Summer 8-2019

The Effects of Anterior Cruciate Ligament Thickness, Knee Joint Laxity, Activity Level, Neuromuscular Imbalances and Lower Extremity Muscular Activation Patterns on Anterior Cruciate Ligament Loading

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A Dissertation Proposal Submitted to the Faculty of
Old Dominion University in Partial Fulfillment of the
Requirements for the Degree of

DOCTOR OF PHILOSOPHY in EDUCATION
HUMAN MOVEMENT SCIENCE
OLD DOMINION UNIVERSITY
August, 2019

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ABSTRACT

THE EFFECTS OF ANTERIOR CRUCIATE LIGAMENT THICKNESS, KNEE JOINT LAXITY, ACTIVITY LEVEL, NEUROMUSCULAR IMBALANCES AND LOWER EXTREMITY MUSCULAR ACTIVATION PATTERNS ON ANTERIOR CRUCIATE LIGAMENT LOADING

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There are numerous known mechanisms of anterior cruciate ligament (ACL) injury risk that span from knee joint laxity to landing mechanics. However, the relationship of these mechanisms to ACL loading during landing is unclear. In addition to landing mechanics, anthropological aspects such as ACL diameter, which can now be obtained via ultrasound, may also be an important mechanism for ACL loading. However, the relationship between ACL diameter to ACL loading and landing mechanics remains unknown. This study was conducted with three primary purposes. The first purpose of this study was to examine the inter and intra-rater reliability of using ultrasound to measure the diameter of the full ACL and anteromedial bundle of the ACL. The second purpose was to examine the relationship between the anteromedial bundle diameter and lower extremity strength, whole body anthropometrics, and previous physical activity levels. The third purpose was to examine the relationship between simulated ACL loading and pre-activation electromyography of lower extremity muscles, anteromedial bundle diameter and knee joint laxity.

Two separate raters with differing levels of experience (low: Rater 1 and moderate: Rater 2) used diagnostic ultrasound to image bilateral ACLs of twenty participants. Rater 1 also performed an additional imaging session with the same population. Interclass correlations were conducted to examine 1) intra-rater reliability between sessions for Rater 1 and 2) inter-rater reliability between the Raters 1 and 2. To address the second purpose of this study, anteromedial

bundle diameter, height, weight, strength measurements of the knee flexors, knee extensors, hip abductors, and hip adductors and previous physical activity were recorded for 17 participants. Pearson product correlations were conducted between anteromedial bundle diameter and the recorded variables. To address the third purpose of this study, 17 participants' pre-activation of lower extremity muscles, anteromedial bundle diameter and passive knee joint laxity were recorded. Next, participants performed single leg landings from a 40 cm height and a height equal to their maximal jumping capabilities. During landings, three dimensional kinematic and kinetic variables were recorded as well as electromyography (EMG) of the Rectus Femoris, Vastus Medialis, Vastus Lateralis, Medial Hamstring, Lateral Hamstring, Medial Gastrocnemius, and Gluteus Medius. Participant specific musculoskeletal models were generated for each participant. Three dimensional ACL loading was calculated using previously established equations. Pearson product correlations were then utilized to analyze the relationship of ACL loading to knee joint laxity, anteromedial bundle diameter and pre-activation of the lower extremity muscles.

Moderate correlations were found for inter and intra-rater reliabilities. There were weak correlations found for the full ACL diameter during inter and intra-rater analyses. The current results show that the anteromedial bundle can be found and measured more reliably than the full ACL. There were no significant correlations between anteromedial bundle diameter and previous exercise activity levels. However, significant correlations were found with quadriceps and adductor strength values. Simulated ACL loading was only found to have significant correlations with lateral hamstring pre-activation for both 40 cm and relative drop landing heights. The current results suggest that the pre-activation of the hamstrings may influence ACL loading if the medial and lateral hamstring groups are activated disproportionately.

The results from these studies establish that the anteromedial bundle can be analyzed with moderate reliability from researchers utilizing ultrasound. Future research involving diagnostic ultrasound should consider analyzing the anteromedial bundle to increase reliability and generalizability of their results. Although it may be possible for stressors to cause hypertrophy in the ACL, few factors known to stress the ACL are significantly associated with the size of the anteromedial bundle diameter. Thus, the association between those mechanisms and ACL loading may not be as clear.

I dedicate this work to my wife, Stephanie, and son, Frank the tank!

Thank you for everything!

ACKNOWLEDGEMENTS

First and foremost I want to thank my current advisor Dr. Hunter J. Bennett and former advisor Dr. Joshua T. Weinhandl. It's hard to describe how much I appreciate all of the time and effort you have invested in me. Each one of you have taught me more and guided more than I can really show appreciation for. I know how often I have asked questions or clarification, or drove you mad with my poor writing. I'm truly grateful for your efforts.

I would also like to thank the rest of the committee members, Dr. Staci Ringleb and Dr. Laura Hill. I have always appreciated the conversations and guidance you have given me. Over my years at Old Dominion I always felt like I could come to you for any advice or help. It has always and forever will be appreciated.

Lastly, I would like to thank my family, especially Stephanie. I know how difficult and cranky I have been over the last few months while trying to finish all of this work.

