Anticipatory and Reactionary EMG and Knee Joint Kinematics Between Male and Female Athletes during Single-Leg Landing Tasks: Relevance to Non-Contact ACL Injuries

by

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Abstract

Background: Females are more likely to suffer anterior cruciate ligament (ACL) injuries. Muscle activity may be responsible in preventing ACL tears during jump-landings. **Purpose:** To determine a) whether sex differences exist in knee joint motion and muscle activity during preparatory and landing phases of single leg drop-jumps (SLJ) and b) whether an association between preparatory muscle activity and sex and landing knee joint motion exist. **Methods:** 33 male and 21 female athletes were recruited. Standardized biomechanical and electromyography procedures were used to record joint motion and muscle activity during the SLJ. **Results:** No sex differences in knee motion and muscle activity were found in preparatory and landing phases ($p>0.05$). Sex was not associated with knee motion in landing phases ($p>0.05$). Preparatory rectus femoris and medial hamstring muscle activity was associated with knee motion during landing phases ($p<0.05$). **Conclusion:** SLJ mechanics do not explain female injury risk in this athletic population.

LIST OF ABBREVIATIONS USED

ACL – Anterior Cruciate Ligament

ANOVA – Analysis of Variance

CCI – Co-contraction Index

EMG – Electromyography

GRF – Ground Reaction Force

IC – Initial Contact

LH – Lateral Hamstring

MH – Medial Hamstring

MVIC – Maximum Voluntary Isometric Contraction

Q:H – Quadriceps:Hamstring

RF – Rectus Femoris

ROM – Range of Motion

SD – Standard Deviation

SLJ – Single Leg Drop Jump

VL – Vastus Lateralis

VM – Vastus Medialis

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Chapter One: Introduction

1.1 Introduction

Anterior cruciate ligament (ACL) injuries are a common but devastating sports injury, with the rates of ACL injuries appearing to still be increasing (Buller et al., 2015). ACL injuries are typically classified as having a contact or non-contact mechanism, with the non-contact ACL mechanism occurring 70% of the time (Boden et al., 2000; Stuelcken et al., 2016). Non-contact ACL injuries typically occur during athletic manoeuvres such as landing from a jump or cutting whereby the athlete must decelerate or change directions quickly (Boden et al., 2000; Stuelcken et al., 2016). Individuals who have ruptured their ACL suffer both short- and long-term consequences, such as decreased knee function scores, quality of life (Daniel et al., 1994) and an increased risk of developing early onset osteoarthritis (Lohmander et al., 2004, 2007; Webster & Hewett, 2022).

It is difficult to ascertain the number of ACL injuries in a given country, though it is estimated in the US that around 69 out of 100,000 individuals will sustain an ACL injury each year (Sanders et al., 2016), with approximately 135,000 individuals undergoing ACL reconstruction surgery (Buller et al., 2015). Sweden, one of the only countries that currently has a national web-tracking database on the number of ACL reconstruction surgeries performed per year, has around 80 out of 100,000 individuals undergoing ACL reconstruction surgery per year (Nordenvall et al., 2012). ACL injuries most commonly occur in individuals between the ages of 21 to 30 (Nordenvall et al., 2012), though Sanders et al. (2016) found that females were more likely to tear their ACL between the age of 14 to 18 while the age range for males was between 19 to 25. The young adult sporting population is most susceptible to ACL injury and differentiating sex-related factors is imperative to thwart continued dysfunction.

Numerous studies have looked at both biomechanical and neuromuscular factors to determine the mechanism of ACL injury as well as what factors are associated with increased risk of the non-contact mechanism. Ireland (1999) proposed a term to describe the position that results in an ACL rupture, which she called the “position of no return”. The high risk “position of no return” that results in non-contact ACL injury is characterized as being an out-of-control landing, on one foot, as well as with less knee flexion (sagittal plane) and greater knee valgus (frontal plane) (Ireland, 1999; Koga et al., 2010). Furthermore, hip abductors and extensors are unable to sufficiently control trunk movement, which indirectly increases the strain sustained at the ACL (Ireland, 1999). Research focusing on video-analysis as well as patient reports supports this description of a non-contact ACL injury mechanism (Boden et al., 2000; Kobayashi et al., 2010; Koga et al., 2010; Krosshaug et al., 2007; Stuelcken et al., 2016). Previous research has noted that sex differences exist in the performance of dynamic manoeuvres. This could be an explanation as to why there is a sex difference in rates of non-contact ACL injury, with females being 2 to 4 times more likely to sustain an ACL injury compared to males participating in the same sport (Arendt et al., 1999; Montalvo et al., 2019; Prodromos et al., 2007).

Research into ACL injury mechanisms has focused on dynamic movements, as they mimic in-game and in-practice movements and are more likely to place the knee in the “position of no return”, which could lead to an ACL rupture. There have been numerous theories developed to improve our understanding of male/female differences during dynamic movements and how it might be relevant to ACL injuries. One such theory is the quadriceps dominant theory, and the second theory is the ligament dominant theory, with both theories being related to one another (Hewett et al., 2010). The quadriceps dominant theory states that females predominantly utilize their quadriceps to help stabilize the knee joint during dynamic movements, while the ligament dominant theory supports that the ligament and other inert tissues of the knee joint help stabilize the knee during movement due to the ineffective motor control of the posterior chain muscles of the lower extremity (i.e., hamstrings, gluteal muscle groups, and the gastrocnemius and soleus). In such an instance, the quadriceps are antagonists of the ACL, which results in an increase in the anterior shear force on the ACL. Furthermore, higher quadriceps activation can lead to a more extended knee during landing from a jump, which results in greater forces travelling, across the knee. Together, this could result in an ACL injury. Some studies have supported these theories, with females displaying a greater quadricep muscle group activation (Hughes & Dally, 2015; Urabe et al., 2005) and significantly less hamstring muscles activity (Hughes & Dally, 2015) to stabilize the knee joint and absorb the forces. Males were more likely to use a synergistic muscle pattern where hamstring/quadriceps co-contractions help absorb the forces that travel though the knee when landing from a jump (Nagano et al., 2007; Urabe et al., 2005). This predominant use of the quadricep muscles to stabilize the knee joint results in females tending to land with straighter, stiffer knees in the sagittal plane (Schmitz et al., 2007; Weinhandl et al., 2010) and greater knee abduction angle compared to males (Ford et al., 2003; Pappas et al., 2007; Russell et al., 2006). Furthermore, research suggests that females are less likely to co-contract their medial thigh muscles compared to their male counterparts (Palmieri-Smith et al., 2008, 2009). This selective activation of the lateral musculature and decreased activation of medial musculature could place females in positions of higher risk to sustain an ACL injury (Palmieri-Smith et al., 2008, 2009). This sex difference in biomechanical and muscle activity could explain why females are more likely to sustain injury and by understanding these differences, sex specific preventive programs may be established to help reduce the risks. Plenty of studies have been conducted on these theories, though a recent systemic review has concluded that no consensus can be drawn to support or disprove these theories based on current studies due to differences in methods and inconsistencies (Otsuki et al., 2021). It should be noted that in some of these studies, the participants were recreationally active. Prior research has shown that in collegiate athletes, males and females have similar knee joint motion and muscle activity when landing on a single leg (Fagenbaum & Darling, 2003). Furthermore, female collegiate athletes moved differently compared to recreational athletes during a single leg drop jump landing task (Morishige et al., 2019). Prior athletic training may have an influence on the sex specific biomechanics and muscle activity that play a role into the sex differences in ACL injury risk, and in this specific population, it may not appear.

Besides looking at the biomechanical and neuromuscular analysis during dynamic manoeuvres, studies have looked at how preparatory muscle activity assists with maintaining joint stability and preventing an ACL injury (Riemann & Lephart, 2002; Wikstrom et al., 2006). As ACL injuries are thought to typically occur within the first 40ms of initial foot contact with the ground (Krosshaug et al., 2007), preventing an ACL injury via feedback mechanisms for neuromuscular control is not possible. This is because time for the cerebral cortex to develop a motor response due to a change in the environment (typically greater than 120ms) (Williams et al., 2001) and the electromechanical delay (i.e. time between onset of muscle excitation and development of tension in the muscle) (Cavanagh & Komi, 1979) exceeds 40ms, which is when an ACL injury will typically occur. Therefore, preparatory muscle activity may play a role in preventing ACL injuries, with studies showing that there is an association between preparatory muscle activity and anterior tibial shear force and peak knee flexion angles (Brown et al., 2014) as well as peak knee valgus angles (Palmieri-Smith et al., 2008). Furthermore, previous research has also shown that there is a sex difference in how males and females prepare before landing. Nagano et al. (2007) found that females had greater rectus femoris muscle activity but males had a greater Quadriceps:Hamstring (Q:H) ratio 50ms prior to landing from a jump, and these results echo the findings from Zazulak et al. (2005). As these factors are associated with an increased risk of sustaining an ACL injury (Leppänen, Pasanen, Kujala, et al., 2017), it is important to understand how preparatory muscle activity influences knee motion in this plane. Since the mechanism of injury is thought to be multi-planar, more research needs to be done to understand the association between preparatory muscle activity and knee motion during dynamic manoeuvres.

Preparatory and reactionary muscle activity as well as knee joint ROM differs between males and females when a dynamic manoeuvre is performed (Ford et al., 2003; Hughes & Dally, 2015; Nagano et al., 2007; Pappas et al., 2007; Russell et al., 2006; Schmitz et al., 2007; Weinhandl et al., 2010; Zazulak et al., 2005). Females tend to have greater preparatory quadriceps muscle activity (Nagano et al., 2007; Zazulak et al., 2005), smaller knee flexion angles (Leppänen, Pasanen, Kujala, et al., 2017; Schmitz et al., 2007; Weinhandl et al., 2010) and larger knee abduction angles (Hewett et al., 2005), which have been identified as biomechanical and neuromuscular risk factors associated non-contact ACL injuries. However, research studying the sex specific relationships between preparatory muscle activity and knee joint ROM during the dynamic movement is still not clearly understood, particularly in a group of male and female athletes engaged in sport and training at a similar level, and more research is needed.

The primary objective of this thesis is to determine the difference in preparatory and reactionary muscle activity as well as the relationship between preparatory muscle activity and knee joint range of motion (ROM) during the landing phase of a single leg drop jump (SLJ) between post-pubescent male and female athletes who engage in a similar level of training and are field or court athletes.

1.2 Objectives

The primary objective of this thesis is divided into two objectives:

1. Determine whether there is a difference in quadriceps and hamstring muscle activity, medial and lateral Q:H co-contractions as well as knee joint ROM during the preparatory and landing phase of a SLJ between male and female participants.
2. Determine whether there is an association between preparatory muscle activity and knee joint ROM during the landing phase and whether sex explains significant variance in this relationship.

1.2.1 Hypotheses for Objectives

The specific alternative hypotheses for the first objective are:

1. Female participants will have smaller sagittal plane ROM at the knee joint compared to male participants (Schmitz et al., 2007; Weinhandl et al., 2010).

2. Females will have greater quadriceps muscle activity compared to males during the preparatory phase (Nagano et al., 2007; Zazulak et al., 2005).
3. Female participants will have lower muscle activity amplitudes of the hamstring compared to male participants during the landing phase (Ebben et al., 2010; Hughes & Dally, 2015; Urabe et al., 2005).
4. Female participants will have smaller medial Q:H co-contractions compared to male participants during the landing phase (Palmieri-Smith et al., 2009).

For the second objective, the hypothesis is:

1. Sex, preparatory quadriceps and hamstring muscle activity, and Q:H co-contractions are predictive of sagittal plane knee ROM during the landing phase.

Chapter Two: Literature Review

This literature review outlines why it is important to study the effect that sex has on biomechanics and muscle activity during a single leg jump landing task and the relevance to non-contact ACL injury. First the negative outcomes associated with an ACL injury as well as the mechanism of injury for an ACL will be reviewed. This will include what knee joint stability is, how the sensorimotor system maintains knee joint stability during movement and what factors place a significant load on the ACL, increasing the risk of an ACL injury. Sex-based risk factors associated with ACL injuries will then be discussed prior to reviewing the different tools used to understand biomechanical and neuromuscular risk factors associated with ACL injuries.

2.1 Burden of ACL Injury

Over the past number of years, the number of individuals participating in sports has increased (National Collegiate Athletic Association, 2021), with the number of ACL injuries being reported reflecting this increase. Prior to 2015, a yearly estimate of 200,000 individuals sustained an ACL injury in the US yearly (Buller et al., 2015), with a significant portion being individuals between the ages of 14 to 25 (Sanders et al., 2016). 70% of ACL injuries are considered to be non-contact and typically occur during sudden decelerations, landing from a jump or pivoting on a planted foot (Boden et al., 2000; Stuelcken et al., 2016), with females being more likely to sustain an ACL injury compared to their male counterparts (Arendt et al., 1999; Montalvo et al., 2019; Prodromos et al., 2007). Therefore, research has focused on understanding how males and females move differently from one another in trying to explain the sex difference in ACL injury risk during dynamic manoeuvres.

ACL injuries are debilitating and are associated with both short- and long-term effects. In the short term, the substantial financial burden by undergoing ACL reconstruction surgery and/or rehabilitation is considerable, with estimates for surgery and rehabilitation totaling over $14,000 US dollars per person in 2013 (Herzog et al., 2017). The time spent absent from sports is significant, with a typical rehabilitation time frame spanning 9 to 12 months (Kaplan & Witvrouw, 2019; Paterno et al., 2014), though the time frame of more than 12 months has been suggested for individuals below the ages of 20 (Nagelli & Hewett, 2017). Getting back to sport can also be difficult. The psychological impact that ACL injuries may have on the individual can reduce the rate of individuals who return to sport post-ACL injury (Ardern et al., 2014).

In the longer term, those sustaining an ACL injury are at an increased risk of developing early-onset osteoarthritis (OA), with research claiming that individuals who have sustained an ACL rupture have an increase in odds by 6.8 in developing OA (Lohmander et al., 2004, 2007; Webster & Hewett, 2022). The chance of developing OA does not change if an ACLR surgery is performed or not (Lohmander et al., 2004; Webster & Hewett, 2022). Furthermore, most individuals who sustain an ACL injury will experience a decrease in quality of life within 15 years from injury, as seen with lower Knee Injury and Osteoarthritis Outcome Score (KOOS) scores, with the largest impact being noted on the knee-related quality of life and sports or recreation function sub-scales (Daniel et al., 1994; Lohmander et al., 2004, 2007). Lower KOOS scores on the different subscales has been correlated with decreased vertical hop height and Test for Substitution Patterns scores (Flosadottir et al., 2016). The Test for Substitution Pattern assesses postural orientation while the vertical hop height is used to assess hop performance, and both can be uses to assess muscle function. This suggests that worse muscle function after an ACL injury is related to lower KOOS sub-scale scores.

Individuals who have previously torn their ACL are also at a higher risk of sustaining a secondary ACL injury compared to individuals who have never torn their ACL (Paterno et al., 2014). Research suggests that one's movement pattern and neuromuscular control is altered after an ACL injury (Goerger et al., 2015; Johnston et al., 2018; Ortiz et al., 2008). This altered neuromuscular control and movement pattern appears to predispose these individuals to further ACL injury.

Due to the many negative consequences associated with ACL injuries, preventing an ACL rupture is key. To prevent an ACL injury from occurring, we must first understand how an ACL injury occurs and what factors are associated that can be modified.

2.2 Mechanism of Injury of the ACL

ACL ruptures can be classified based on the mechanism of injury: contact or non-contact, with non-contact ACL injuries occurring more frequently (Boden et al., 2000). A non-contact mechanism is when no large external force comes into direct contact with the knee at the time of the injury (Marshall, 2010), typically occurring when an individual is involved in a sport were dynamic movements, such as sudden decelerations, pivoting or landings are frequent (Boden et al., 2000; Koga et al., 2010; Krosshaug et al., 2007; Stuelcken et al., 2016). Of all jump landings, single leg landings appear to result in an increase in biomechanical and neuromuscular demands (Pappas et al., 2007; Taylor et al., 2016), and may result in increased risk of sustaining an ACL rupture. Observation video analysis studies appear to show only a slightly higher incidence of ACL injuries occurring when landing on a single leg compared to two legs (Boden et al., 2000; Koga et al., 2010; Stuelcken et al., 2016; Olsen et al., 2004). In most instances, the injury occurred when an opponent was close by and the athlete was reacting to them or the players attention was focused elsewhere such as attempting to score, which may have led to them putting themselves in a position that resulted in an ACL injury (Boden et al., 2000; Olsen et al., 2004; Stuelcken et al., 2016). It is important to study how individuals respond when landing on a single leg as it could reveal risk factors associated with ACL injury.

Panjabi (1992) proposed that three subsystems (passive, active, and neural) make up the sensorimotor system responsible for maintaining joint stability (Figure 1). When a dysfunction of one of the three stabilizing subsystems occurs, this could result in an injury due to abnormal loading of the joint (Panjabi, 1992).

