020): Normalized Symmetry Index (NSI)[233]

$$NSI=\frac{Greater\ Limb-Lesser\ Limb}{(\max(greater_t\ ,\ lesser_t))-(\min\ (greater_t\ ,lesser_t))}\ x\ 100$$

$$NSI=\frac{0.94235802-0.351042}{(1.4645)-(0.3510)}\ x\ 100=\mathbf{53.10\%\ \mathit{Asymmetry}}$$

Appendix J: Flow Chart of Statistical Tests

Note: Red arrows distinguish initial decision path. Green lines distinguish final path. Black lines are the path not taken.



Appendix K: Results of Normality and Homogeneity of Variance Assumption Tests

Table 6.1. Results of Shapiro-Wilks test for normality of peak sagittal plane joint angles and sagittal plane joint angles at initial contact for all groups and across all tasks

		Untransformed				Transformed			
Variable	Task	Pre-Male	Post-Male	Pre-Female	Post-Female	Pre-Male	Post-Male	Pre-Female	Post-Female
pHFA	Walk	0.001*	0.056	0.181	0.000*	0.253	0.788	0.986	0.263
	Run	0.021*	0.000*	0.038*	0.000*	0.908	0.293	0.222	0.917
	Cut	0.251	0.000*	0.003*	0.029*	0.187	0.110	0.369	0.968
HFA_IC	Walk	0.000*	0.05*	0.322	0.000*	0.188	0.400	0.541	0.208
	Run	0.302	0.00*	0.358	0.001*	0.641	0.837	0.085	0.661
	Cut	0.266	0.000*	0.004*	0.001*	0.827	0.597	0.240	0.237
pKFA	Walk	0.312	0.000*	0.000*	0.000*	0.935	0.158	0.000*	0.192
	Run	0.012*	0.350	0.062	0.028*	0.438	0.985	0.801	0.531
	Cut	0.000*	0.004*	0.375	0.007*	0.499	0.986	0.960	0.841
KFA_IC	Walk	0.043*	0.062	0.117	0.001*	0.336	0.744	0.424	0.307
	Run	0.012*	0.083	0.064	0.000*	0.725	0.573	0.578	0.033*
	Cut	0.239	0.252	0.021*	0.096	0.572	0.969	0.747	0.762

*** Indicates non-normal finding and a violation of the normality assumption (p<0.05).**

Table 6.2. Results of Shapiro-Wilks test for normality of peak sagittal and frontal plane joint moments for all groups and across all tasks

		Untransformed				Transformed			
Variable	Task	Pre-Male	Post-Male	Pre-Female	Post-Female	Pre-Male	Post-Male	Pre-Female	Post-Female
	Walk	0.047*	0.022*	0.006*	0.005*	0.739	0.241	0.187	0.729
pHEM	Run	0.056	0.006*	0.161	0.00*	0.982	0.428	0.698	0.088
	Cut	0.001*	0.005*	0.001*	0.016*	0.103	0.895	0.384	0.959
	Walk	0.006*	0.031*	0.399	0.079	0.194	0.055	0.881	0.857
pHAM	Run	0.029*	0.013*	0.029*	0.013*	0.581	0.308	0.888	0.244
	Cut	0.828	0.197	0.509	0.004*	0.785	0.656	0.205	0.305
	Walk	0.028*	0.009*	0.045*	0.001*	0.972	0.232	0.881	0.555
pKEM	Run	0.002*	0.111	0.182	0.007*	0.551	0.275	0.607	0.649
	Cut	0.000*	0.029*	0.607	0.003*	0.046*	0.309	0.979	0.378
	Walk	0.107	0.014*	0.386	0.024*	0.560	0.968	0.425	0.363
pKAM	Run	0.053	0.116	0.230	0.010*	0.867	0.155	0.134	0.099
	Cut	0.344	0.165	0.186	0.876	0.368	0.846	0.743	0.564

*** Indicates non-normal finding and a violation of the normality assumption (p<0.05).**

Table 6.3. Results of Levene's test for homogeneity of variance of peak sagittal plane joint angles and sagittal plane joint angles at initial contact.

Variable	Task	**Untransformed**	**Transformed**
pHFA	Walk	0.256	0.823
	Run	0.508	0.986
	Cut	0.023*	0.065
HFA_IC	Walk	0.023*	0.113
	Run	0.703	0.809
	Cut	0.001*	0.051
pKFA	Walk	0.607	0.312
	Run	0.042*	0.288
	Cut	0.013*	0.061
KFA_IC	Walk	0.076	0.443
	Run	0.275	0.626
	Cut	0.001*	0.009*

*** Indicates a statistically significant finding, or a violation of homogeneity of variance assumption ($p<0.05$).**

Table 6.4. Results of Levene's test for homogeneity of variance of peak sagittal and frontal plane joint moments.

Variable	Task	Untransformed	Transformed
pHEM	Walk	0.396	0.234
	Run	0.830	0.999
	Cut	0.025*	0.295
pHAM	Walk	0.040*	0.442
	Run	0.298	0.200
	Cut	0.107	0.094
pKEM	Walk	0.068	0.068
	Run	0.019*	0.275
	Cut	0.012*	0.002*
pKAM	Walk	0.457	0.716
	Run	0.765	0.846
	Cut	0.132	0.355

*** Indicates a statistically significant finding, or a violation of homogeneity of variance assumption ($p<0.05$).**