NOMENCLATURE

EMG	Electromyography
MVIC	Maximal Voluntary Isometric Contraction
ACL	Anterior Cruciate Ligament
PCL	Posterior Cruciate Ligament
MCL	Medial Collateral Ligament
LCL	Lateral Collateral Ligament
GRF	Ground Reaction Force
AM	Anteromedial Bundle of the ACL
RF	Rectus Femoris
VM	Vastus Medialis
VL	Vastus Lateralis
MH	Medial Hamstring
LH	Lateral Hamstring
MG	Medial Gastrocnemius
GM	Gluteus Medius
BW	Body Weight
40 cm	40 cm landing height
RH	Relative landing height

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CHAPTER 1: INTRODUCTION

BACKGROUND AND RATIONALE

Anterior Cruciate Ligament (ACL) injuries are devastating and can require months to years for athletes to return to full participation in athletics, if they ever return to a pre-injury state at all (Ardern, Webster, Taylor, & Feller, 2011; Kvist, 2004; Griffin et al., 2006; Langford, Webster, & Feller, 2009). Through three-dimensional biomechanical analysis of kinematic and kinetic properties of human movement, it has been shown that there are mechanical and physiological factors (ACL injury risk factors) that influence the incidence of ACL injuries, such as reduced knee flexion angles, increased knee abduction angles, increased tibial translation, increased frontal plane loading, increased anteroposterior shear knee joint force, lower extremity muscular strength, muscular imbalances, hormonal changes, ligament size as well as many other biomechanical and physiological factors (Alentorn-Geli et al., 2009; Cochrane, Lloyd, Buttfield, Seward, & McGivern, 2007; Griffin et al., 2006; Hewett, Myer, Ford, et al., 2005; Myer, Ford, Palumbo, & Hewett, 2005).

Disparity also exists between sexes in the incidence of ACL injuries. Women are three times as likely to have an ACL injury compared to their male counterparts (Petushek, Sugimoto, Stoolmiller, Smith, & Myer, 2019). Women are thought to be more predisposed to ACL injury due to increased joint laxity during their menstrual cycle, anthropometric differences, Q-angle, pelvis width, femoral notch widths, muscular imbalances and differences in muscle activation and contraction timing compared to men (Deie, Sakamaki, Sumen, Urabe, & Ikuta, 2002; Eiling, Bryant, Petersen, Murphy, & Hohmann, 2007; Hewett, Myer, & Ford, 2006; G. D. Myer, Ford, Divine, et al., 2009; G. D. Myer, Ford, & Hewett, 2005; Park, Stefanyshyn, Ramage, Hart, & Ronsky, 2009).

Although there is a disparity in ACL injuries between men and women, both sexes have similar mechanical and physiological factors that increase the risk of ACL injuries. It has been shown that joint laxity increases injury risk at the knee in both men and women (Fremerey et al., 2000; Griffin et al., 2000; G. D. Myer, Ford, Paterno, Nick, & Hewett, 2008; Shultz, Carcia, & Perrin, 2004). There are many potential causes of joint laxity that increase ACL injury risk, including a reduction in proprioception around the knee joint, muscular fatigue, muscular imbalances, activation patterns and hormonal factors (Fremerey et al., 2000; M. G. Hall, Ferrell, Sturrock, Hamblen, & Baxendale, 1995; Hurley, Rees, & Newham, 1998; McNair & Marshall, 1994; D. Roberts, Ageberg, Andersson, & Friden, 2007; Rozzi, Lephart, & Fu, 1999; Rozzi, Lephart, Gear, & Fu, 1999). The lack of proprioception may influence activation patterns of the lower extremity during dynamic tasks (Schultz, Miller, Kerr, & Micheli, 1984). As such, lower extremity muscular imbalances and muscular fatigue during dynamic activity increase ACL injury risk (Dunnam, Hunter, Williams, & Dremsa, 1988a; G. D. Myer, Ford, Barber Foss, et al., 2009; Orchard, Marsden, Lord, & Garlick, 1997). Joint laxity, muscular fatigue, and muscular imbalances allow for greater translation and rotation of the tibia during dynamic activity, placing greater ACL injury risk on the athlete. Furthermore, pre-activation patterns of the lower extremity muscles before a dynamic task can influence ACL injury (Cowley, Ford, Myer, Kernozek, & Hewett, 2006).

Measuring pre-activation patterns of lower extremity muscles can be beneficial in assessing potential ACL injury risk. Understanding the pre-activation of lower extremity muscles are key because ACL failure can occur under 50 ms after initial contact with the ground (Krosshaug et al., 2007). Understanding the pre-activation patterns of the lower extremity muscles can enhance our insight to the landing mechanics and how previous training, or previous experience performing the task, may influence muscle contractions at the time of initial contact. Past research indicates that experience level influences pre-activation of the lower extremity muscles during dynamic tasks (Medina, Valovich McLeod, Howell, & Kingma, 2008). Pre-activation of the lower extremities muscles can influence shear forces acting at the knee, potentially increasing ACL loading (Hewett, Zazulak, Myer, & Ford, 2005; Myer et al., 2005; Sell et al., 2007). This previous experience may either reduce potential injury risk or generate a potential injurious situation during landing. It has also been shown that within 50 milliseconds of initial contact with the ground, both peak GRF and peak tibial translation will take place (J. Hashemi et al., 2010; Kernozek & Ragan, 2008; Krosshaug et al., 2007; Pflum, Shelburne, Torry, Decker, & Pandy, 2004).