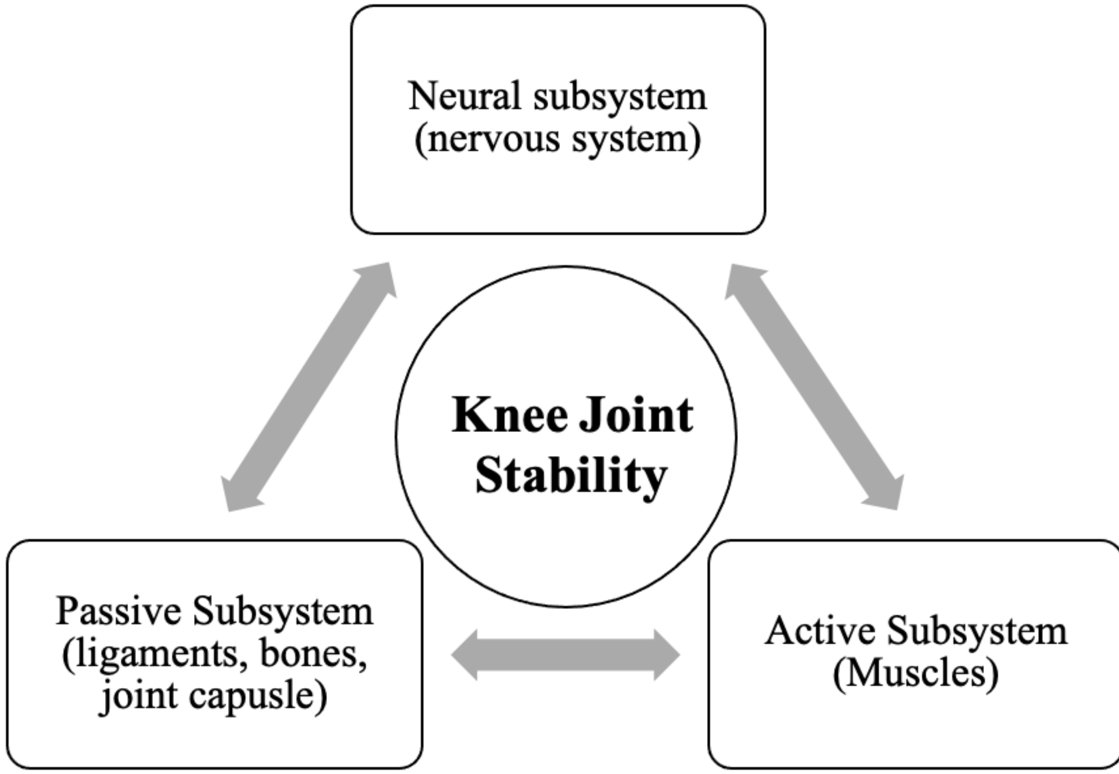

Figure 1*:* Joint stability subsystems as proposed by Panjabi. Adapted from "The stabilizing system of the spine. Part I. Function, dysfunction, adaptation, and enhancement," by M. M. Panjabi, 1992, *Journal of Spinal Disorders, 5*(4), 384.

In terms of an ACL injury, this dysfunction could occur in the neural or active subsystem, with dysfunction of the active subsystem being the inability of muscles producing sufficient stiffness to assist in dissipating the forces experienced at the knee (Brown et al., 2014; Li et al., 1999) while the neural subsystem dysfunction is an abnormal neuromuscular pattern (Nagano et al., 2007; Palmieri-Smith et al., 2008; Weinhandl et al., 2010), which results in the excessive loading of the ACL.

2.2.1 The Knee Joint and its Stabilizing System

In our day to day lives, the knee joint experiences a great amount of force. For these tasks to be achieved successfully and efficiently, the knee joint stability is crucial. If the knee joint is not stable during movements, this could predispose an individual to injuries (Panjabi, 1992; Wikstrom et al., 2006; Williams et al., 2001). Multiple definitions have been given in an attempt to explain what joint stability is, though there is still no consensus on the definition (Noyes et al., 1980; Riemann & Lephart, 2002; Wikstrom et al., 2006; Williams et al., 2001). For this paper, joint stability is defined as "the state of a joint remaining or promptly returning to proper alignment through an equalization of forces" (p. 72) when the body is in motion (Riemann & Lephart, 2002). Joint stability is achieved when there is a balance in the internal and external forces experienced at a joint. The movement or activity of an individual causes the external forces while the ligaments, muscles and bones of the joint are the structures that produce the internal forces and counteracts the external forces (Riemann & Lephart, 2002). The sensorimotor system is responsible for achieving joint stability (Riemann & Lephart, 2002). When the external forces exceed the forces produced by the body structures that surround the knee joint, this could result in these body structures such as the ACL becoming injured. Studying how deficits in the sensorimotor system could affect joint stability and its relation to injury is important, as it could help identify factors within the sensorimotor system that may increase the risk of injury. By identifying these factors, it will then become possible to develop programs to mitigate these risk factors and reduce the risk of the injury.

2.2.1.1 The Sensorimotor System and Joint Stability

The sensorimotor system is responsible for maintaining knee joint stability. If there is a deficit in the sensorimotor system, this results in knee joint instability and could lead to an injury. The sensorimotor system is made up of numerous systems: i) somatosensory, ii) visual, and iii) vestibular (Riemann & Lephart, 2002; Wikstrom et al., 2006). These systems provide afferent information, and the information is then integrated as part of the motor response to help maintain knee joint stability. The motor response is implemented via neuromuscular mechanisms (Riemann & Lephart, 2002; Wikstrom et al., 2006). To maintain knee joint stability, afferent information that is measured by mechanoreceptors in the muscles, ligaments, joint capsule, and skin are relayed to the central nervous system. Muscles activations occur in response to this afferent information and are regulated through feedforward and feedback loops (Riemann & Lephart, 2002). Feedforward loops are pre-programmed muscular controls based on prior experiences while feedback loops are muscular control that may be reflexive in nature and is influenced by sensory information (Riemann & Lephart, 2002; Wikstrom et al., 2006). This process is a simplified picture as to how knee joint stability is maintained through dynamic movement, and hints at the necessary anatomical and physiological systems that are involved.

The passive subsystem as proposed by Panjabi (1992) is compromised of the bones making up the joint, ligaments, and joint capsule. The tibiofemoral and patellofemoral joints, meniscus, knee joint capsule and the ligaments make up the passive subsystem of the knee joint. One of the four major ligaments of the knee is the ACL (Abulhasan & Grey, 2017). The ACL runs in a posterior-lateral direction superiorly, with the point of origin starting from the anterior intercondylar area of the tibia and the point of inserting being the posterior part of the medial side of the lateral condyle of the femur (Goldblatt & Richmond, 2003; Starkey & Brown, 2015). Two distinct bundles of fibers make up the ACL, the anteromedial and posterolateral bundles. Depending on the knee position, the bundles will be relaxed or taut (Goldblatt & Richmond, 2003; Starkey & Brown, 2015). At full knee extension, the anteromedial bundle is relaxed while the posterolateral bundle is taut. However, the anteromedial bundle starts to become taut while the posterolateral bundle begins to relax when flexion of the knee occurs (Goldblatt & Richmond, 2003; Starkey & Brown, 2015). The primary purpose of the ACL is to prevent anterior translation of the tibia relative to the femur and also serves a secondary function to help prevent hyperextension of the knee, excessive internal rotation movement of the tibia relative to the femur as well as varus and valgus movement (Goldblatt & Richmond, 2003; Starkey & Brown, 2015).

The neural subsystem, made up of mechanoreceptors found in the ligaments, tendons, and muscles, assists with feedforward and feedback mechanisms (Panjabi, 1992; Solomonow & Krogsgaard, 2001). This subsystem is important in preparatory muscle activity and changing the muscle activity in response to a change (Riemann & Lephart, 2002). The active subsystem of Panjabi's model is made up of the muscles and tendons that act on a joint. The muscles produce an internal moment around a joint and counteracts external moments, thus helping to maintain joint stability. In the case of the knee joint, the muscles surrounding the knee joint can be divided into an anterior, posterior, medial and lateral musculature group. The anterior aspect of the knee is made up of all the muscles in the quadriceps muscle group: i) rectus femoris, ii) vastus medialis, iii) vastus lateralis and iv) vastus intermedius. The muscles in the posterior chain of the lower extremity include the hamstrings and gastrocnemii. The hamstring muscle group is comprised of the biceps femoris, semimembranosus and semitendinosus. Previous research studying muscle activity and knee joint stabilization by generating an internal moment found hamstring muscle activity produced a flexion moment while quadriceps muscle activity generated an extension moment at the knee joint (Lloyd & Buchanan, 2001; Krishnan et al., 2008; Buchanan & Lloyd, 1997; Lloyd et al., 2005). For internal abduction or adduction moments at the knee joint, the co-contraction of the medial or lateral quadricep and hamstring muscle group is suggested to be primarily responsible for contributing to such moments, with smaller contributions from other muscles (Lloyd et al., 2005; Lloyd & Buchanan, 2001; Zhang & Wang, 2001). The medial musculature of the thigh is thought to have a varus moment arm while the lateral musculature of the thigh has a valgus moment arm (Besier et al., 2003; Lloyd & Buchanan, 2001; Zhang & Wang, 2001). This is in part due to the quadriceps muscle inserting to the midline of the tibia which could counter an external varus and valgus force (Lloyd & Buchanan, 2001). With regards to the hamstring muscle group, if an external load is applied to the knee joint, causing the femur to rotate about one condyle, this will result in the moment arms that opposes such motion to be lengthened while the moment arms that produce such motion will be shortened (Besier et al., 2003). Therefore, when the medial Q:H muscles of the thigh contract, this produces an internal knee adduction load and resists external knee abduction loads (Zhang & Wang, 2001), which could unload the ACL and reduce the risk of an ACL rupture (Palmieri-Smith et al., 2009).

2.2.2 Loading of the ACL

Understanding how the ACL is loaded during movement is important as it can reveal the mechanism of an ACL injury as well as risk factors associated with ACL injury. One factor that contributes to ACL loading is when anterior shear force is applied to the proximal end of the tibia (Berns et al., 1992; Fleming et al., 2001; Markolf et al., 1995). Previous cadaveric studies investigating the load on the ACL when external forces are placed on the knee found that when an anterior shear force was applied at the proximal end of the tibia, it resulted in an increase in ACL strain while a pure internal and external tibial rotation moment did not significantly strain the ACL (Berns et al., 1992; Markolf et al., 1995). When a varus or valgus moment is applied to the femur relative to the tibia in combination with anterior shear force, there is a significant increase in the strain measured in the ACL (Markolf et al., 1995). When the knee is weightbearing in healthy adults, the results from the cadaver studies are only partially supported. Under weightbearing conditions, anterior shear load placed on the tibia resulted in an increase in measured ACL strain (Bates et al., 2019; Fleming et al., 2001), as well as an external knee abduction moment (Bates et al., 2019) and internal tibial rotation moment (Bates et al., 2019; Fleming et al., 2001). The aforementioned studies suggest that multiple planar loads results in peak ACL strain, with anterior tibial translation coupled with valgus, varus or internal rotation moments causing the greatest amount of ACL strain. Some caution should be used when interpreting the results from all mentioned studies as the devices that were used to measure ACL strain was implemented only into the anteromedial bundle of the ACL and might not represent the strain experienced in the posterolateral bundle of the ACL.

Previous studies looking at how knee joint angles influence ACL loading have found that when knee flexion angles ranged from 0° to 30°, the ACL would experience the greatest amount of force (Li et al., 2004; Markolf et al., 1995; Renström et al., 1986), with forces peaking at 30° knee flexion (Li et al., 2004). Prospective studies appear to support this, with a smaller peak knee flexion angle and larger vertical ground reaction force (GRF) during a jump landing test being predictors of ACL injury (Leppänen, Pasanen, Kujala, et al., 2017) as well as peak external knee flexion moment (Leppänen, Pasanen, Krosshaug, et al., 2017). These models suggest that individuals who have a stiffer landing (i.e. smaller peak knee flexion angle) are more likely to suffer an ACL injury, as a stiffer landing places more stress on the ACL. It should be noted that while these factors were predictors, a number of uninjured participants also displayed similar numbers for these variables (Leppänen, Pasanen, Krosshaug, et al., 2017; Leppänen, Pasanen, Kujala, et al., 2017), which indicates that these variables cannot be used in isolation as part of a screening test.

On the other hand, there are studies that disagree with the conclusion that frontal plane variables do not result in a non-contact ACL injury (Hewett et al., 2005; Kobayashi et al., 2010; Koga et al., 2010). Hewett et al. (2005) proposed that individuals who display a dynamic knee valgus position, whereby the individual will display a combination of hip adduction, hip internal rotation and tibial abduction relative to the femur, will have a higher likelihood to have an ACL injury and sagittal plane factors do not play a role in predicting ACL injury. Females who sustained an ACL injury had greater external knee abduction moments and knee abduction angles when landing from a jump while measured sagittal plane joint kinematics and kinetics were not significant predictors for ACL injury (Hewett et al., 2005). While the strain experienced by the ACL was not measured directly, this study suggests that large external knee abduction moments load the ACL to the point of failure. Previous observational studies also support the theory of dynamic knee valgus position as being the mechanism of ACL injury, with video analysis revealing that at the point of ACL injury, the knee's position is abducted and coupled with internal tibial rotation (Kobayashi et al., 2010; Koga et al., 2010). One proposed explanation as to why there are differences between the prospective studies is related to how the ground reaction forces and motion data are filtered. While both studies used a Butterworth filter, Hewett et al. (2005) filtered their force data with a cut-off frequency of 50Hz and their motion data at 9Hz, while Leppänen et al. (2017) filtered both their force and motion data at 15Hz. These differences in methods to filter data can have a significant impact on calculated knee joint moments, with different cut-off frequencies for filtering force and motion data resulting in sudden spikes in joint moments during the first 100ms of a dynamic movement (Kristianslund et al., 2012). Furthermore, both studies used a different method for their regression models to determine what variables were associated with an ACL injury.

More recent research appears to suggest that this differing opinion on which movement is the primary culprit in the mechanism of ACL injury discounts the multi-planar nature of an ACL injury (Boden & Sheehan, 2022; Quatman et al., 2010). This highlights the importance of considering how all planes of motion should be studied when trying to understand the mechanism of ACL injury.

Besides the position of the knee that could cause the strain on the ACL to exceed the failure threshold, muscle activity also influences the forces that travel through the ACL (Li et al., 1999, 2004; Maniar et al., 2020; Renström et al., 1986; Solomonow et al., 1987). Muscles are responsible for joint position and motion during a movement (Riemann & Lephart, 2002), and act as a dynamic restraint against external forces in trying to maintain knee joint stability (Panjabi, 1992; Solomonow & Krogsgaard, 2001). The feedforward and feedback mechanism that make up neuromuscular control is responsible for turning muscles on and off as well as regulating the magnitude of muscle activity (Riemann & Lephart, 2002). Understanding how muscles prepare and react to a dynamic manoeuvre, this can reveal how altered neuromuscular control could load the ACL and is indicative of an increased risk of sustaining an ACL injury.

The quadriceps and the hamstring muscle groups contribute to the forces experienced by the ACL. Isolated quadriceps muscle activity results in the highest amount of ACL loading (Li et al., 1999, 2004) while isolated hamstring muscle activity results in the smallest, especially as the knee joint is flexed from 30° to 90° (Li et al., 2004; Renström et al., 1986). The quadriceps muscles are primarily responsible for producing an anterior shear force while the hamstring muscle produced a posterior shear force (Maniar et al., 2020). As previously mentioned, there is an increase in ACL strain when an anterior shear force is applied to the tibia (Berns et al., 1992; Markolf et al., 1995), thus isolated quadriceps muscle activity producing anterior shear force will cause the strain experienced by the ACL to increase. However, during movement, muscles do not work in isolation from one another but in conjunction with one another in an agonistic-antagonistic relationship (Li et al., 1999, 2004; Renström et al., 1986). The co-contraction of both the quadriceps and hamstring muscle group prevents excessive anterior tibial translation (Li et al., 1999) and anterior shear force experienced by the ACL is also smaller compared to just pure quadriceps muscle group contraction (Li et al., 1999, 2004; Renström et al., 1986).

The respective medial and lateral thigh musculature (i.e. Q:H groups) is also responsible for opposing the dynamic knee valgus position that is associated with an increased risk of a non-contact ACL injury as well as less elongation of the ACL (Palmieri-Smith et al., 2009; Serpell et al., 2015). Higher peak knee valgus angles and moments are associated with greater lateral Q:H muscle activity and smaller peak knee valgus angles are associated with greater medial Q:H muscle activity (Palmieri-Smith et al., 2008, 2009). As previous research found that increased knee abduction moments and angles are associated with ACL injury (Hewett et al., 2005), this suggests that the preferential activation of the medial thigh muscles is important to help reduce the load transferred through the ACL.