Since the incidence of ACL injuries per year are extremely high (Griffin et al., 2006) and costs to repair ligamentous injuries are expensive, numerous studies have been conducted to attempt reduce ACL injury mechanisms. Studies have examined the kinematic and kinetic properties of various athletic maneuvers to assess ACL injury risk and how specific physical activity or training parameters can reduce ACL injury mechanisms (Alentorn-Geli et al., 2009; Cochrane et al., 2007; Deie et al., 2002; Eiling et al., 2007; Griffin et al., 2006; Hewett et al., 2006; Hewett, Myer, Ford, et al., 2005; Myer et al., 2005; G. D. Myer, Ford, Divine, et al., 2009; G. D. Myer et al., 2005; Park et al., 2009). Multiple studies have also examined specific training techniques to improve lower extremity function and coordination, including resistance training, plyometric and dynamic stability exercises (Hewett, 2009; Hewett, Myer, Ford, et al., 2005; Hewett, Torg, & Boden, 2009; G. D. Myer, Ford, McLean, & Hewett, 2006).

Individuals that are more physically active, performing more physically dynamic tasks such as plyometrics, resistance training, or all around sport training, may have reduced lower extremity injuries (Myer et al., 2011). Individuals with lower physical activity levels prior to new dynamic activity or new training have a greater incidence of injury (Jennifer M Hootman et al., 2002; Jones et al., 1993). Since resistance, plyometric and balance training have been shown to reduce ACL injury risk mechanisms in adults, the potential for reduction of ACL individuals in youth athletics may be equivalent. It has been shown that specific warm-up routines can prevent knee injuries in youth athletics (Olsen, Myklebust, Engebretsen, Holme, & Bahr, 2005). Athletes performing 10-15 minutes of an intervention program per practice had a 71% reduction of injuries during practice and an 80% reduction of within game injuries (Wedderkopp, Kaltoft, Lundgaard, Rosendahl, & Froberg, 1999). Furthermore, performing resistance, flexibility and balance training for 30-45 minutes three times a week for six weeks was effective at reducing the number of knee injuries throughout a soccer season (Emery & Meeuwisse, 2010). Research has also shown that strength training can alter lower extremity collagenous structures such as the Achilles tendon and patellar tendon (Kongsgaard et al., 2007). The amount of time and intensity at which people exercise is directly related to collagen adaptation (Kjær, 2004). In addition, prolonged intense exercise can increase the net collagen synthesis in collagen structures (i.e. tendon and ligament), which can help those collagen structures resist greater amounts of load (Kjær, 2004). Thus, further research is needed to investigate the relationship between physical activity measures, ACL diameter and ACL loading/injury risks.

The gold standard for intracapsular examination of the knee is diagnostic arthroscopy (Mir, Sultan, & Bashir, 2015); however, the procedure is invasive and time consuming. Magnetic resonance imaging (MRI) allows medical professions the option to examine the intracapsular space of the knee with a less invasive and time consuming effort (Crawford, Walley, Bridgman, & Maffulli, 2007; Friemert et al., 2004). MRI has been shown to be viable and reliable measure for knee injuries compared to arthroscopy when assessing: osteoarthritis, menisci injuries, ACL injuries and PCL injuries (Jah, Keyhani, Zarei, & Moghaddam, 2005; Lundberg, Odensten, Thuomas, & Messner, 1996; Oei, Nikken, Verstijnen, Ginai, & Myriam Hunink, 2003; Pessis et al., 2003). Although MRI is an effective tool, less invasive and more cost effective technologies such as diagnostic ultrasound are becoming more widely available.

Diagnostic ultrasound can be a cost effective and feasible way to examine the intracapsular structures of the knee, such as the ACL (Chen et al., 2013; Mahajan, Chandra, Negi, Jayaram, & Hussein, 2015; Skovgaard Larsen & Rasmussen, 2000). Ultrasound has been shown to be reliable to diagnose injuries when compared to ultrasound and arthroscopy when diagnosing meniscal tears, medial and lateral collateral ligaments injuries, patellar tendon dysfunction, PCL injuries, and ACL injuries (Cho et al., 2001; Larsen & Rasmussen, 2000; Najafi, Bagheri, & Lahiji, 2006; C. S. Roberts, Beck Jr, Heinsen, & Seligson, 2002). Focusing on the ACL, recent work has found that ultrasound measurements of ACL are reliable compared with MRI (an intra-class correlation of 0.87; (Mahajan et al., 2015)). With the advances in ultrasound, analyses of ACL morphological parameters can be more readily available to researchers and used to relate knee mechanics with ACL parameters. Although ultrasound has been established as a reliable, less invasive measure compared to MRI, further investigation is needed. Specifically, research is needed to examine the intra and inter-rater reliability of ultrasound measures between researchers when MRI or arthroscopic comparisons are excluded/unavailable.

Although many ACL injury mechanisms have been identified from the current literature, it appears further research is required to examine the relationships between ACL loading and these previously defined variables. Specifically, further research within ACL loading could include measurements of ACL anthropometrics, muscle strength, muscle pre-activation patterns, knee laxity, and whole body anthropometrics. In addition, as the ability of ligaments to hypertrophy is unclear in humans, measurements of physical activity level could also provide important information regarding ACL morphology and loading.