Besides muscle activity that occurs during a dynamic manoeuvre, preparatory muscle activity can also have an influence on the forces experienced by the ACL. As the ACL injury typically occurs within the first 40ms of IC with the ground, preparatory muscle activity could prevent the lower extremity from being in an at-risk position when landing from a jump (Koga et al., 2010). Therefore, understanding the relationship between preparatory muscle activity and knee joint motion should be studied. Smaller peak knee flexion angles are associated with greater preparatory Q:H co-contraction ratio (due to a smaller mean hamstring muscle activity) (Walsh et al., 2012) as well as larger mean quadriceps muscle activity (Brown et al., 2014). Such muscle activity and Q:H co-contractions also are associated with an increase in anterior tibial shear force (Brown et al., 2014; Shultz et al., 2009). A smaller knee flexion angle at IC is a risk factor associated with ACL injuries, which indicates that preparatory muscle activity patterns should be studied to understand the relationship between preparatory muscle activity in joint kinematics. Furthermore it was noted that in female athletes, a higher peak knee valgus angle was associated with increased preparatory vastus lateralis and lateral hamstring activity, while a lower peak valgus angle was associated with increased preparatory vastus medialis activity (Palmieri-Smith et al., 2008).

While measuring ACL strain in-situ is not possible, some factors can be looked at in an indirect way in explaining whether there could be an increase in ACL strain. Factors associated with an increase in ACL loading includes a smaller knee flexion angle (Li et al., 1999; Markolf et al., 1995; Renström et al., 1986), greater knee abduction angle, greater preparatory and reactionary quadriceps muscle activity and smaller hamstring (Li et al., 1999, 2004; Maniar et al., 2020; Renström et al., 1986) and gluteus medius muscle activity (Maniar et al., 2020). Studies also look at muscle activity and joint movement during dynamic movement, such as landing on a single or double leg in hopes of understanding how these movements could result in ACL injury.

2.2.3 Single Leg Versus Double Leg Dynamic Tasks

Within the literature, double leg and single leg dynamic manoeuvres have been analysed as these reflect the movements when a non-contact ACL injury often occurs. Examples of such dynamic tasks include double leg and single leg drop landings (Pappas et al., 2007; Yeow et al., 2010), double leg (Goerger et al., 2015; Hewett et al., 2005) and single leg drop vertical jumps (Rocchi et al., 2018), countermovement jumps and forward jumps (Donohue et al., 2015; Palmieri-Smith et al., 2008; Taylor et al., 2016). Studies comparing single-leg and double-leg jump landing tasks have noted that for single leg landings, there is a larger knee valgus angle but smaller knee flexion angle at IC (Pappas et al., 2007; Yeow et al., 2010), smaller peak knee flexion angle (Donohue et al., 2015; Pappas et al., 2007; Taylor et al., 2016; Yeow et al., 2010), larger knee abduction angles (Pappas et al., 2007) and moment (Taylor et al., 2016). The smaller knee flexion angle has been associated with increased ACL loading (Li et al., 1999, 2004; Markolf et al., 1995) while a large knee valgus angle and peak knee abduction moment is a predictor for a non-contact ACL injury (Hewett et al., 2005), making the single leg jump important to study. Besides the difference in lower extremity biomechanics, lower extremity muscle activity is greater when a single leg landing is performed compared to a double leg landing (Pappas et al., 2007). This significant increase in muscle activity suggests that single leg landings challenge the neuromuscular control of an individual to a greater degree compared to double leg landings. This is because individuals are trying to absorb the forces that are experienced when landing while also trying to maintain their balance (Pappas et al., 2007). The increased biomechanical and neuromuscular demands during single leg landings may result in increased risk of sustaining an ACL rupture. Past observation video analysis studies appear to support this notion with studies conducted by Boden et al. (2000) and Koga et al. (2010) finding that athletes who suffered ACL injuries landed on a single leg more often compared to two legs. It is important to study how individuals respond when landing on a single leg as it could reveal risk factors associated with ACL injury.

By investigating the mechanism of a non-contact ACL as well as factors influencing strain in the ACL, factors associated with increased risk of non-contact ACL injury can be found and studied. With single leg dynamic movements being more challenging to one's neuromuscular control compared to double leg manoeuvres, research should look at the muscle activity as well as knee joint kinematics during the preparatory and landing phases of single leg jump movements to understand how the ACL might be injured during this movement and why females have an increased risk of sustaining an ACL injury.

2.4 Sex-Related ACL Injury Risk Factors

Females who participate in sports that primarily involve jumping and cutting are 2-4 times more likely to incur an ACL injury compared to their male counterpart (Arendt et al., 1999; Montalvo et al., 2019; Prodromos et al., 2007). However, this sex difference only becomes apparent after puberty. Prior to puberty, there is no difference in the likelihood of sustaining an ACL injury between male and female athletes (Stracciolini et al., 2015). Research has shown that there is a sex disparity in the rate of non-contact ACL injuries between male and female athletes.

Some of this sex disparity is attributed to non-modifiable factors such as anatomical structure and hormones. Females tend to have smaller ACL in terms of length, cross-sectional area and volume compared to males (Anderson et al., 2001; Chandrashekar et al., 2006). These smaller ACLs may be more prone to failure at lower levels of loads. Female ACLs also tend to have different tensile properties compared to male ACLs, with the female ACLs failing at lower loads compared to males (Chandrashekar et al., 2006). Female sex hormones during the menstrual cycle have also been studied to determine whether they play a role in increasing the risk of an ACL injury. It has been proposed that the changes in female sex hormones can alter the neuromuscular control or knee joint movement (Balachandar et al., 2017; Dedrick et al., 2008; Park et al., 2009), with this change in neuromuscular control and knee joint motion being similar to the mechanisms identified for non-contact ACL injuries. However, there is no clear consensus on whether sex hormones affect the risk in sustaining an ACL injury due to limited evidence and mixed quality of studies (Abt et al., 2007; Balachandar et al., 2017; Hertel et al., 2006).

More research is needed to understand how sex hormones influences neuromuscular control and could lead to an increased risk of an ACL injury.

While anatomical and hormonal factors have been identified as being partially responsible for the increased risk of non-contact ACL injuries, these factors are non-modifiable and so, they cannot be altered to try and reduce the risk of an ACL injury. However, by identifying modifiable biomechanical and neuromuscular risk factors associated with ACL injuries in females, we may be able to identify individuals at a higher risk of an ACL injury and intervene to try and prevent an ACL injury from occurring.

2.4.1 Biomechanical and Neuromuscular Control

Biomechanical risk factors have been looked at in trying to help explain the sex related differences in ACL injury incidence. Previous studies have found that females landed with greater knee valgus angles (Ford et al., 2003; Pappas et al., 2007; Russell et al., 2006), greater internal tibial rotation (Nagano et al., 2007) and less knee flexion angles (Decker et al., 2003) and ROM (Schmitz et al., 2007; Weinhandl et al., 2010) compared to males when landing from a jump. This motion of the knee joint is similar to the "position of no return", which is described as landing with less knee flexion and increased knee valgus when a non-contact ACL injury occurs (Ireland, 1999; Koga et al., 2010; Krosshaug et al., 2007), which may explain to a certain degree why females are more likely to tear their ACL. Furthermore, these movement patterns place greater strain on the ACL, as evidenced by previous in-vivo studies revealing certain combinations of movement, such as applying a valgus moment when the knee is nearly extended, resulting in an increase in the strain sustained by the ACL (Berns et al., 1992; Markolf et al., 1995).

Muscle activity is, in part, responsible for the knee movement patterns that are measured when a dynamic manoeuvre is performed. Therefore, neuromuscular factors have also been implicated in explaining ACL injuries. When males undergo puberty, they experience a neuromuscular ‘growth spurt’ to accompany their actual growth spurt. This allows for males to produce greater rates of power, peak quadriceps and hamstring moment and better coordination that allow them to control their longer limbs (Hewett et al., 2004; Quatman et al., 2006). However, females do not experience this neuromuscular ‘growth spurt’, which places them at a higher risk of sustaining an ACL injury. This is because the longer limbs result in longer moment arms, but if insufficient force is produced to prevent excessive moment experienced by the knee, this will result in a rupture of the ACL (Hewett et al., 2004; Quatman et al., 2006). This weaker neuromuscular control of the lower extremity may also explain the neuromuscular theories that have been proposed.

Females also display different neuromuscular patterns compared to males, with Hewett et al. (2010) and Pappas et al. (2016) proposing the quadriceps dominance and the ligament dominance theories as being common altered neuromuscular profiles associated with an increased risk of ACL injury. The ligament dominance theory states that the muscles of individuals at a higher risk of sustaining an ACL injury do not sufficiently absorb the forces that are experienced during dynamic movements, resulting in the individual having large knee valgus ROM as well as valgus moments (Pappas et al., 2016). This results in the ACL having to absorb more force over a short period of time and could result in the failure of the ACL (Hewett et al., 2010). The ligament dominance theory is characterized by the posterior muscle chain of the lower extremity being underutilized when performing a dynamic manoeuvre, such as landing from a jump (Ahmad et al., 2006; Ebben et al., 2010; Hewett et al., 2010). The quadriceps dominance theory states that individuals who primarily use the quadriceps muscle group to stabilize the knee joint during dynamic movement are more likely to sustain an ACL rupture (Hewett et al., 2010). Both theories are thought to be related, such that when there is insufficient hamstring and stronger quadriceps muscle activity, this results in a smaller knee flexion angle and increased ACL strain occurs. Furthermore, with the insufficient hamstring muscle activity, the ACL is preferentially loaded to help maintain knee joint stability instead of the posterior chain muscle group, resulting in an increase in ACL loads.

2.4.1.1 Sex Differences in Biomechanical and Neuromuscular Control During Single-Leg Jump-Landings and its relation to ACL injury

As previously mentioned, jump landings are a common mechanism of non-contact ACL injury, as it is a movement that produces high GRF over a short period of time that is experienced across the knee and could lead to an ACL rupture (Boden et al., 2000). Sex differences when completing single leg jump landing tasks have been studied due to the increased biomechanical and neuromuscular demands that are placed on the knee and by extension, the ACL when landing on a single leg (Pappas et al., 2007; Taylor et al., 2016).

It has been suggested that males and females have different biomechanical and muscle activity patterns when landing from a jump, with females landing with smaller knee flexion angles and knee flexion ROM (Schmitz et al., 2007; Weinhandl et al., 2010), larger knee valgus angles (Pappas et al., 2007; Russell et al., 2006) and greater internal tibial rotation (Nagano et al., 2007) compared to males. During the preparatory phase of a jump-landing, females tend to have greater quadriceps muscle activity and Q:H co-contraction ratios (Nagano et al., 2007; Zazulak et al., 2005) compared to males. During the landing phase, female athletes again preferentially use their quadriceps (Hughes & Dally, 2015; Urabe et al., 2005; Walsh et al., 2012; Zazulak et al., 2005), which results in a smaller knee flexion angle (Pappas et al., 2016) and increased anterior tibial shear force (Li et al., 1999; Maniar et al., 2020; Renström et al., 1986). Females also tend to have smaller hamstring muscle activity (Hughes & Dally, 2015), and greater Q:H co-contraction ratios (Nagano et al., 2007; Palmieri-Smith et al., 2009), which suggests that there was low hamstring muscle activity and/or high quadriceps muscle activity when they co-contracted with one another. This result is also supported by a study conducted by Palmieri-Smith et al. (2009), which calculated at Q:H co-contraction index (CCI) as proposed by Ruldoph et al (2000). Palmieri-Smith et al. (2009) found that females had lower Q:H CCI compared to male participants, specifically the medial Q:H CCI. This was in part related to the smaller medial hamstring muscle activity that female participants had compared to male participants. They also found that in females, there was a significant association between medial Q:H CCI and peak external knee abduction moment in females. All these factors have been shown to be associated with increasing strain experienced by the ACL, suggesting that the different neuromuscular control employed by females during the preparatory or landing phase during a single-leg jump-landing task places them at a higher risk of rupturing their ACL compared to their male counterparts.

There are studies that have found conflicting results in which no sex difference in muscle activity for the quadriceps (Fagenbaum & Darling, 2003; Palmieri-Smith et al., 2009; Pappas et al., 2007) and the hamstring muscle group (Fagenbaum & Darling, 2003; Nagano et al., 2007; Pappas et al., 2007) during similar tasks. These differences in results could be due to the variability in the study designs, such as skill level of the participants, heights of platform for the single leg jump tasks and sample sizes. However, the consensus is that females exhibit a different neuromuscular pattern compared to males when landing from a jump and it is this difference that is addressed in neuromuscular training programmes to prevent an ACL injury (Hewett et al., 2010).

While these studies have shown that differences between males and females exist in terms of joint movement and muscle activity, there are less studies comparing the relationship between preparatory muscle activity and landing sagittal plane knee joint kinematics and whether there is a sex difference. A study conducted by Brown et al. (2014) only examined the relationship between preparatory muscle activity and sagittal plane knee joint kinematics in females only while the study conducted by Walsh et al. (2012) collapsed the males and females into a combined group as there were no sex differences when they performed separate 1-way ANOVAs. Palmieri-Smith et al. (2008) looked at the relationship between preparatory muscle activity and peak knee valgus angles and found that for females, a higher peak knee valgus angle was associated with increased preparatory vastus lateralis and lateral hamstring and decreased preparatory vastus medialis activity, while none of the preparatory muscle activity was associated with knee joint angles in males. Therefore, more studies are required to address the limitations to create a clearer picture on how muscles control the lower extremity and dissipate the forces that travel across the knee to prevent ACL injuries from occurring.

There has been a large emphasis on identifying the biomechanical and neuromuscular risk factors specific to each sex as these are modifiable risk factors as opposed to anatomy and sex hormones. Presently, it is accepted that females tend to land with smaller knee flexion angles, hamstring muscle activity and Q:H co-contractions as well as having larger quadriceps muscle activity compared to their male counterparts, though not all studies support these findings. There is also less understanding on how preparatory muscle activity is associated with knee joint movement when landing from a jump, and whether sex is a factor. By identifying the biomechanical and neuromuscular characteristics associated with ACL injuries between sexes, it could allow for identification of individuals who may be at a higher risk of an ACL injury.

2.5 Assessment of Knee Joint Stability

Biomechanical and neuromuscular analyses are commonly performed to quantify lower extremity joint function and is often used to differentiate between normal and aberrant movement patterns that could increase the risk of ACL injury (Kleissen et al., 1998; Wikstrom et al., 2006). These analyses are achieved by using different technologies, such as electromyography and motion capture systems.

2.5.1 Electromyography

To understand the contribution of skeletal muscle to joint stability, electromyography (EMG) has been employed. EMG is a tool that measures the superposition of muscle action potentials; detecting electrical activity at the surface of the skin associated with muscle excitation (Riemann et al., 2002). It can be used to measure and understand how muscles turn on and off as well as the magnitude of the muscle activity when performing movement as muscles are considered to be dynamic restraints of the body and they assist in maintaining joint stability (Riemann et al., 2002; Theisen et al., 2016; Wikstrom et al., 2006).

In most studies performing a neuromuscular analysis of a dynamic manoeuvre, surface EMG (sEMG) is used to measure muscle activity of the superficial muscles (Palmieri-Smith et al., 2008; Pappas et al., 2007; Zazulak et al., 2005). Research has established that sEMG provides reliable data of the muscles of the lower extremity when performing maximum voluntary isometric contraction (MVIC) trials as well as dynamic movements such as walking (Rutherford et al., 2020), landing from a jump and performing a run and rapid change of direction (cut) manoeuvre (Fauth et al., 2010). It was noted that while there was less reliability during dynamic movements, all intraclass correlation coefficient (ICC) values were greater than 0.80, which indicates good to excellent reliability (Fauth et al., 2010; Rutherford et al., 2020). Some considerations for the accuracy of sEMG includes crosstalk between other muscles (Fauth et al., 2010; Riemann et al., 2002), skin-electrode impedance and the electromagnetic radiation from power sources that are either 50 or 60 Hz (Tankisi et al., 2020). To decrease the risk of crosstalk and to ensure adequate reliability, the Surface Electromyography for the Non-Invasive Assessment of Muscles (SENIAM) group proposed a standardized guideline for best practice in placement of the electrodes as well as prepping the skin on limbs so that the signal measured by the electrode is of the best quality (SENIAM, 1999a, 1999b).

When reporting sEMG data, the data is typically normalized to a standard value to allow comparisons to be drawn between participants and muscles (De Luca, 1997). However, there does not appear to be a consensus as how to normalize sEMG data, though the Journal of Electromyography and Kinesiology noted that the EMG values should be normalized to a maximal voluntary contraction ("Standards for Reporting EMG Data," 2018). MVICs appear to be the most common normalization used to determine the standard value (Hughes & Dally, 2015; Nagano et al., 2007; Rudolph et al., 2000; Rutherford et al., 2020; Shultz et al., 2009). By normalizing EMG to the maximal amplitude of MVICs, it provides information on how much a muscle activates with respect to the maximal activation of the muscle, thus providing a physiological reference in which to compare EMG between different muscles and participants (Burden, 2010).