SPECIFIC AIMS AND HYPOTHESES

There are 3 primary purposes of this study, addressed by separate research aims and analyses:

Aim #1: Identify the reliability of measuring the ACL through ultrasound through inter-rater and intra-rater reliability.

Working hypotheses: Measuring ACL diameter through ultrasound will produce an inter-rater and intra-rater reliability measure with a minimal ICC of 0.6 and a desired ICC of 0.8 with an α of 0.05 and β of 0.20.

Aim #2: Identify the relationship between of physical activity levels, knee flexor strength, knee extensor strength, hamstring to quadriceps and hip abductor to adductor strength ratios and anthropometric measurements to the anteromedial bundle diameter of the ACL.

Working hypotheses: There will be a strong relationship ($r \geq 0.7$) between physical activity levels, knee flexor strength, knee extensor strength, hamstring to quadriceps strength ratio, hip abductors to adductor strength ratio, height and weight to the diameter of the anteromedial bundle of the ACL. It is also hypothesized that as physical activity levels increase strength of the lower extremity muscles will increase.

Aim #3: Identify the relationship between ACL diameter, pre-activation of the lower extremity musculature and passive knee joint laxity to estimated ACL loading.

Working hypotheses: During drop landings, there will be an inverse relationship between ACL loading compared to ACL diameter and pre-activation of the lower extremity musculature. Next, there will be a direct relationship between ACL loading and passive knee joint laxity.

LIMITATIONS OF THE STUDY

1. Utilizing ultrasound measures of the ACL without a “gold standard” comparison technique such as MRI or arthroscopy. Ultrasound has been previously establish as a reliable ACL measurement tool compared to MRI.
2. Participant recollection of time spent exercising and the intensity at which the exercise was performed for the previous six months.
3. The Rajagopal musculoskeletal model only utilizes a single degree of freedom knee joint, which has been validated for estimated joint contact forces of the lower extremities.
4. The equation implemented for computing ACL loading only includes muscular contributions to the knee’s sagittal plane. Loads provided from the frontal and transverse plane are derived from inverse dynamics.

DELIMITATIONS OF THE STUDY

1. Only male participants will be included in this study. Therefore any findings may not be applicable to females.
2. This study will only test specific muscles through EMG and isometric strength testing, while it is known that other muscles will cause alterations about the lower extremity.
3. Participant’s previous experience performing the desired movements.

ASSUMPTIONS OF THE STUDY

1. Participant responses were accurate when answering all questionnaires. During inverse dynamics it is assumed that all body segments are a perfectly rigid body.
2. Participants will provide maximal effort during maximal voluntary isometric contractions.
3. During inverse dynamics it is assumed that each joint is a frictionless pin joint. During inverse dynamics it is assumed that the moment of inertia remain constant during the movement.
4. Participants will not experience peripheral fatigue from the study protocol.

SIGNIFICANCE OF THE STUDY

ACL injuries are one of the most common knee injuries among athletic populations. Research has primarily focused on ACL injuries in women or the comparison of injury mechanisms between women and men. Few studies have focused on the exact mechanisms that are associated with ACL injuries in men alone. While men are less pre-disposed for ACL injuries then women, the mechanisms that are associated with the injury in men alone are unclear. Diagnosing ACL injuries or ACL anthropometrics also has complications. Although MRI is considered the "gold standard" for the evaluation of the ACL, it is expensive and time consuming. As technology improves, diagnostic ultrasound has begun to emerge as a potential tool to examine and evaluate the ACL. Few studies have evaluated the reliability of ultrasound measures of the ACL. Therefore, accomplishing the aforementioned aims will bring further knowledge to the diagnosis and evaluation of the ACL, the relationship between ACL size and known mechanisms that increases ACL loading and the relationship of ACL loading and ACL injury mechanisms in men.

CHAPTER 2: LITERATURE REVIEW

ANATOMY OF THE KNEE

The knee is complex system within the lower extremity, which makes human movement possible. There is slight debate on the exact type of joint the knee is. For example, the knee may be considered a dual or double condyloid joint (S. J. Hall, 2012; Hamill, Knutzen, & Derrick, 2015; Loudon, Manske, & Reiman, 2013), condylar joint (Hamill et al., 2015), or modified Hinge joint (Biel, 2014; Hamill et al., 2015; Oatis, 2009; Tortora, 2012). Although textbooks may classify the joint differently, they all consider the knee to be a crucial component of lower extremity movement that allows for required mobility and extreme weight bearing activities (S. J. Hall, 2012; Hamill et al., 2015; Oatis, 2009; Tortora, 2012). The knee can have various joint classifications because of its ability to actively move in multiple planes of motion. For example, the knee can be considered a hinge, modified hinge, condylar or double condylar joint (Hamill et al., 2015; Loudon et al., 2013; Oatis, 2009).