2.5.1.1 Calculation of Muscle Co-Contractions

Muscle co-contraction is the simultaneous contraction of both agonist and antagonist muscles. Muscle co-contraction is important in preventing sport injuries as it helps to maintain joint stability (Solomonow et al., 1987) and to help produce efficient movement (Solomonow & Krogsgaard, 2001). One way to measure muscle co-contraction is to record EMG data and use equations to estimate muscle co-contractions. Previous research has shown that muscle co-contractions during the preparatory phase of landing (prior to IC) as well as during the landing phase is correlated to knee flexion angles (Ohji et al., 2019; Podraza & White, 2010). It is important to note that depending on the method used to calculate the muscle co-contraction, different conclusions can be drawn from the calculated co-contraction.

The CCI proposed by Rudolph and colleagues (2000) looks at both temporal and amplitude variables, as it provides an estimation of the relative activity of two muscles, while at the same time as providing that amount of co-contraction over the period of interest. The equation as proposed by Rudolph and colleagues (2000) is provided below:

$$Co - Contraction\ Index\ (CCI) = \int_{0}^{101} \frac{\left(\frac{EMGS}{EMGL} \times (EMGS + EMGL)\right)}{101}$$

EMGS represents the normalized EMG data of the less active muscle activity and EMGL represents the normalized EMG data of the more active muscle activity measured at a specific point of time. In the study conducted by Rudolph et al. (2000), this period of interest was determined to be the entire weight acceptance phase during a landing task. A low calculated CCI was described to represent one of two situations, one being that both muscles have low levels of activity, and the second being that one muscle has a high level of muscle activity while the other muscle has a low level of muscle activity. A high calculated CCI represented a high level of muscle activity in both muscles over the time of interest. Rudolph et al. (2000) suggested that a low CCI value indicated smaller amount of muscle co-activity, resulting in the individual displaying a more selective muscle activation pattern during an activity while a high CCI value indicated a greater amount of muscle co-activation, resulting in the individual having a more generalized muscle activation pattern being used. This method to determine the CCI has good to excellent reliability in healthy individuals when walking on a treadmill, with the ICC values ranging from 0.88-0.98 for the lateral and medial Q:H group (Mohr et al., 2018).

2.5.2 Kinematics

Kinematics is the study of movement without respect to the forces that causes the motion (Riemann et al., 2002). Kinematics can be measured by using motion capture systems to track body segments in space and finding the joint angles. These motion capture systems are comprised of high-speed cameras and markers that are placed on the skin and are the gold standards for motion capture technology. Motion capture systems are considered to be passive when the markers reflect light while active systems are considered to be when markers emit light (Pappas et al., 2013).

Kinematic measurement is used to identify movement patterns that the body experiences when performing a dynamic manoeuvre. This can help reveal risk factors that are associated with increased risk in sustaining an injury, such as ACL tear (Hewett et al., 2005). However, to ensure that biomechanical analyses are accurate, data acquisition using technology that is both accurate and reliable is required.

The accuracy of a motion capture system is influenced by the locations of the different cameras with respect to one another, the distance between the markers and the cameras, how many markers are being used as well as the motion of the markers that are being recorded (van der Kruk & Reijne, 2018). Additionally, markers placed on the skin may not accurately represent bone movement, the goal of motion capture analyses in biomechanics. This is referred to as skin motion artifact (Benoit et al., 2006; Cereatti et al., 2017). Calculated joint kinematics were found to be repeatable when measured by skin markers compared to markers mounted on bone pins within a participant. However, it was reported that there was a large error between both marker types between subjects due to skin movement artifact (Benoit et al., 2006). Furthermore, it was reported that skin markers underestimated knee abduction angles and overestimated internal rotation angles during a cutting movement (Benoit et al., 2006). Therefore, skin markers may not fully mimic lower extremity bone movement, particularly movements in the frontal and transverse plane, thus caution should be used in interpreting the joint kinematics being measured when using skin markers.

Reliability of kinematic analyses has also been conducted. Using intra-class correlation coefficients (ICC), previous studies found that knee joint ROM in the sagittal (ICC: 0.73-0.94) and frontal (ICC: 0.77-0.87) plane during gait between sessions (Rutherford et al., 2020) and sagittal, frontal and transverse knee joint ROM during landing from a jump within sessions (ICC: 0.91-0.94) (Mok et al., 2016) and between sessions (ICC: 0.73-0.81) (Ford et al., 2007; Mok et al., 2016) to have good to excellent reliability, though sagittal knee joint excursion reliability from the study conducted by Ford et al. (2007) was fair to good (ICC: 0.553). Therefore, the use of motion capture and force plates to capture data so that kinematic analysis can be performed is reliable and accurate.

The passive, active and neural subsystems must work together in harmony to maintain knee joint stability and prevent ACL injuries from occurring during dynamic activity. The use of motion capture and EMG to indirectly measure aspects of these subsystems as proposed by Panjabi (1997) that are responsible for joint stability allow us to determine how a non-contact ACL injury occurs. By identifying risk factors associated with ACL injuries, this could potentially help reduce the burden associated with ACL injuries.

2.6 Conclusion

Females are at a higher risk of sustaining an ACL injury compared to males, with one explanation being that dysfunction of one of the three stabilizing subsystems that helps maintain knee joint stability. EMG and motion capture has been used to indirectly measure these subsystems to identify factors associated with increased loads experienced by the ACL, with the combination of anterior tibial shear force, knee abduction moment and internal tibial rotational moment. Knee joint kinematics and preparatory and reactionary muscle activity has been associated with these loading patterns. Neuromuscular control is challenged when landing on a single leg or under a perturbed condition, and sex specific movement and muscle activity patterns which are associated with an increased risk of ACL injury could occur. However, more research is needed to determine the sex specific relationship between preparatory muscle activity and knee joint kinematics during single leg landings.

Chapter Three: Methodology

3.1 Participant Recruitment

Participants provided written informed consent for study procedures. Approval was granted by the Acadia University Research Ethics Board (REB) (14-39) for the project titled, "The effect of sex, age and leg dominance on lower limb biomechanics during athletic maneuvers: Relevance to preventing Anterior Cruciate Ligament (ACL) injuries". Participants were recruited either from the varsity athletics teams at Acadia University via email, word of mouth or through an email sent out to the Acadia university community as well as the minor basketball and soccer associations. My role on this research project was to assist in the collection and processing of data (2018-2019). To address the thesis objectives, which are in part addressing the objectives of the project plan, deidentified drop jump data was obtained from the John MacIntyre motion Laboratory of Applied Biomechanics (mLAB) in 2021, collected using the methodology described below.

For recruitment, researchers contacted interested participants via email or telephone with a script explaining the study as well as having questions to determine eligibility for the study (Appendix A). The inclusion criteria were as follows:

i. Engaged in a cutting/jumping field or court sport
ii. No previous history of major injuries to the lower extremity or back
iii. Met return-to-sport criteria if sustained sprained ankle 3 months prior to data collection as determined by a trained health care practitioner

If a participant was eligible for the study, an informed consent form explaining the study purpose and methods was sent to the participant or parent(s)/guardian(s) (if the participant was under the age of 18) to review and sign prior to testing.

3.2 Sample Size

The sample size was based on an estimate from the literature studying muscle activity and knee joint ROM during a jump landing task (Hughes & Dally, 2015; Palmieri-Smith et al., 2008; Pappas et al., 2007; Russell et al., 2006; Schmitz et al., 2007; Urabe et al., 2005; Walsh et al., 2012; Weinhandl et al., 2015; Zazulak et al., 2005). Sample sizes for the female group ranged from 10-21 while sample sizes for males ranged from 9-20 in previous studies.

One study comparing total knee flexion ROM during the landing phase from a single leg jump found that males had 12.9° (standard deviation (SD) = 6.9°) of knee joint ROM while females had 8.3° (SD = 5.9°) of knee joint ROM (Schmitz et al., 2007). Another study looking at differences in muscle activity found that females had greater preparatory rectus femoris muscle activity (mean = 33.6 %MVIC, SD = 18.5 %MVIC) compared to males (mean = 18.7 %MVIC, SD = 8.2 %MVIC) (Zazulak et al., 2005). Another study looking at the differences in peak muscle activity found that females had higher peak rectus femoris muscle activity (mean = 209.2 %MVIC, SD = 53.0 %MVIC) and smaller peak biceps femoris muscle activity (mean = 114.5 %MVIC, SD = 58.2 %MVIC) compared to males (rectus femoris: mean = 149.3 %MVIC, SD = 42.9%; biceps femoris: mean = 234.9 %MVIC, SD = 85.9%) (Hughes & Dally, 2015). Based on a 2-sample power calculator (G*Power), with a power of 80%, a Beta of 0.20 and an alpha value of 0.05, it was found that:

i) Male sagittal plane knee ROM = 12.9°±6.9°; Female sagittal plane knee ROM = 8.3°±5.9°; the sample size required was 32 in each group.

ii) Male preparatory rectus femoris activity = 18.7%±8.2%; Female preparatory rectus femoris activity = 33.6%±18.5%; the sample size required was 18 in each group.

iii) Male landing rectus femoris activity = 149.2%±42.9%, male landing biceps femoris activity = 234.9±85.9%; Female landing rectus femoris activity = 209.2%±53.0%, female landing biceps femoris activity = 114.5%±58.2%, the sample size required was 7 in each group.

Therefore, the recruitment goal was to include at least 32 participants in each group to ensure statistical power to detect changes in knee range of motion and muscle activation levels during the SLJ task.

3.3 Procedure

3.3.1 Participant Information and Anthropometric Measurements

Upon arriving at the mLAB, participants were given a brief explanation of the data collection and questions were answered at this time. Next, participants completed an informed consent form, an eligibility questionnaire as well as a puberty assessment questionnaire (Appendix B). Once all forms were completed, participants were asked to change into compression shorts and tank top that was provided to them. Anthropometric measures of each participant's height, body mass, thigh and calf circumference as well as foot width was measured and recorded. This was used for descriptive purposes as well as data processing.

3.3.2 Electromyography Setup and Instrumentation

A standardized protocol, in line with the SENIAM guidelines, was used to collect electromyographic data of each participant (Del Bel et al., 2017; SENIAM, 1999b, 1999a). A wireless, 16-channel EMG Delsys Trigno system (Delsys Inc., Natick, MA) was used to collect bilateral EMG data of eight muscles in the lower extremity (interelectrode distance (IED)=10mm, bandwidth: 20-450Hz, preamp gain: 1000x, CMRR<-80dB).

To determine the placement of the surface electrodes on the muscle bellies of the gluteus medius (GM), rectus femoris (RF), vastus lateralis (VL) and medialis (VM), medial (MH) and lateral hamstring (LH), and medial (MG) and lateral gastrocnemius (LG), a measurement guideline set by, and in line with the SENIAM guidelines was used (SENIAM, 1999a, 1999b) (Table 1).

Table 1: SENIAM guidelines on placement of electrodes on the lower extremity

Muscle	Muscle Site
Medial gastrocnemius	35% of the distance from medial knee joint line to tubercle of the calcaneus
Lateral gastrocnemius	30% of the distance from lateral knee joint line to tubercle of the calcaneus
Vastus medialis	20% of the distance from knee's medial joint line to the anterior superior iliac spine (ASIS)
Vastus lateralis	33% of the distance from the knee's lateral joint line to the ASIS
Rectus femoris	50% of the distance between patellar base and ASIS
Medial hamstring	50% of the distance between the ischial tuberosity and medial tibial epicondyle
Lateral hamstring	50% of the distance between the ischial tuberosity and lateral tibial epicondyle

Muscle	Muscle Site
Gluteus medius	50% of the distance from the iliac crest to the greater trochanter

Once the appropriate EMG electrode location was found, the skin of the participant was prepared and electrode site confirmed with an isometric contraction of the muscle. Skin preparation involved the shaving of the area with a razor to remove hair and dead skin before rubbing alcohol swabs were used to clean the shaved area. 3M medical tape was used to dab the prepared area to remove any remaining hair and dead skin that could interfere with the EMG signal. The 16 surface EMG electrodes were then placed on the prepared muscle bellies and were affixed with double-sided tape. The electrodes were placed so that they ran in the direction of the underlying muscle fibers of the muscle bellies. Participants underwent a validation test compromised of a series of manual muscle tests to ensure that the signal quality of each surface EMG electrode was satisfactory. Participants were asked to activate each muscle group by resisting plantarflexion, knee flexion, knee extension, hip extension and hip abduction while EMG signals were visually inspected for quality. Signals that were considered unsatisfactory resulted in re-measurement, skin preparation and replacing the EMG electrode until the signal was adequate. Immediately after the validation test, a one-second bias trial was performed where resting muscle activation was recorded. During this trial, participants were lying supine and instructed to relax. The EMG were then secured to the skin using Coverlet Cover-Roll Stretch Adhesive Bandages to ensure that the EMG units did not detach from the skin and to ensure proper contact during the data collection (Figure 2a and 2b). Fabrifoam wrap covered the electrodes to prevent excessive electrode movement during the data collection (Figure 2c).

a) b) c)

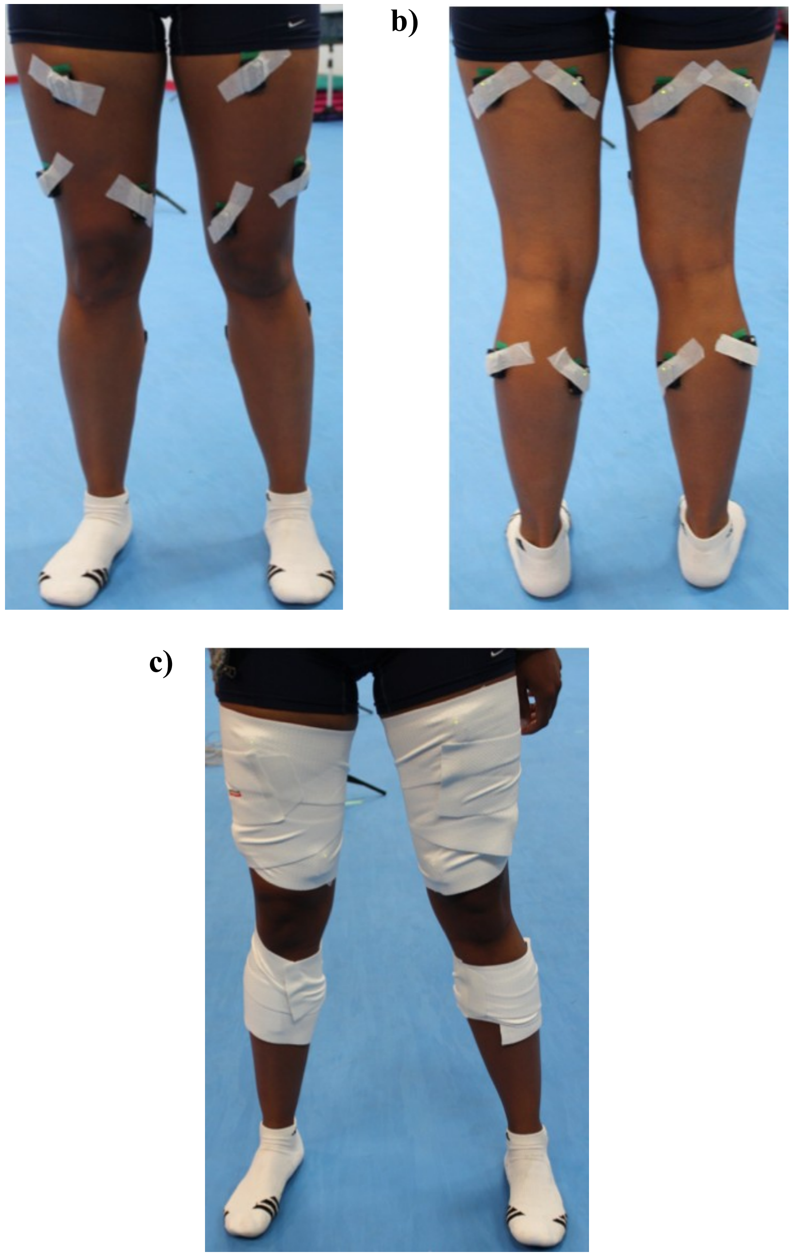

Figure 2: a) Anterior and b) posterior view of electrode placement; c) anterior view of EMG electrodes covered with Fabrifoam wrap

3.3.3 Maximum Voluntary Isometric Contraction

MVICs exercises were completed to record maximum levels of EMG to normalize the jump trials to a percentage of MVIC (%MVIC). Five MVIC exercises were performed on the Biodex System 3 dynamometer (Biodex Medical Systems Inc., New York, NY), in addition to one manual resisted muscle test where participants performed a resisted standing unilateral plantarflexion movement with a researcher applying force down onto the participant's shoulder (Table 2). The manual resisted muscle test was performed in addition to the seated plantarflexion on the Biodex dynamometer to ensure that the trial that was selected elicited the highest MVIC. Each MVIC exercise was performed twice, with participants being instructed to give their maximum effort during each trial for the left and right limb.

Prior to each exercise, lever arms, seat straps and seat position were adjusted appropriately, and lever arm length was recorded onto the data collection sheet. Gravity correction trials were then performed, with participants being instructed to relax their limb. When performing the MVIC exercise on the Biodex, participants were instructed to cross their hands over their chest or on their lap so that participants were unable to leverage themselves when performing the MVIC exercises. Each MVIC exercise was performed for five seconds with EMG and torque data being recorded for a three-second window in the middle of the trial. A minimum of a 30-second rest period between each trial for a MVIC exercise was given for participants to recover. During the MVIC trials, researchers provided verbal encouragement to the participants for them to perform and sustain the maximum contraction.

Table 2: MVIC exercises performed 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°)

Muscle	Exercise
Medial and Lateral gastrocnemii	Standing Plantarflexion*
Gluteus medius	Hip Abduction (Semi-prone)

Note. * Performed without Biodex

3.3.4 Motion Analysis Setup and Instrumentation

For the collection of three-dimensional kinematic data of the entire body, a 12-camera Qualisys motion capture system (Oqus 4) and one high-speed video camera (Oqus 2) was used (Qualisys, Gothenburg, Sweden). A total of 75 14mm retro-reflective makers were placed on bony landmarks of the participants' feet, shanks, thighs, pelvis, trunk, and arms based on a standardized lab marker setup. Single retro-reflective markers were secured with double-sided tape and Coverlet® Cover-Roll Stretch Adhesive Bandages on bony landmarks of the participants while rigid marker clusters with four retro-reflective markers were placed on the left and right thigh and shank of each participant which were wrapped with Fabrifoam Supra wraps. A rigid marker cluster with three retro-reflective markers was secured to the heel of each foot (Figure 3). Finally, participants were asked to wear a headband with five retro-reflective markers around their head. The collection sheet for the surface marker setup as well as the electrode placements for participants can be found in Appendix C.

Figure 3: Anterior (left) and posterior (right) placement of retro-reflective markers

3.3.5 Calibration

Prior to the start of the testing protocol, a 5-second standing calibration trial was performed for each participant. Participants were instructed to stand in the anatomical position while standing on two force plates. This was to allow for joint centers, segment anatomical coordinate systems and reference angles of the ankle, knee, and hip joints to be established. Afterwards, a movement calibration trial was performed, with participants being instructed to move each of their limbs separately, before twisting their trunk and their head. This movement calibration was performed to build an AIM model, which helps automatically identify each marker during dynamic movement trials (Qualisys AB, 2018). Following the completion of these calibration trials, 10 markers located on the medial aspects of the elbows, heels, ankles, and knees, as well as on the base of the 2nd metatarsal were removed for the movement trials. These markers were removed to prevent participants from altering their movements due to apprehension of dislodging these markers and instead, were considered virtual markers.

3.3.6 Testing Protocol

The volume for which the movements were being performed was calibrated using a Qualisys calibration wand and a L-shaped reference system (L-frame) prior to the arrival of the participant for the data collection session. The L-frame was aligned to the corner of one of the force plates (FP3) that was embedded in the ground (Figure 4). This was performed to allow for the accurate tracking and calculation of 2D data to 3D data. The wand is moved in the desired volume, which allows for system to calculate the position of each camera and their orientation in space as well as to align the force plates and marker coordinate system (Qualisys AB, 2018).

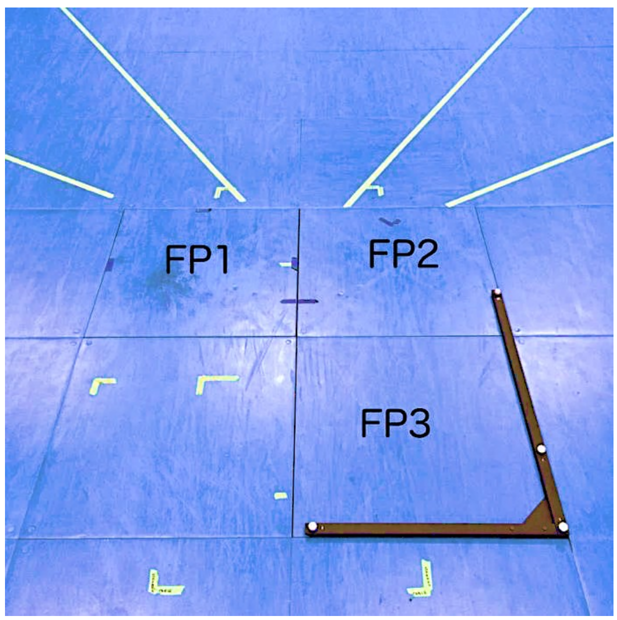

Figure 4: Calibration 'L-frame' and embedded force plate set-up in the mLAB floor

Three AMTI force plates (Advanced Mechanical Technology Inc., Watertown, MA) were used throughout the data collection session to collect GRF data, though only FP1 (for right leg) or FP2 (for left leg) (Figure 3.4) was used to collect GRF data for the SLJ task to determine IC with the ground. Force data was sampled at a rate of 2,000Hz. Force plate, motion capture and EMG data were simultaneously captured using Qualisys Track Manager (QTM) software.

After the calibration trials, a 30cm box was placed 15cm behind the two embedded force plates. Participants were given verbal instructions on how to perform the SLJ trials (Appendix D). The instructions were that participants were to stand on a single leg on top of the box, with their first toe lined up directly to the middle of the box. They were then instructed to lean as far as they could off the box until they "fall" off the box. Upon landing on the force plate on the same leg that they started on, they were to jump as high as possible vertically and then land on the same leg on the same force plate. Upon the second landing, participants were told to maintain their balance for a few seconds. After the verbal instructions, participants were allowed to practice the jumps to ensure they were performing the task correctly. A total of eight successful trials, four trials for the right leg and four trials for the left leg were collected for each participant. A trial was considered to be unsuccessful if the participant was visually observed to have jumped off the box instead of "falling" off (i.e. head/torso displaced vertically upwards as/after foot left box), lost their balance at any point of time during the trial, their non-landing foot touched the ground or the box before the end of the trial, or the landing foot did not completely land on a single force plate. Prior research has shown that during failed jump landings, EMG data was statistically different compared to the successful landing trials (Wikstrom et al., 2008).

3.4 Data Processing

For this study, only the EMG data of the lower extremity and kinematic data of the dominant leg knee joint during the first landing in the SLJ trials was analyzed. The dominant leg was determined as being the limb that the participant used to kick a ball (Del Bel et al., 2017).

3.4.1 Force Plate Data

Force plate data was processed in Visual3D biomechanical software (C-motion Inc., Rockville, Maryland, USA), and was filtered using a low-pass 4th order recursive Butterworth filter (cut-off frequency: 12Hz). Within Visual3D, events were created to reflect the two phases within the jump, the preparatory and the landing phase. The IC event was determined to be the point in time at which measured GRF by the force plate >10N. The preparatory phase was the time period between 200ms before IC to IC (Zazulak et al., 2005) while the landing phase was considered to be the time period from IC to the lowest vertical position of the calculated average of the three sacral markers. Prior research has noted that a sacral marker can be used to approximate the total body centre of mass during level walking and running (Jeong et al., 2018; Napier et al., 2020). This phase during the landing is thought to capture the time period in which the energy generated during the landing has been absorbed or transferred by the lower extremity (Barrios et al., 2016; Pozzi et al., 2017).

3.4.2 EMG

EMG data from the quadriceps and hamstring muscles were examined and processed using custom MatLab™ R2019b (The Mathworks Inc., Massachusetts, USA) software. Raw EMG were visually inspected for movement artifacts, dynamic range saturation or 60Hz noise before being bias corrected, full-wave rectified and linear enveloped using a zero-lag 4th order Butterworth filter (cut-off frequency: 6Hz) (Del Bel et al., 2017; Hubley-Kozey et al., 2006). To amplitude normalize the EMG data, the maximum amplitude for each MVIC trial was found using a 100ms moving window algorithm for each trial. EMG data was then amplitude normalized as a percentage of MVIC (%MVIC). After, EMG data during the preparatory and landing phase was time normalized so that each phase could be described by 101 points with 0% being the start of the phase and 100% being the end point of the phase. This was done for both the preparatory and landing phase, though it is more important for the landing phase as the period for the landing phase will differ within and between participant trials, while the time period for the preparatory phase is consistent between participants and trials. Ensemble average EMG waveforms were calculated for males and females for each muscle for both phases.

To calculate the average root mean square (RMS) for each quadriceps and hamstring muscle during each phase, the RMS function in MatLab™ R2019b was used, with the equation that the function was derived from stated below (The MathWorks, Inc., n.d.), where *x* represents a matrix that is *N*-by-*M* in size, where $N > 1$:

$$x_{RMS} = \sqrt{\frac{1}{N}\sum_{n=1}^{N}|x_n|^2}$$

3.4.3 Kinematics

Tracked retro-reflective marker data was imported into Visual3D biomechanical software (C-motion Inc., Rockville, Maryland, USA) and filtered using a low pass 4th order bidirectional Butterworth filter (cut-off frequency: 12Hz). From the standing calibration, anatomical coordinate systems of each rigid segment of the body were established. During the movement trials, virtual and tracked markers were used to determine the anatomical coordinate systems of the various body segments. A cardan rotation (x-y-z sequence) was used to calculate joint angles, with knee flexion/extension occurring about the x-axis. The rotation sequence flexion/extension followed by adduction/abduction and then internal/external rotation was used, with flexion, adduction and internal rotation being positive. Knee flexion ROM was calculated as being the difference between maximum and minimum knee joint angle during each phase.

3.5 Statistical Analysis

The statistical analysis was performed in SPSS Version 27 and Minitab Version 19.2020.2.0. No missing data was noted for all variables. The mean and SD for the descriptive data was calculated and normality was checked using the Kolmogorov-Smirnov test. If normality was present, a 1-way analysis of variance (ANOVA) was performed to determine if there was statistically significant difference between male and female participants' descriptive data. If the descriptive data did not meet the assumption of normality, a Mann-Whitney U-Test was performed to determine if there was any statistical difference between male and female participants' descriptive data.

The mean and SD for knee joint ROM during the preparatory (200ms from IC to IC) and landing (IC to lowest point that sacral markers descended) phase and EMG data during the preparatory and landing phase (i.e. VM, VL, RF, MH, LH, VM:MH CCI and VL:LH CCI) was calculated. A Kolmogorov-Smirnov test was used to check normality while Levene's test was used to check homogeneity of variance. Skewness and kurtosis were also checked to determine if the data was normally distributed.

For the first primary objective, the data was analyzed in two different ways. To determine if there was a difference between male and female participants for sagittal knee joint ROM, VM:MH and VL:LH CCI during the preparatory and landing phase, a 1-way

ANOVA was performed. If the data was not normally distributed, a Mann-Whitney U test was performed. Cohen's d_s (denominator used was pooled SD) was calculated using:

$$Cohen's\ d_s = \frac{\bar{X}_1 - \bar{X}_2}{\sqrt{\frac{(n_1 - 1)SD_1^2 + (n_2 - 1)SD_2^2}{n_1 + n_2 - 2}}}$$

Where *X* represents the mean for each group, *n* represents the number of participants in each group and *SD* represents the standard deviation for each group. To interpret Cohen's d_s, a small effect size is considered to be equal to 0.2, a medium effect size is equal to 0.5, and a large effect size is equal to 0.8 (Cohen, 1988, 1992; Lakens, 2013).

For the EMG variables, mixed ANOVAs were performed. If the data was not normally distributed or had significant skewness or kurtosis (z-score ≥ 1.96 or ≤-1.96), a Johnson transformation was performed using Minitab in preparation for statistical tests. A 2x3 mixed ANOVA was performed for the quadriceps muscle group while a 2x2 mixed ANOVA was performed for the hamstring muscle group for both the preparatory and the landing phase. The between-subject variable was sex while within-subject variable was mean EMG activity for the respective muscle groups and phases. The muscles that comprise the quadriceps group were VM, VL and RF, while the muscles that made up the hamstring groups were the MH and LH. Mauchly's test of sphericity was performed to ensure that the assumption of sphericity was not violated. If this assumption was violated, either the Greenhouse-Geisser value or Huynh-Feldt value was used. The effect size for the ANOVA was reported as partial eta-squared (η^2) (small effect: 0.01, medium effect: 0.06, large effect: 0.14) (Cohen, 1988; Lakens, 2013). If a main effect was noted, pairwise comparisons were performed. Cohen's d_{av} (denominator used was the average standard deviation of both measures) was calculated using the equation:

$$Cohen's\ d_{av} = \frac{M_{diff}}{\frac{SD_1 + SD_2}{2}}$$

Where M_{diff} represents the mean difference between groups and SD represents the standard deviation for each group. As Cohen's d_{av} is similar to Cohen's d_s, Cohen's d_{av} effect size can be interpreted as 0.2 = small effect size, 0.5 = medium effect size, and 0.8 = large effect size (Cohen, 1988, 1992).

To address the second primary objective, a backward entry linear regression was performed, with predictor variables being VL, VM, RF, MH, and LH RMS and VM:MH and VL:LH CCI, and the dependent variable being sagittal plane knee ROM during the landing phase. Pearson correlation tests were performed to determine if there was a significant correlation between VM:MH CCI, and mean VM and MH EMG activity as well as for VL:LH CCI, and mean VL and LH EMG. As there was a statistically significant correlation between VM:MH CCI and mean VM and MH amplitude, as well as a statistically significant correlation between VL:LH CCI and VL mean amplitude, VL:LH and VM:MH CCI were excluded from the linear regression. VIF (<10), tolerance statistics and Eigenvalues were also calculated to determine if multicollinearity was present. Cook's distance (<1) and Standardized DFBeta values were looked at to determine if there were influential cases that could have an impact on the model. Cohen's f^2 was calculated for the whole model and the individual variables as:

$$Cohen's\ f^2 = \frac{R^2_{included} - R^2_{excluded}}{1 - R^2_{included}}$$

Where for the whole model, R^2 excluded is equal to 0 as none of the variables are excluded, while for the individual variables, R^2 excluded is the calculated R^2 of the model with the independent variable of interest excluded from the model. A small, medium, and large effect is as follows: $f^2 = 0.02$, $f^2 = 0.15$, and $f^2 = 0.35$ (Cohen, 1988, 1992). Alpha values were set at 0.05 for statistical significance.

Chapter 4: Results

4.1 Descriptive Data

A total of 54 participants (males = 33; females = 21), aged 16 to 25 were included in this study. Descriptive data of the participants such as age, height, weight, body mass index (BMI), and self-reported number of years playing their primary report was analyzed (Table 3). The average BMI and number of years played were not considered to be normally distributed, thus a Mann-Whitney U-Test was performed on these variables.

Table 3: Mean and SD of descriptive data

	Age (years)	Height (cm)	Weight (kg)	BMI	Years Playing
Male	20 ± 2	$180.6 \pm 6.2^{*}$	$81.2 \pm 12.5^{*}$	$25 \pm 3^{\ddagger}$	11 ± 5
Female	20 ± 2	$167.4 \pm 5.9^{*}$	$64.5 \pm 8.6^{*}$	$23 \pm 3^{\ddagger}$	11 ± 4

Note: * represents $p < 0.01$; ‡ represents $p < 0.05$

From Table 4.3, males and females did not differ in terms of age and number of years playing their primary sport. However, male participants were statistically heavier and taller ($p < 0.01$) as well as having a higher BMI ($p = 0.012$) compared to female participants.

Table 4 displays the breakdown of number of participants participating in their primary sport as self-reported. Two participants reported soccer and basketball as being their primary sports, as they played soccer in the summer months and basketball in the winter months. Furthermore, while there is one participant whose primary sport was reported as cycling, their kinematic and EMG data did not vary significantly from the rest of the group, and so was not discarded.

Table 4: Primary sport as reported by participants

Sport	N
Soccer	21
Basketball	9
Football	9
Soccer + Basketball	2
Volleyball	2
Rugby	4
Hockey	2
Baseball	1
Ultimate Frisbee	3
Cycling	1

4.2 Objective 1

4.2.1 Knee ROM and CCI

Sagittal plane knee joint ROM, VM:MH and VL:LH CCI were compared between males and females during the preparatory and landing phases during the first landing of the SLJ task. Table 5 shows the mean and SD of sagittal plane knee ROM during both phases. Caution should be used when interpreting Cohen's d_s for the variables that were not normally distributed. There was no evidence that knee ROM, VM:MH and VL:LH CCI differed between male and female participants during the preparatory and landing phases of the SLJ task. However, a small effect size was noted for knee ROM during the preparatory phase, VM:MH CCI during both phases and VL:LH CCI during the landing phase.

Table 5: Mean ± SD, p-value, and effect size for sagittal plane knee ROM and medial and lateral CCI

	Male	Female	p-value	Cohen's d_s
Knee ROM (°)				
Preparatory Phase	43.4 ± 6.9	46.0 ± 8.4§	0.104	0.359
Landing Phase	48.7 ± 8.2	48.0 ± 7.8	0.748	0.090
VM:MH CCI				
Preparatory Phase	29.2 ± 16.8	36.6 ± 16.7	0.120	0.441
Landing Phase	47.3 ± 28.6§	63.2 ± 37.7	0.121	0.489
VL:LH CCI				
Preparatory Phase	34.2 ± 18.3	37.5 ± 15.4	0.500	0.190
Landing Phase	63.8 ± 42.2§	80.9 ± 44.4	0.125	0.397

Note: § represents that the variable was not normally distributed, Mann-Whitney U test performed.