The knee is comprised of three primary bones: femur, tibia, and patella. Some will also consider the proximal portion of the fibula as part of the knee or have influence about the knee since it serves as an attachment site for muscles that articulate movement at the knee (Hamill et al., 2015; Oatis, 2009; Tortora, 2012). The primary focus of this review of literature will be on the tibiofemoral joint, which is comprised of the tibia and the femur (Tortora, 2012), and the muscles and ligaments that place a crucial role in the stability, mobility and overall movement of the knee. There are four primary ligaments that increase the stability, maintain proper alignment, and decrease unwanted movements (Hamill et al., 2015; Oatis, 2009; Tortora, 2012). Two primary ligaments are extracapsular ligaments, medial collateral ligaments (MCL) and lateral collateral ligament (LCL). While the two primary intracapsular ligaments are the anterior cruciate ligament (ACL) and posterior cruciate ligament (PCL). The MCL is comprised of two, superficial or anterior and deep portions (Ozkan, Akalan, & Temelli, 2007; Robinson, Bull, & Amis, 2005), and new research shows a possible third portion, the ligaments and joint capsule that comprise the posteromedial capsule (Robinson et al., 2005; Robinson, Bull, Thomas, & Amis, 2006). The multiple bundles of the MCL help resist valgus forces, knee abduction, and internal and external rotation (J. C. Gardiner, Weiss, & Rosenberg, 2001; Hamill et al., 2015). The LCL is comprised single cord like bundle, which slackens as knee flexion increases and is responsible for resisting varus forces, adduction and external rotation (Hamill et al., 2015; Meister, Michael, Moyer, Kelly, & Schneck, 2000).

The ACL is responsible for countering anterior translation and medial rotation of the tibia relative to the femur. The ACL is comprised of two bundles, an anteromedial and posterolateral bundle. Amis and Dawkins (1991) found there to be a third, intermediate bundle separating the posterolateral and the anteromedial bundles. However, the focus here will be on the two dominate ACL portions, the anteromedial and posterolateral bundles.

The anteromedial and posterolateral bundles of the ACL have different functions to help support knee during flexion/extension and internal/external rotation. Each individual bundle of the ACL has different material properties, which allow for different strength and strain characteristics (Mommersteeg, Blankevoort, Huiskes, Kooloos, & Kauer, 1996). For example, the posterolateral bundle is tight or tense during knee extension while the anteromedial is tense during knee flexion. This suggests that the anteromedial bundle stabilizes the knee against anterior movement and rotation when the knee is flexed, while the posterolateral bundle stabilizes the knee at full extension (Kato et al., 2012). Thus, as the knee flexes and extends during a squat, loading of the two bundles will vary with as ACL length changes (Yoo et al., 2010). In the ACL, the anteromedial bundle is smaller than the posterolateral. In the first 30º of knee flexion, both bundles shorten and from there each bundle performs separate functions. The anteromedial bundle increases in length after 30º of knee flexion until 120º of flexion. The posterolateral bundle continues to decrease in length until 90-100º of flexion and then starts to increase in length (Amis & Dawkins, 1991).

ACL INJURY PREVALENCE

During a 10 year period following 17,307 patients, 7769 injuries to the knee were reported. Of the 7769 knee injuries, 3482 injuries involved internal structures of the knee. Out of the 3482 internal knee injuries 45.38% involved the ACL, 24.01% the medial meniscus, 17.58% the MCL, 8.16% the lateral meniscus, 2.53% the LCL, and 1.46% the PCL (Majewski, Susanne, & Klaus, 2006). Over a 16 year period, 15 NCAA sports were followed to diagnose injury rates. During this period, over 50% of all injuries were to the lower extremity and ACL injuries accounted for 3% of all injuries across all 15 sports (J. M. Hootman, Dick, & Agel, 2007). It is estimated that between 80,000-250,000 ACL injuries occur and 100,000 ACL reconstructions are performed annually in 15-25 year olds (Dick et al., 2007; Griffin et al., 2006). After an ACL injury and reconstruction a substantial amount of time may be needed for an athlete to return to pre-injury form, if they ever make it back to a high level of competition again. On average, 50% of athletes that suffer a significant ACL injury require longer than 12 months to return to competition (Ardern, Webster, Taylor, & Feller, 2011; Kvist, 2004; Langford, Webster, & Feller, 2009).

In the occurrence of a full rupture of the ACL there are many options to repair the ligament. Patients have the option for various ACL reconstruction surgeries, including: a contralateral or ipsilateral patellar graft (Biggs, Jenkins, Urch, & Shelbourne, 2009), iliotibial graft, semitendinosus and gracilis grafts (Silver, Tria, Zawadsky, & Dunn, 1991) along with single and double bundle replacement (Kaz, Starman, & Fu, 2007). After ACL surgery, with this possible 12 month recovery period, patients will require some form of rehabilitation to bring the reconstructed ACL back to the pre-injury state of the ACL or bring the reconstructed ACL to equal standards to the contralateral ACL (Biggs et al., 2009).

Although various life events may play a role in deterioration of the surrounding articulating cartilage and bone within the knee joint, ACL reconstructions have shown to cause early signs of osteoarthritis, as soon as 10-11 years post injury (Bahr & Krosshaug, 2005; Kessler et al., 2008), but most patients will experience some degree of osteoarthritis between 12 and 20 years post injury regardless of rehabilitation and reconstruction choices (Bahr & Krosshaug, 2005; L. S. Lohmander, Ostenberg, Englund, & Roos, 2004; Oiestad, Holm, Engebretsen, & Risberg, 2011; Von Porat, Roos, & Roos, 2004). Post ACL reconstruction may also lead to many other changes in the lower extremity. For instance, ACL reconstruction can alter lower extremity kinematics and kinetics and can lead to a subsequent ACL rupture.