The waveforms for knee kinematic during the preparatory and landing phase also show that there is little difference between sagittal plane knee kinematics during the preparatory and landing phase between male and female participants (Figure 5). This supports the findings in Table 5.

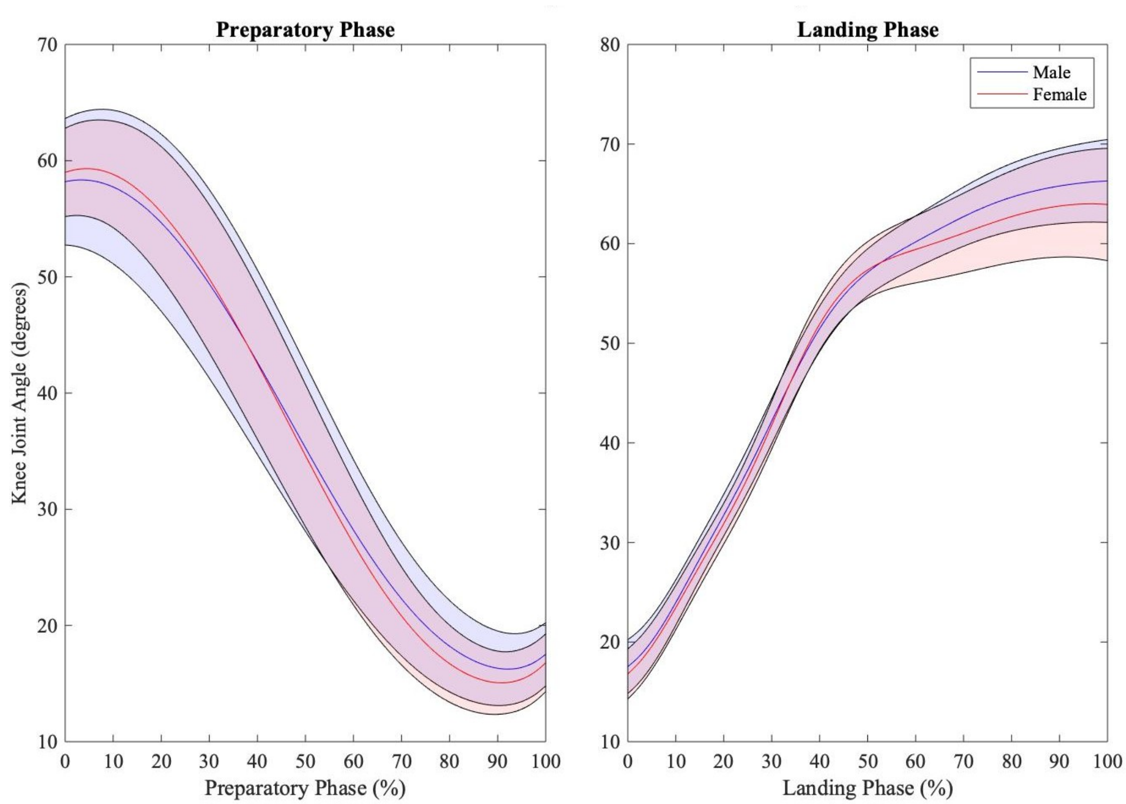

Figure 5: Mean and SD of sagittal plane knee joint angle during the preparatory and landing phase of SLJ

4.2.2 EMG

4.2.2.1 Quadriceps Muscle Group

For the preparatory phase, the Johnson Transformation failed to find an adequate equation to ensure that the VM, VL and RF data could be transformed to achieve a normal distribution. Therefore, the mixed ANOVA was performed on the untransformed data. However, for the landing phase, the VM, VL and RF EMG data was transformed using the equation -15.28 + 3.00 * Ln(X+65.53), where X represents the EMG variable.

Table 6 contains the mean and SD of mean amplitude quadriceps muscle activity during the preparatory and landing phase. There was a significant main effect for the quadriceps muscles in the preparatory ($F_{2,104} = 71.74$, $p < 0.001$, $\eta^2 = 0.58$) and landing phase ($F_{2,104} = 89.57$, $p < 0.001$, $\eta^2 = 0.63$). However, there was no main effect for sex nor an interaction effect.

Table 6: Mean and SD of quadriceps EMG during preparatory and landing phase

	Male	Female
Preparatory Phase		
VM (%MVIC)	45.4 ± 17.5	43.1 ± 22.5
VL (%MVIC)	39.0 ± 14.8	38.4 ± 16.1
RF (%MVIC)	14.5 ± 9.4	18.9 ± 11.5
Landing Phase		
VM (%MVIC)	150.9 ± 67.6	136.7 ± 70.6
VL (%MVIC)	123.8 ± 41.4	125.7 ± 48.2
RF (%MVIC)	55.6 ± 30.6	65.8 ± 28.1

Pairwise comparisons revealed that for the preparatory phase, mean amplitude of RF was significantly smaller compared to mean VM ($t(53) = 10.44$, $p<0.001$, Cohen's $d_{av} = 1.89$) and mean VL ($t(53) = 10.34$, $p<0.001$, Cohen's $d_{av} = 1.76$) muscle activity as well as there being a large effect size. A similar within muscle were seen for the landing phase (VM: $t(53) = 12.82$, $p<0.001$, Cohen's $d_{av} = 1.89$; VL: $t(53) = 10.69$, $p<0.001$, Cohen's $d_{av} = 1.85$).

Pairwise comparisons also revealed that VL was significantly smaller compared to VM ($t(53) = 2.64$, $p = 0.011$, Cohen's $d_{av} = 0.33$) during the preparatory phase and the landing phase ($t(53)=2.33$, $p = 0.024$, Cohen's $d_{av} = 0.32$).

Figures 6 and 7 are the ensemble averaged waveforms of the quadriceps muscle group during the preparatory and landing phase of the SLJ task. Visually, there is little difference in VM and VL activity during the preparatory phase, while there is a slight increase in RF muscle activity for female participants compared to males (Figure 6).

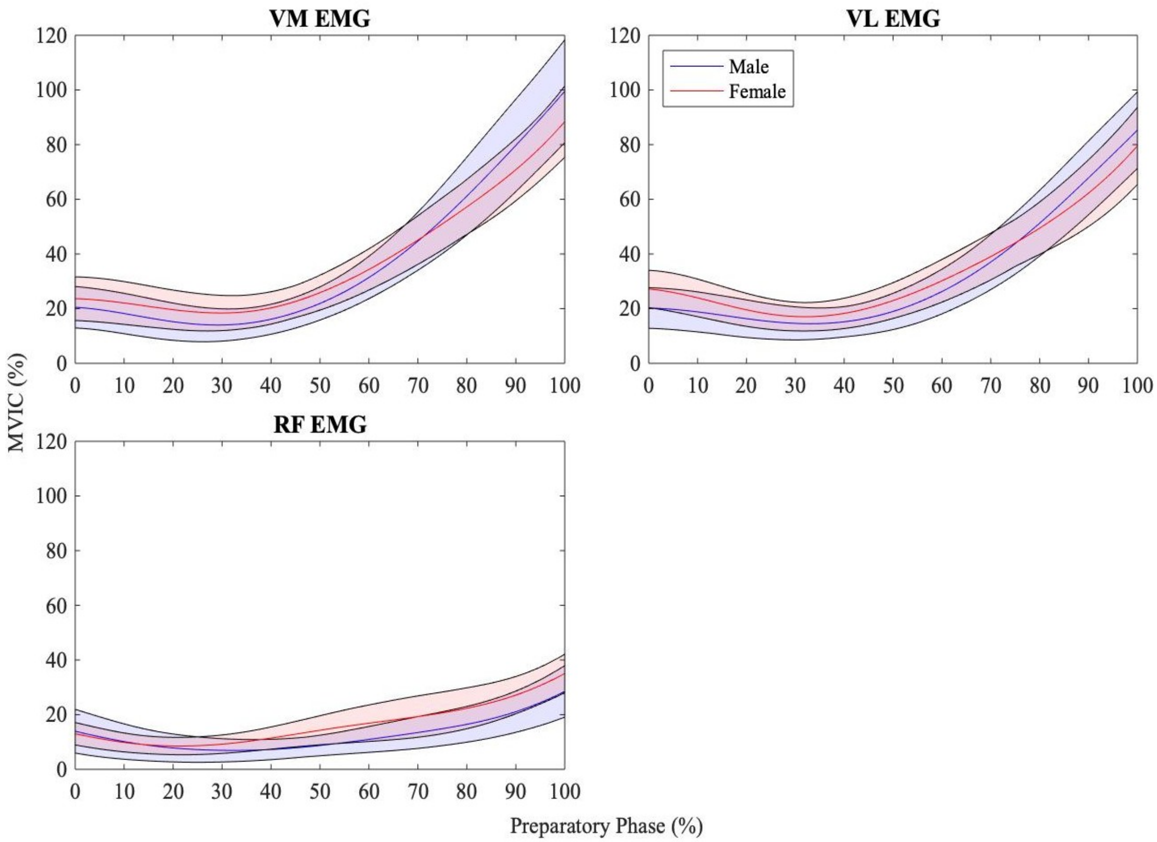

Figure 6: Mean and SD of dominant leg quadriceps muscle group (%MVIC) during the preparatory phase of SLJ

During the landing phase, VM waveforms for males were slightly higher compared to females while females had higher RF waveforms compared to males (Figure 7). Only the mean VL waveforms were visually similar between males and females. Furthermore, the variability in waveforms was greater in the landing phase compared to the preparatory phase. These waveforms support results obtained from the mixed ANOVA tests described above.

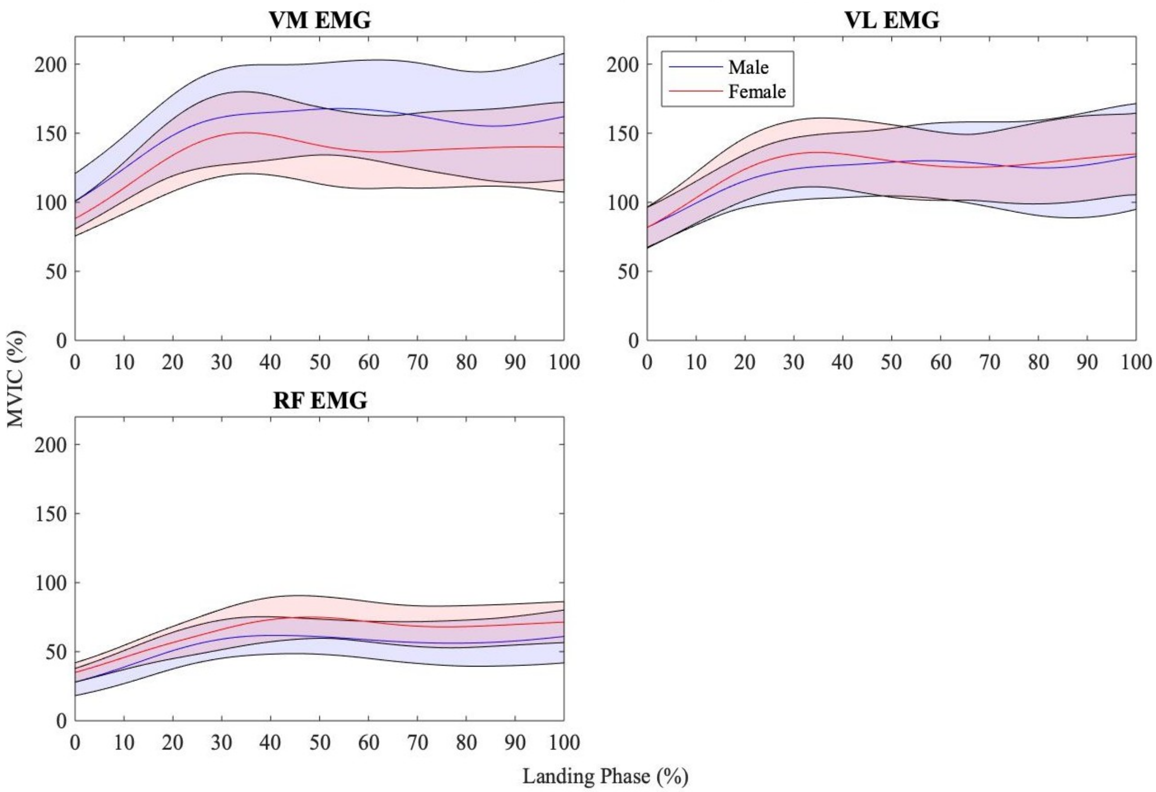

Figure 7: Mean and SD of dominant leg quadriceps muscle group (%MVIC) during the landing phase of SLJ

4.2.2.2 Hamstring Muscle Group

Johnson Transformation for the hamstring muscle group occurred for both the preparatory and landing phase, with the equation for the preparatory phase being 1.62 + 1.15 * Ln ((X-3.50)/(114.74-X)) and for the landing phase being 0.947 + 0.977 * Ln((X-5.00)/(130.11-X)), where X represents the EMG variable.

Table 4.7 contains the mean and SD of mean amplitude hamstring muscle activity during the preparatory and landing phase. A main effect for hamstring muscle during the preparatory phase ($F_{1,52} = 5.46$, $p = 0.023$, $\eta^2 = 0.095$) and the landing phase ($F_{1,52} = 5.10$, $p = 0.028$, $\eta^2 = 0.089$) was found. No main effects for sex or phase interactions occurred.

Table 7: Mean and SD of hamstring EMG during preparatory and landing phase

	Male	Female
Preparatory Phase		
MH (%MVIC)	25.5 ± 14.6	30.7 ± 16.4
LH (%MVIC)	29.8 ± 14.7	37.7 ± 19.3
Landing Phase		
MH (%MVIC)	36.1 ± 17.8	44.2 ± 21.9
LH (%MVIC)	44.4 ± 24.5	55.0 ± 25.3

Pairwise comparisons revealed that LH is statistically more active during the preparatory (t(53) = -2.39, p = 0.020, Cohen's d_{av} = 0.36) and in the landing phase (t(53) = -2.23, p = 0.030, Cohen's d_{av} = 0.37) compared to MH muscle activity.

There were no interaction effect for both phases and while the main effect for sex was not significant for both the preparatory ($F_{1,52}$ = 3.42, p = 0.070, η^2 = 0.062) and landing phase ($F_{1,52}$ = 3.81, p = 0.056, η^2 = 0.068), the p-value was close to 0.05, which suggests that there is weak evidence that females have greater hamstring muscle activity during both phases of the jump compared to male participants. Furthermore, there is a medium effect of sex during the preparatory and landing phase.

These results are supported by the ensemble waveforms as displayed in Figure 8 and 9. Figure 8 shows that females had greater LH and MH muscle activity throughout the preparatory phase. Furthermore, while the general shape of the waveforms was similar, LH activity was greater compared MH during the preparatory phase. These general findings were also reflected in Figure 9, which shows that females had greater mean LH and MH muscle activity throughout the landing phase, and that LH activity was greater compared to MH.