RISK FACTORS

Through biomechanical analysis of 3D kinematic and kinetic properties of human movement, it has been shown that there are mechanical and physiological factors (ACL injury risk mechanisms) that influence the incidence of ACL injuries. These factors include: increased knee abduction angles, increased frontal plane loading, increased anteroposterior shear knee joint force, neuromuscular imbalances, fatigue, hormonal changes, ACL thickness as well as many other biomechanical and physiological factors (Alentorn-Geli et al., 2009; Cochrane et al., 2007; Griffin et al., 2006; Hewett, Myer, Ford, et al., 2005; Myer et al., 2005). During activities where side step cutting, jumping and landing are prevalent, increased ACL injury risk may be present due to neuromuscular imbalances that can lead to further altered mechanics. It has been shown that decreased neuromuscular control may lead to increased knee injuries (Ekstrand & Gillquist, 1983), more specifically ACL injuries (Boden, Dean, Feagin, & Garrett, 2000; Hewett, Myer, Ford, et al., 2005; G. D. Myer et al., 2005). Furthermore, the time to achieve peak hamstring torque and quadriceps to hamstring strength ratios may influence ACL loading, since co-contraction of the hamstring has been shown to reduce anterior translation of the tibia generated by contraction of the quadriceps (Ahmad et al., 2006; G. D. Myer, Ford, Barber Foss, et al., 2009). Research has also shown that neuromuscular fatigue may be a contributing factor in non-contact ACL injury (Chappell et al., 2005).

It has been established that athletes should have a hamstring to quadriceps strength ratio of 0.50 to 0.80 to prevent lower extremity injury (P. Kannus, 1988a; P. Kannus & Jarvinen, 1990). Others have predicted a more exact ratio of 0.60 hamstring to quadriceps strength ratio to prevent lower extremity injury (Dunnam et al., 1988a; Orchard et al., 1997). G. D. Myer, Ford, Barber Foss, et al. (2009) found that females with decreased hamstring strength compared to female and male controls had increased ACL injuries. The same female controls who did not have ACL injuries had equal hamstring strength as the male controls, but decreased quadriceps strength. also found that females had decreased hamstring strength when compared to men of equal quadriceps strength. It appears that females also do not increase hamstring strength and torque during increased velocities of an isokinetic activity as compared to males performing the same activity (Hewett, Myer, & Zazulak, 2008).

Muscular fatigue can also alter mechanics and muscle responses during activity. For example, it has been found that increased fatigue is associated with excessive external rotation (Schoenfeld, 2010). Additionally, isolated hamstring fatigue reduces joint coordination between the knee and hip (Samaan, Hoch, Ringleb, Bawab, & Weinhandl, 2015), which have been assessed to increase ACL injury risk (Pollard, Heiderscheit, van Emmerik, & Hamill, 2005).

Along with neuromuscular imbalances and muscular fatigue, reduced proprioception around a joint may decrease its functional ability. It has been shown that the ACL has proprioceptors that may influence mechanical functions of the lower extremity in injury prevention during dynamic tasks (Schultz et al., 1984). Proprioception can be defined as "the conscious perception of the orientation of different parts of the body with respect to another" or "the sense of rates of movement" (Jerosch & Prymka, 1996). It has been shown that as healthy patients age, a decrease in strength requires increased reliance on proprioception around a joint, which decreases the postural stability of the joint (Hurley et al., 1998). Evidence shows that ACL proprioceptors are more active, meaning increased signal transduction, at near full extension of the knee (Jerosch & Prymka, 1996). Furthermore, it was found that participants that had hypermobility syndrome had a lack of proprioception compared to controls. It has been suggested that this lack of proprioception can lead to undesirable mechanics during activity (M. G. Hall et al., 1995). Therefore, participants or athletes that have hypermobility may be placed at increased ACL injuries near full extension of the knee. As athletes land or cut near full extension, the lack or proprioceptive ability will alter their ability to respond to perturbations of the joint. After injury or rupture to the ACL, proprioception around the knee may be altered (D. Roberts et al., 2007). After ACL reconstruction, some patients have a full return to athletics with no additional issues, while others may return to play and continue to have instability and increased laxity in the joint (Fremerey et al., 2000). Participants with more than average joint laxity have been shown to have a decrease in the hamstring reflex, particularly the bicep femoris, when controlled forward, internal and external perturbations were applied to the lower leg during a weight bearing single leg stance (Shultz et al., 2004). It has also been shown that after participation in athletics that involve prolonged dynamic movements can increase knee joint laxity (Steiner, Grana, Chillag, & Schelberg-Karnes, 1986).