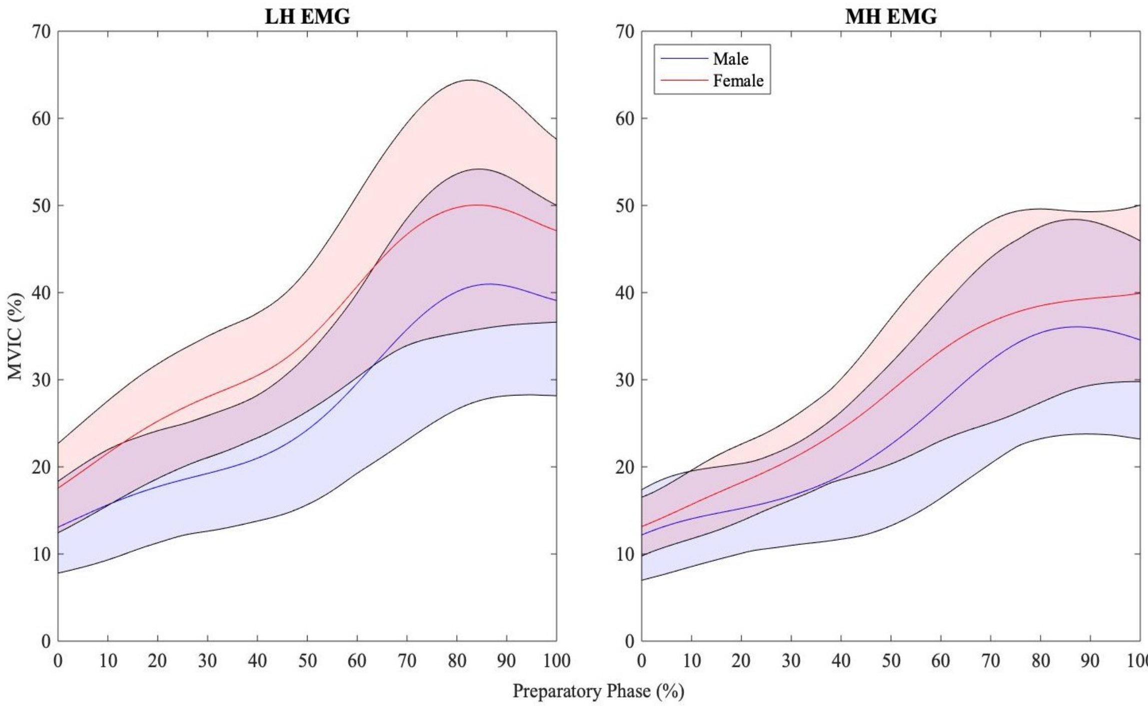

Figure 8: Mean and SD of dominant leg hamstring muscle group (%MVIC) during the preparatory phase of SLJ

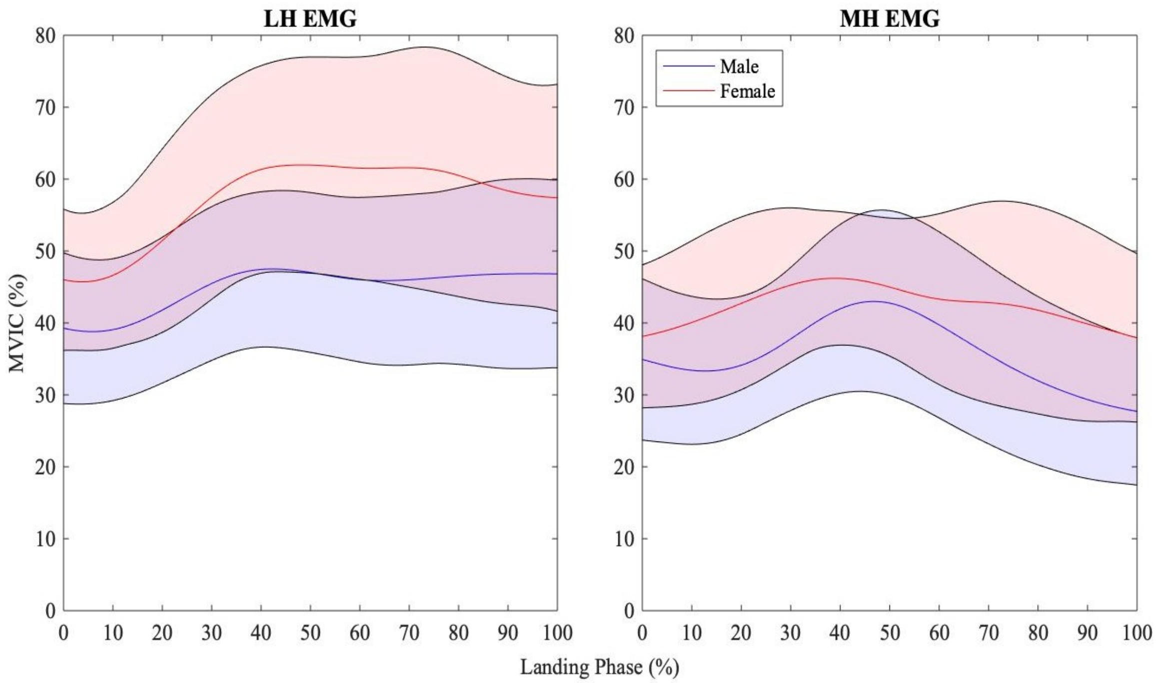

Figure 9: Mean and SD of dominant leg hamstring muscle group (%MVIC) during the landing phase of SLJ

4.3 Objective 2

Data from male and female participants were collapsed into a combined group to determine whether sex was a significant predictor of knee joint ROM during the landing phase alongside preparatory muscle activity. The backwards linear regression found that mean RF and MH EMG activity during the preparatory phase was associated with knee ROM during the landing phase. As sex was not a predictor of knee ROM during the landing phase of SLJ, linear regressions for solely the male and female participants were not performed. Table 8 is the summary table of predictors of knee ROM during the landing phase.

Table 8: Linear model of predictors of knee ROM during the landing phase

	b	Std. Error B	β	p
Constant	48.70 (43.28 – 54.11)	2.70		<0.001
RF RMS	0.22 (0.017 – 0.41)	0.099	0.28	0.034
MH RMS	-0.14 (-0.27 – -0.003)	0.067	-0.26	0.046

Note: $F=4.70$, $R=0.39$, $R^2 = 0.16$, $p=0.013$

15.5% of the variation of knee ROM during the landing phase is explained by this model (R^2), with an increase in mean RF activity and a decrease in mean MH activity during the preparatory phase of the SLJ being predictive of an increase in knee ROM during the landing phase. If mean MH EMG is kept constant, a 1%MVIC increase in mean RF EMG during the preparatory phase will lead to an increase in knee ROM by 0.22° during the landing phase. If mean RF EMG is kept constant, a 1%MVIC decrease in MH EMG during the preparatory phase will lead to an increase in knee ROM by 0.14° in the landing phase. The model's effect size was $f^2= 0.18$, which suggests that the model has a moderate effect size while the effect size for the individual predictors mean RF ($f^2=0.092$) and MH ($f^2=0.082$) EMG activity was a small effect size each.

Chapter 5: Discussion

Much research has been conducted to try and understand how differences in male and female knee joint biomechanics and muscle activity during athletic maneuvers could contribute to the sex difference in rates of non-contact ACL injuries. It has been postulated that females tend to land from a jump in a more knee extended position (Decker et al., 2003; Schmitz et al., 2007; Weinhandl et al., 2010) and utilize a quadriceps-dominance and/or ligament-dominance neuromuscular control strategy to stabilize the knee joint while landing (Hewett et al., 2010; Hughes & Dally, 2015; Nagano et al., 2007; Pappas et al., 2016; Urabe et al., 2005; Zazulak et al., 2005), which are factors that have been attributed to an increased risk of a non-contact ACL injury. However, the results from these studies are mixed and usually have not looked at preparatory knee joint kinematics and muscle activity. Furthermore, there are even fewer studies looking at how preparatory muscle activity could be associated with landing phase knee joint ROM in male and female athletes during a SLJ task. Previous research suggest that it takes 50-80ms for a motor response to occur when landing from a jump (Williams et al., 2001). As a non-contact ACL injury typically occurs within the first 40ms during a landing from a jump, this makes the preparatory muscle activity even more important to ensure that the knee can absorb the forces without an injury occurring (Koga et al., 2010). Therefore, the objective of this thesis was two-fold: a) to determine if there was a sex difference in knee joint ROM, muscle activity and Q:H CCI during the preparatory and landing phase of the SLJ, and b) to determine if sex, preparatory muscle activity and Q:H CCI was associated with knee joint ROM during the landing phase.

5.1 Objective 1

The first objective of this study was to determine whether there is a sex difference in quadriceps and hamstring EMG activity, medial and lateral Q:H CCI as well as knee joint ROM during the preparatory and landing phase of a SLJ in active participants. It was hypothesized that female participants would have smaller sagittal knee joint ROM in both phases, greater preparatory quadriceps muscle activity, smaller hamstring muscle activity and smaller medial Q:H co-contractions during the landing phase compared to male participants. The results of this present study did not support these hypotheses.

Previous research looking at jump-landings has found that females tend to go through less knee flexion when landing (Schmitz et al., 2007; Weinhandl et al., 2010) and that they have greater quadriceps muscle activity (Nagano et al., 2007; Zazulak et al., 2005) and smaller hamstring muscle activity (Ebben et al., 2010; Hughes & Dally, 2015; Urabe et al., 2005). Landing with a straighter knee increases ACL injury risk due to the increased load that the ACL undergoes (Leppänen, Pasanen, Kujala, et al., 2017), and this may partly be due to the smaller hamstring muscle activity and larger quadriceps muscle activity (Renström et al., 1986). Factors exist with the study procedures that may explain this discrepancy with the original hypotheses where the lack of standardization in height of platform, movement assessed, and even data processing across studies could have influenced the results reported (Fagenbaum & Darling, 2003; Hughes & Dally, 2015; Nagano et al., 2007; Palmieri-Smith et al., 2009; Urabe et al., 2005; Zazulak et al., 2005).

The utilization of a fixed height for the present SLJ protocol may not place the same amount of demand on male participants as it does for female participants. When considering demand, it refers to how individuals attenuate the landing impact, which is achieved by the muscles of the lower extremity and the lower extremity joints (Yeow et al., 2011). Previous studies found that females on average had a maximum jump height of 27.0-28.2cm compared to males, whose average maximum jump height ranged from 43.0-44.0cm (Weinhandl et al., 2010, 2015). When these participants landed from these heights during a unilateral jump-landing tasks in the respective studies, the female participants had greater knee extension moments (Weinhandl et al., 2015) and tended to land with a smaller knee flexion ROM compared to males (Weinhandl et al., 2010). However, when the landing height was standardized at 30cm, there was no sex difference in knee joint kinematics. In terms of muscle activity, a study conducted by Ebben et al. (2010) found that females had smaller LH muscle activity during the landing phase of a jump compared to males when the height of the jump was calculated relative to the participant's maximum jump height. In this study, females had a jump height of 43.28cm while males had a jump height of 57.61cm, which is larger compared to the previous studies conducted by Weinhandl et al. (2010, 2015). These studies suggest that absolute landing height of 30cm may be sufficiently challenging for females but not for males, resulting in similar lower extremity joint biomechanics and muscle activity when landing as it more closely mimics the landing demands that females generally experience in their daily activities compared to their male counterparts. By utilizing a fixed height of 30cm within this study, our female participants may have had greater demands placed on them, resulting in an increase in muscle activity and a change in knee joint kinematics similar to male participants, thus explaining why no sex difference in these variables were found.

One advantage to utilizing a standardized landing height is that it allows for the ease of comparing lower extremity joint biomechanics and muscle activity between studies as a 30cm landing height is a commonly used method for single-leg landing tasks. Another advantage of using a standardized landing height is that it could help standardize and control some of the task parameters, such as minimizing vertical and horizontal center of mass displacement. If different landing heights are used, it could mean that peak joint moments and vertical GRF would be normalized by the square root of the participants landing height besides being normalized by body mass to allow for the comparison between participants and to reduce individual variability. This is based on the impulse-momentum relationship and uniform acceleration motion which assumes that the average GRF during the landing phase of a jump is proportional to the square root of landing height (Hass et al., 2005; Weinhandl et al., 2010). However, this normalization may not account for all the variability in the joint moments and ground reaction forces caused by the difference in landing height if this assumption is not linear. Therefore, by using a standardized landing height, this eliminates the need for the utilization of landing height to normalize joint moment and vertical GRF.

The participants' athletic abilities may also have an influence on muscle activity and knee joint movement when landing from a jump and could explain why no sex difference was found. Some research has suggested that how collegiate level athletes move compared to recreational athletes can differ from one another, and that female collegiate level athletes tend to move more similarly to male collegiate level athletes compared to female recreational athletes. A study conducted by Morishige et al. (2019) reported that female collegiate athletes had significantly greater peak knee flexion angles compared to female recreational athletes. They also noted that recreation athletes were more likely to land in positions that increased their risk of a non-contact ACL injury compared to the collegiate athletes. Fagenbaum and Darling (2003) found that male and female collegiate level athletes move similarly to one another when landing from jumps, though females tended to land with greater knee flexion angles, either when landing from their maximum jump height or from a fixed height. However, there are some studies that contradict these findings. A study conducted by Chrisman et al. (2012) found that recreational and elite level soccer athletes had similar knee valgus alignment when landing from a jump and they concluded that the results from the elite soccer athletes was generalizable to recreational athletes. However, the result from this study should be interpreted with some caution as the participants in this study were between the ages of 11-14 years, which is younger than the studied population of this current study and the studies conducted by Morishige et al. (2019) as well as Fagenbaum and Darling (2003). It is possible that the athletes underwent training that could have potentially addressed factors that were associated with ACL injury, thus resulting in both male and female athletes moving in a similar fashion. Therefore, the participants' athletic capabilities could have an influence on how males and females perform a SLJ in a controlled laboratory setting and could explain why sex differences in knee joint ROM and muscle activity were not found.

The lack of sex difference in sagittal knee joint kinematics and muscle activity during the preparatory and landing phase of a SLJ task suggests these factors may not fully explain why non-contact ACL injuries occur in an athletic sub-population. This concurs with previous research that has shown that non-contact ACL injuries typically occur during a dynamic task under a multi-planar loading (Myer et al., 2011; Olsen et al., 2004; Quatman et al., 2010; Stuelcken et al., 2016). Preparatory muscle activity may play an important role in preventing non-contact ACL injury as such injuries typically occurs within 40ms from initial landing (Krosshaug et al., 2007). Identifying if preparatory muscle activity has an influence on landing kinematics and if sex is a factor could help with understanding how non-contact ACL injuries occur.

5.2 Objective 2

The second objective of this thesis was to determine whether preparatory muscle activity and muscle co-contractions were associated with sagittal plane knee joint ROM during the landing phase of a SLJ and whether sex explains a significant variance in this relationship. It was hypothesized that sex, preparatory quadriceps and hamstring EMG activity and Q:H CCI would be predictive of landing phase sagittal plane knee joint ROM. The hypothesis for objective 2 was partially supported. It is important to note that the regression model only accounted for 15.5% of the variance. This suggests that there may be other factors such as preparatory muscle activity of muscles not analyzed in this thesis (e.g. gastrocnemius, soleus, gluteus maximus) that could play a role in landing phase knee joint ROM.

Sex was not a variable associated with sagittal plane knee joint ROM during the landing phase of a SLJ. This is reflected also in the ensemble averaged kinematic and muscle activity waveforms (Figure 4, 5 and 7), which show that males and females tend to have similar preparatory muscle activity and knee joint movement during the landing phase. Previous research has shown similar results of no sex difference in kinematic and EMG variables during a landing task, and thus the analysis was performed on a combined group only (Walsh et al., 2012). As mentioned earlier, the differences in participants' backgrounds across studies could explain the lack of agreement in findings of sex differences.

For objective 2, it was found that an increase in preparatory RF muscle activity and a decrease in preparatory MH muscle activity was associated with an increase in knee joint ROM during the landing phase ($F=4.70$, $R= 0.39$, $R^2 = 0.16$, $p=0.013$). This result was surprising as it was thought that an increase in RF muscle activity and a decrease in MH muscle activity would result in less knee flexion to occur in the preparatory phase.

Previous research suggests that stiffer knee landings (i.e. less knee flexion) is associated with increased quadriceps muscle activity and decreased hamstring muscle activity, which are risk factors associated with non-contact ACL injuries. (Hewett et al., 2010; Krosshaug et al., 2007; Olsen et al., 2004; Walsh et al., 2012). However, a more recent study looking at the relationship between peak knee flexion angle and preparatory muscle activity found that a smaller knee flexion angle was significantly associated with greater preparatory MH muscle activity (Malfait et al., 2016).

A possible explanation for the greater preparatory RF muscle activity being associated with an increased knee joint ROM during the landing phase is that it prepares the knee for the increased load that it would experience when trying to control the descent of the body's center of mass while landing on a single leg (Brown et al., 2014; Hashemi et al., 2010). An *in vitro* simulation study found that during the initial landing phase, an increase in quadriceps muscle activity resulted in a decrease in ACL strain. This was attributed to the quadriceps muscle undergoing an eccentric contraction to help slow knee flexion (Hashemi et al., 2010). This result is supported in the study conducted by Brown et al. (2014), were they found that greater RF during the preparatory phase was associated with a decrease in external knee flexion moment. Brown et al. (2014) concluded that this increase in preparatory RF muscle activity was to help stabilize the knee joint and slow the descent of the body's center of mass. Therefore, the positive association between increased preparatory RF muscle activity and landing phase knee joint ROM could be reflecting the participant's attempts to successfully land on a single leg.

Another explanation as to why the greater preparatory RF muscle activity was found to be associated with increasing knee flexion in the landing phase was that it was causing hip flexion. As the RF is a bi-articular, it works to extend the knee and flex the hip joints. An increase in hip flexion during the landing phase is a factor that is also thought to reduce the risk of a non-contact ACL injury (Hashemi et al., 2011; Shultz et al., 2009). Shultz et al. (2009) found that there was an increase in the anterior tibial shear forces when an individual landed with a smaller hip flexion angle and larger knee flexion range of motion. This suggests that individuals who land in a more upright posture results in the vector of the resultant ground reaction force to be more posterior to the knee joint center, which results in the tibia to translate forward and place a greater anterior shear force on the ACL, resulting in an increase in the risk of an ACL rupture. If an individual lands with a flexed hip and knee, this decreases the anterior shear force on the knee, and is more protective of the ACL. Therefore, the positive association between increased preparatory RF muscle activity and landing phase knee joint ROM could be reflecting the hip joint co-flexion that occurs to help protect the ACL.