Architecture within the joint capsule may also play a role in ACL injuries. Simon, Everhart, Nagaraja, and Chaudhari (2010) concluded that steeper lateral posterior tibial slopes and smaller intercondylar notches were more pronounced in patients with ACL injuries. Narrowing of the intercondylar notch has been associated with ACL injury, where a smaller intercondylar notch would cause the ACL to excessive stretch over the inner margin during dynamic task generating failure in the ACL (Uhorchak et al., 2003) and may increase injury risk (Good, Odensten, & Gillquist, 1991). The thickness of ACL can also be predictor of ACL injury risk, as participants with a non-contact ACL injury had smaller ACL than a matched control (Chaudhari, Zelman, Flanigan, Kaeding, & Nagaraja, 2009; Mahajan et al., 2015). It seems as though participants that had ACL injuries had a negative relationship between ACL diameter and weight. showing ACL diameter decreased with increased weight. However, the healthy matched controls had a positive relationship with ACL diameter to weight, showing increased ACL diameter increased with increased weight (Mahajan et al., 2015). Individuals who have suffered a non-contact ACL injury have been shown to have smaller ACL volume than healthy matched controls (Chaudhari et al., 2009; Simon et al., 2010), as well as a steeper lateral posterior tibial slope. Typically it has been found that women have smaller ACL thickness than men (A. F. Anderson, Dome, Gautam, Awh, & Rennirt, 2001; N. Chandrashekar, Slauterbeck, & Hashemi, 2005; Charlton, St John, Ciccotti, Harrison, & Schweitzer, 2002; Dienst et al., 2007), and that the size of the ACL may be correlated to height (Charlton et al., 2002). However, when ACL size is corrected for lean body mass there is no significant difference between men and women (A. F.

Anderson et al., 2001). It has also been found that as height of male participants increases the size of the ACL increases, but this correlation may not be present in female participants (A. F. Anderson et al., 2001). When male and female participants are matched for height, women have significantly smaller ACL thickness (Dienst et al., 2007). Some research has shown that ACL thickness is correlated to the weight, and not height (Chaudhari et al., 2009; Mahajan et al., 2015). Furthermore, it has been observed that athletes with stronger quadriceps muscles had larger ACL thickness (A. F. Anderson et al., 2001). It was found that during ACL reconstruction, graft thickness and diameter was correlated to both height and weight (Thomas, Bhattacharya, Saltikov, & Kramer, 2013). Although, ACL size and strength has been associated with height and weight, the relationship of estimated ACL loading and ACL size has yet to be established. Further research is needed to establish is ACL size is a factor to further consider for ACL injury risk during dynamic movements.

PRE-ACTIVATION

Olsson, Jonsson, Larsson, and Nyberg (2008) have shown through functional magnetic resonance imaging that novice and experienced athletes will recruit different portions of the brain when asked to visualize the completion of an athletic task. The utilization of various portions of the brain may initiate different mechanics during actual athletic tasks. During athletic maneuvers, novice and experience participants also used different muscular recruitment strategies that generate different joint moments in the lower extremity (Sigward & Powers, 2006). It has been suggested that the less experienced participants recruit a greater co-contraction pattern for a more protective strategy. Furthermore, previous experiences can influence mechanics to activities before they take place, such as pre-activation of muscle around joints or reduction in joint moments (Beard, Kyberd, Fergusson, & Dodd, 1993; Besier, Lloyd, Ackland, & Cochrane, 2001).

During landing maneuvers, pre-activation of the lower extremity musculature can influence mechanics during landing (Cowley et al., 2006; Gehring, Melnyk, & Gollhofer, 2009; Hewett, Zazulak, et al., 2005). It has been found that men and women utilize different pre-activation strategies when performing drop landings (Gehring et al., 2009; Zazulak et al., 2005). Men were found to activate both the medial and lateral quadriceps, while also activating the lateral hamstring earlier than the medial hamstrings. Medina et al. (2008) demonstrated that pre-activation differences were observed between athletes and non-athletes, but did not find significant differences in activation timing between men and female athletes. Medina et al. (2008) also found that there was no difference in activation timing of the medial and lateral hamstring in male participants. The peak and integrated signal as well as the timing of the pre-activation of the lower extremity musculature during dynamic activities play a large role in the dynamics of landing. As athletes prepare for the landing, depending on gender and previous experience the activation of the lower extremity can change altering landing mechanics. This alteration of muscle pre landing activations may prepare the athletes lower extremity to handle force differently during landing thus altering ACL loading during landings. The extent of the relationship between pre-activation of the lower extremity and ACL loading is lacking from the current literature.

EVENT TIMING

It is thought that during dynamic activity excessive anterior translation of the tibia caused by forces generated by the quadriceps place the ACL at greater risk of injury (Withrow, Huston, Wojtys, & Ashton-Miller, 2006). It has been shown that during a drop landing the peak GRF happens between 40-57 ms (Kernozek & Ragan, 2008; Pflum et al., 2004), while ACL injury or failure can take place between 17-50 ms (Krosshaug et al., 2007). It has been shown that the force generated by the quadriceps during a landing maneuver may not be the reason for excessive anterior translation of the tibia that result in ACL injury (J. Hashemi et al., 2010). Furthermore, during drop landings, fluoroscopy measurements show peak anterior tibial translations occur within 40-50 ms after initial contact, with peak GRF occurring around the same time frame (Torry et al., 2011).