In terms of the MH, the negative association between MH and landing knee joint ROM may be due to the MH acting more on the hip than the knee joint. As hip angle can have an influence on knee flexion angle as well as ACL strain, it is possible the decreased preparatory MH activity could allow more hip flexion to occur, thus allowing for the hip and knee to flex appropriately. A modeling study conducted by Bakker et al. (2016) supports this thought, whereby higher hamstring forces were correlated with an increase in measured ACL strain. It should be noted that this study did not look at preparatory hamstring muscle activity though. However, a study conducted by Brown et al. (2014) found that an increase in LH preparatory muscle activity was associated with a decrease in peak hip flexion during the landing phase in female athletes. Therefore, while MH preparatory muscle activity may be associated to knee flexion ROM during the landing phase, it may be more at the hip than at the knee, though it is outside the scope of this present study.

An increase in MH preparatory muscle activity may also be associated with decreased knee joint flexion ROM during the landing phase because the moment arm of the MH is past its optimal length when the knee is in a more extended position (Maniar et al., 2020, 2022). Malfait et al. (2016) had similar results in their study, and they attributed this relation to the decreased optimal length of the moment arm of the MH at more extended knee positions, as more force would need to be applied by the muscles to produce the same moment around the knee joint. Previous studies have shown that there is an inverse relationship in length of the moment arms of hamstring muscles with measured hamstring muscle activity, whereby an increase in the hamstring moment arms as the knee moves through knee flexion corresponds with a decrease in hamstring muscle activity (Baratta et al., 1988; Herzog & Read, 1993). It is possible that in this present study, the increase in preparatory MH activity is associated with a decreased knee joint

ROM during landing (i.e. stiffer landing with less knee flexion) as more force is required to produce the same amount of internal moments due to the less than optimal length of the MH moment arms. However, this should be interpreted with caution as the relationship between measured EMG amplitude and force is not a linear relationship (De Luca, 1997).

The above explanation does not explain why the LH was not associated with landing knee joint ROM as both the MH and LH experiences a shortening in its moment arm at smaller knee flexion angles, and one would expect it would be associated as well. Another explanation for the increase in preparatory MH activity could also be related to coronal plane knee joint ROM. A study conducted by Pollard et al. (2010) found smaller hip and knee flexion angles when landing was associated with an increase in knee valgus angles. If an individual landed with a more extended knee and hip, there was an increase in frontal plane knee excursion to decelerate. In this present study, the negative association with preparatory MH muscle activity with landing sagittal plane knee joint ROM could be that as sagittal plane knee joint ROM decreases, there is also an increase in knee valgus movement. As the MH produces an internal varus joint moment while the LH produces an internal valgus joint moment (Maniar et al., 2020), there might be an increase in preparatory MH activity prior to landing in preparation for this knee valgus movement. As this increase in knee valgus movement during landing is associated with smaller knee flexion, this could explain this the negative association between preparatory MH and knee joint ROM during the landing phase.

Preparatory RF and MH is associated with sagittal plane knee joint ROM during the landing phase, which suggests that preparatory muscle activity plays a role in the lower extremity kinematics when landing and could help prevent ACL injuries from occurring. As sex was not a factor, this suggests that in this athletic sample, both males and females used similar preparatory muscle activity patterns prior to landing.

5.3 Limitations

This study is not without limitations. First, a limitation of this study was that only 21 female participants were analyzed in the end, which is below the calculated number of 32 female participants that were required for the study to be adequately powered. This could partially explain why there were no sex differences detected when comparing the hamstring muscle groups during both phases as the p-values were approaching 0.05, with the effect size being considered moderate. This suggests that if the sample size increased, the results could be statistically significant. Secondly, by solely analyzing sagittal knee plane ROM, quadriceps and hamstring muscle activity, it does not account for these other variables that all affect ACL loading. It is possible sex differences in these other variables exist. Research has shown that a non-contact ACL injury is a multiplanar injury that is influenced by the ankle, hip, and trunk besides just the knee joint (Ireland, 1999; Koga et al., 2010; Krosshaug et al., 2007; Stuelcken et al., 2016). Coronal and transverse plane joint kinematics were collected for the SLJ task but was not included in this thesis. Furthermore, not all muscles that play a role in modulating the load on the ACL such as the gastrocnemius and the gluteal muscle group were not analyzed in the present study (Maniar et al., 2018, 2020; Neamatallah et al., 2020). Gastrocnemii and gluteus medius muscle activities were collected for the SLJ task but was not included in this thesis due to a significant number of participants having these muscle groups exceeding 300 %MVIC during the landing phase. This suggests that participants did not perform maximal voluntary contraction when recording the muscle activity during the MVIC trials for these muscle groups. Therefore, these muscle groups were excluded from the analysis. Thirdly, the CCI was used as a method to capture co-contractions of muscle pairs during the execution of the landing task based on the equation proposed by Rudolph et al. (2000). While this CCI calculation has been used extensively to understand co-contractions, it remains limited as many different activation patterns may exist to provide the same CCI value (Hubley-Kozey et al., 2009). For objective 2, we used individual muscle activation levels in the final regression equation model, in part for this reason. Fourthly, the box height was standardized between participants. As previously mentioned, some studies suggest using individualized box heights based on the participants' maximum jump height would be more appropriate in simulating the loads they would experience during their trainings and games. However, by standardizing the box height (e.g. 30cm), this standardizes the vertical displacement of the center of mass and could potentially standardize the vertical velocity the individual may have at IC. This could potentially reduce the difference in kinetic energy which could affect the landing mechanics. Furthermore, prior studies have also used a standardized 30cm for their SLJ tasks, which would make it easier to compare the present results to their results. Lastly, results from this study cannot be generalized to recreationally active individuals which narrows the applicability and only discrete variables were statistically analyzed. By selecting only discrete variables, distinct waveform profiles could be missed between participant groups, as can be seen in this current study whereby the waveforms for the LH appeared to be different but the discrete variables were not statistically different between sexes.

5.4 Future Implications

To our knowledge, no other study exists in trying to determine if there is a difference in the association between preparatory muscle activity on sagittal plane knee joint kinematics during a SLJ task between males and females to help explain the discrepancy in ACL injury risks. The clinical implication of this study is that preparatory muscle activity plays a role in knee joint kinematics during the landing phase. As ACL injuries typically occur within the first 40ms of a jump-landing, it is important that preparatory muscle activity is sufficient to help prevent the knee from going into the "position of no return" which could lead to increased risks of injuries.

The findings of this thesis have laid the groundwork for future research, which includes:

1. This study did not find any sex difference in knee joint kinematics in the sagittal plane, however, I did not explore ankle, hip or trunk kinematics associated with the landing maneuver. Joint kinematics and kinetics have been shown to be important factors for ACL loading as well as the ACL being loaded in multi-planar fashion during non-contact ACL injuries (Myer et al., 2011; Quatman et al., 2010; Stuelcken et al., 2016). It would be important to explore whether there is a sex difference in knee joint biomechanics in the frontal and transverse plane as well as joint movements of the foot, hip, and trunk and whether preparatory muscle activity is associated to these movements.
2. As this study was done in a controlled environment, this does not mimic in-game situations. Non-contact ACL injuries often occur when an individual

is being challenged by an external perturbation, such as reacting to an opponent or being pushed-off balance when landing from a jump (Ireland, 1999; Koga et al., 2010; Krosshaug et al., 2007; Olsen et al., 2004; Stuelcken et al., 2016). External challenges such as reacting to an unanticipated perturbation while landing on a single limb might better replicate in-game situations more closely. This could potentially shed some light on differences in movement strategies between males and females when they react to these external perturbations and how it may be related to non-contact ACL injuries.

3. This study suggests that there is no sex difference in neuromuscular activation and sagittal plane knee joint biomechanics in a healthy athletic population. However, whether this is true in post-pubescent athletes who are returning to sports after an ACL injury is still a relatively novel area of research. Future studies should investigate whether preparatory and landing phase muscle activity and joint biomechanics in athletes returning to sports post-ACL injury (with a further division into those who underwent ACL reconstruction and those who did not) differ from healthy athletes.

5.5 Conclusions

The objective of this thesis was twofold: 1) to determine if there is a sex difference in sagittal plane knee joint biomechanics and muscle activity during the preparatory and landing phase of a SLJ task, and 2) to determine the relationship between preparatory muscle activity and knee joint ROM during the landing phase of the SLJ in male and female athletes. This study found that there was no sex difference in knee joint ROM and neuromuscular activity and sex was not a factor in the relationship between preparatory muscle activity and landing phase knee joint ROM in this athletic sample. Preparatory RF was positively associated while MH muscle activity was negatively associated with knee joint ROM during the landing phase, though it only accounted for a small portion of the relationship.

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APPENDIX A: Participant Recruitment Email

Email sent out to Acadia FYI and General Public

John MacIntyre Motion Laboratory of Applied Biomechanics
Longitudinal Non-Contact ACL Cutting Jumping Study

Is your child between 8 and 25 years of age and do they participate in a sport (e.g. basketball, soccer, football, rugby, lacrosse or volleyball) where they often have to perform athletic maneuvers such as side-cutting and jumping?

WHO? Acadia research students are looking for young athletes (between 8-25 years old) to participate in a research project related to sport.

WHAT? Using motion capture technology, young athletes who are interested in participating in our study will have their muscle strength and movement patterns analyzed as they perform walking, running, jumping and side-cutting maneuvers.

WHEN? Participation in this study would involve an approximately two-and-a-half (2.5) hour data collection session this summer at a time convenient for you. After testing, we may also ask you and/or your child if you would be interested in being contacted in a year's time to do follow-up testing. This follow-up testing, however, would be completely optional to you or your child.

WHERE? The John MacIntyre mLAB (motion Laboratory of Applied Biomechanics), which is located in the Acadia University Athletic Complex (Room 2010).

WHY? The long-term goal of this research is to use the knowledge gained to improve current injury prevention programs and help reduce the number of devastating sporting injuries, particularly those to the knee's anterior cruciate ligament (ACL).

Facts about the research taking place in the mLAB:

- The technology used in the lab is very similar to the motion capture technology used to make video games and animation movies:
    - https://youtu.be/6XZpVLJU8oE?t=1m35s
- We were recently featured on Rogers Hometown Hockey:
    - https://youtu.be/JCA582m7slk?t=54s

For more information please feel free to drop by the mLAB (Room 2010 in the Athletic Complex, down the hall from the Physio Clinic) or please contact one of the following individuals:

Phone (902-585-1937) or email (mlab@acadiau.ca) the mLAB and please leave a message if there is no answer.
You can also contact one of the following individuals if preferred:
Carolynn Tan (BKineH): email 132936t@acadiau.ca
Nick Cooke (BKineH): email 135688c@acadiau.ca
Chrissy Smith (BScH - Biology): email 132640s@acadiau.ca
Ellen Hatt (BScH – Biology): email 127455h@acadiau.ca
Will Sutherland (BScH – Math): email 136915s@acadiau.ca
Nick DeAdder (MSc Candidate): email nick.deadder@dal.ca
Scott Landry (Supervisor): Office Phone 902-585-1286 or email scott.landry@acadiau.ca

The Acadia University Research Ethics Board has approved this study.

APPENDIX B: Eligibility Questionnaire

John MacIntyre motion Laboratory of Applied Biomechanics
Longitudinal Non-Contact ACL Cutting Jumping Study

Collection Information:

Participant's Name:	St. Dev of Wand Length (Lab):
File Extension:	Max Residual & Cam #:
Collection Date: YY MM DD	St. Dev of Wand Length (Treadmill):
Collector(s):	Max Residual & Cam #:
*Participant's Email:	*Participant's Mailing Address
*Participant's Phone:	*Street or Box #:
	*Town/City:
	*Postal Code:

Participant Metrics:

*Sex: Male Female	Height (cm):	
*Dominant Hand (Writing): L / R	Weight (kg):	
*Dominant Hand (Throwing): L / R	Thigh Circumference (cm)	
*Dominant Foot (Kicking): L / R	Left:	Right:
*Date of Birth: YY MM DD	Calf Circumference (cm)	
*Age:	Left:	Right:
	Foot Width (cm)	
Necessary Paperwork Collected: ☐	Left:	Right:
Additional Comments:		

Questionnaire:

Playing Experience:

Primary Sport (Soccer/Basketball/Football/Rugby/Lacrosse etc): ________________.
Total years playing primary sport: __________________ years.

Position(s) most commonly played: ____________________ (e.g. winger, striker, midfield, defender, keeper, point guard, guard, forward)

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 one): 1 2 3 4 5

For the current season, answer the following for the 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 _________

Subject Name: Collection Date: Subject ID:

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?

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.

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 the above question, explain the diagnosis of the injury or pain, the date that it occurred and if it is preventing you from participation (games/practice)?

1. ______________________________

2______________________________

3. ______________________________

Subject Name: Collection Date: Subject ID:

APPENDIX C: Surface Marker Setup and Electrode Placements

Checklist: EMG and Marker Placement

EMG	RIGHT	LEFT		Measurement	Placement
Lat Gastroc	1	9	30%	Lateral joint line of knee → calcaneus	Toward lateral joint line
Med Gastroc	2	10	35%	Medial joint line of knee → calcaneus	Toward medial joint line
Lat Ham	3	11	50%	Lateral joint line of knee → ischial tuberosity	Toward ischial tuberosity
Med Ham	4	12	50%	Medial joint line of knee → ischial tuberosity	Toward ischial tuberosity
Vastus Lat	5	13	33%	Lateral joint line of knee → ASIS	30°
Rec Fem	6	15	50%	Superior part of patella → ASIS	Toward ASIS
Vastus Med	7	14	20%	Medial joint line of knee → ASIS	45°
Glute Med	8	16	50%	Iliac spine → greater trochanter	Along measurement

Marker Placement (76 total markers – 66 for moving trials and 10 virtual markers)

Trunk (13)
- ASIS (2)
- PSIS (2)
- Iliac Crest (2)
- Sacrum (1)
- Sternum: Jugular Notch
- Sternum: Xiphoid Process
- Acromion Process (2)
- Spine: T2
- Spine: Inferior Angle of Scapula

Arm (12)
- Medial Epicondyle (2)
- Lateral Epicondyle (2)
- Ulnar Styloid Process (2)
- Radial Styloid Process (2)
- Posterior Triceps (2)
- 50% of Radial Aspect of Forearm (2)

Thigh (6)
- Medial Epicondyle of Femur (2)
- Lateral Epicondyle of Femur (2)
- Greater Trochanter (2)

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

Feet (14)
- 1^{st} Distal Hallux (2)
- 1^{st} Metatarsal Head (2)
- 2^{nd} Metatarsal Head (2)
- 5^{th} Metatarsal Head (2)
- Medial Calcaneus (2)
- Lateral Malleolus (2)
- Medial Malleolus (2)

Clusters (27)
- Headband (5)
- Thigh Cluster (2x4)
- Shank Cluster (2x4)
- Heel Triad (2x3)

Removable Markers
(Following Standing Calibration)
- Medial Malleolus (2)
- Medial Epicondyle of Femur (2)
- Medial Epicondyle of Humerus (2)
- 2^{nd} Metatarsal Head (2)
- Medial Calcaneus (2)

APPENDIX D: Verbal Instructions for Standing Calibration and Movement Trials

Verbal Instructions for Trials

Standing Calibration:

"You want to start with the middle of your foot at the level of the green line. Spread your **feet shoulder width apart** and rotate your **palms forward** so they face straight in front of you. Make sure feet are also pointing straight ahead. Stand as tall as possible and stare at camera number 13. Hold that position for 5 seconds."

Wide Calibration:

"Keep your hands in the same position by your side, palms forward but **spread your feet** so one foot is in the **middle of each X** on the floor. Again, stand as tall as possible and stare at camera 13 for 5 seconds."

Move Calibration:

"I will do these movements along with you so you don't have to memorize these but, you want to stand on the square with just one of the X's. You will being with 2 right **legs swings forward and backwards**. Then you will swing your right **leg to the side** twice, and now make 2 **circles**. Now repeat this process with the left leg. Now you will do **running man arms**, **then jumping jack arms, now twist your torso, and rotate your head**."

Hip Joint Center:

"It is really important for this trial that you keep your **torso and hips as still as possible**. You can put your hands on your hips or across your chest – whatever is comfortable for you – but you can't cover any of the markers. If you lose your balance at any point just reset and continue the movement from where you left off. You're going to start with 5 **tiny leg swings** forward and backwards, then 5 **out to the side**, and five **circles**. Now switch and **repeat** with the other leg."

Double-Leg Drop Jump:

"Alright, so you're going to step up on the box here and you can see the little red marks at the tip of the box here… So you want to align the markers on your big toe with these lines. But they can't hang over the front. Then when *[insert name of person on the computer here]* says "Go" you will lean forward until you fall off the box onto the two squares with the X's. Once you land you're going to jump back up as high as possible as if you're doing a header or catching a rebound. And you want to leave the floor as quick as possible. When you land the second time you have to have one foot in each of the squares. As a side note, you can ignore the X's for these trials. If you find the balance hard its more important that you get your balance and then jump up as high as possible rather than trying to complete the second jump as fast as possible. It is also important that you don't jump off the box, but just fall. And final instruction is: when you land just hold that position for an extra second or two."
-ASK IF THEY WANT PRACTICE

Single-Leg Drop Jump:

"This is basically the same thing you just did but now you're going to do it on just one foot, so it's a bit tougher. We will start on your right foot, so you will put your right foot right in the