Raunest, Sager, and Burgener (1996) has shown that as load is applied to the ACL, proprioception from the ligament will increase activation of the hamstrings. The time frame between initial contact and peak ACL force and potential ACL injury is extremely small, small enough that conscious effort to counter movement is nearly impossible. It has been shown that it can take 89 ms to initiate a signal from intracapsular cruciate ligaments to induce a muscular response (Raunest et al., 1996). Muscle spindle response to achieve muscle response can take 50-200 ms (Herzog, Scheele, & Conway, 1999). Komi (2000) suggest that an initiated response from a stretch in the muscle spindle can take 40 ms, while a muscular response will take an additional 50-55 ms. This timing information paints a picture that mechanics after about 50 ms after initial contact may not influence ACL injury during drop landings. This may suggest that ACL injury risk is associated with pre-activation of the lower extremity muscles, skeletal geometry, previous experience (previous physical activity) or training, and ACL thickness. The association between the aforementioned variables and ACL loading can provide more insight to the exact mechanisms that may contribute to ACL injuries.

EXERCISE

During resistance training, the body will go through many neurological and morphological changes, some understood well, while others are not completely understood. During resistance training programs, the goal is to increase neuromuscular control, muscular strength, or muscular size. It has been shown that increases in muscular strength and neuromuscular control can reduce injury risk. A simple expression of muscle force can be equated to $F_m = PCSA*\sigma$ (Enoka, 2008; Hamill et al., 2015; Kawakami, Abe, Kuno, & Fukunaga, 1995; McGinnis, 2013), or force is equal to the physiological cross sectional area times the specific tension of the muscle. During training, there are two ways to increase muscle force. The first way is to increase the PCSA of the muscle and the second is to increase the stimulation to the muscle by either strength of signal or rate coding (Enoka, 2008; Hamill et al., 2015; McGinnis, 2013). To increase neuromuscular control, participants may include training of the proprioceptors around joints to increase motor control of joints by performing plyometric exercises.

Research among adults has shown that during the first six weeks of resistance training there will be a dramatic increase in the neurological factors, such as: increased coordination of the central and peripheral nervous systems, neural stimulation to the agonist from the central nervous system (CNS) (Aagaard et al., 2000; Van Cutsem, Duchateau, & Hainaut, 1998), motor unit activation, increased conduction velocity and increased firing of the neurons (Folland & Williams, 2007; Kenney, Wilmore, & Costill, 2015), rate coding to the agonist group (Rich & Cafarelli, 2000), , inhibition of the antagonist muscles, synchronization of the signal from the CNS peripheral nervous system (Folland & Williams, 2007; P. F. Gardiner, 2011; Kenney et al., 2015; Staron et al., 1994), and decreased inhibition of the agonist group (Aagaard et al., 2000).

As individuals complete resistance training programs and strength increases have been observed, less muscle activation will be needed to complete movements at similar loads from pre intervention (Ploutz, Tesch, Biro, & Dudley, 1994). After a resistance training program EMG activity of the lower extremity will be increased during maximal effort exercises (Aagaard, Simonsen, Andersen, Magnusson, & Dyhre-Poulsen, 2002; Aagaard et al., 2000). Furthermore, it was also observed that resistance training can reduce coactivation of the antagonist muscle during isometric contractions (Carolan & Cafarelli, 1992). Along with changes in coactivation, there is evidence that suggests a decrease in Golgi tendon organ reflex of the agonist muscle during contraction after training (Aagaard et al., 2000). With training, alterations of activation patterns can change, this changes may play a key role in injury mechanisms during dynamic activities.

With prolonged resistance training beyond six weeks, physiological and mechanical changes can be seen within the muscle, including: increase physiological cross sectional area (PCSA) through increase in size and number of fibers, increase in pennation angle, increase in enzyme number and size, and increased energy storage (P. F. Gardiner, 2011). It has also been found that prolonged dynamic resistance training can increase contractile velocity of the trained muscle group (Van Cutsem et al., 1998).

ALTERATIONS TO KNEE LOADING

The effects of resistance training on the knee is complicated and currently not comprehensively understood because of the coordinated actions of multiple lower extremity muscles and joints to perform dynamic activities. Numerous studies have shown training protocols targeting resistance training, plyometric and dynamic stability exercises can reduce ACL injury rates in athletes (Hewett, 2009; Hewett, Myer, Ford, et al., 2005; Hewett et al., 2009;

G. D. Myer, Ford, McLean, et al., 2006). Hewett, Lindenfeld, Riccobene, and Noyes (1999) demonstrated that untrained female athletes are 2.4 to 3.6 times more likely to obtain a serious knee injury compared to female athletes who participated in a six week training program which consisted of: flexibility, plyometric and resistance training. Neuromuscular training that consisted of resistance, plyometric, core, balance and speed training can alter kinematics, particularly at the knee, of young athletes to reduce injury risk (Myer et al., 2005). Next, it has been shown that plyometric training alone can alter kinematics during athletic maneuvers to reduce injury risk in young athletes, and helps correct a quadriceps to hamstring imbalance within the non-dominate limb (Hewett, Stroupe, Nance, & Noyes, 1996). Snyder, Earl, O'Connor, and Ebersole (2009) found that after a six week resistance training program lower extremity kinematics and joint moments during running were altered from pre-training values, specifically reducing knee abduction moments. It may seem that physical training may alter forces during athletic maneuvers, but the specificity of training may also play a role in ACL injury prevention.

During a 7 week training program, diagnosed high risk participants were able to reduce knee abduction moments while predetermined low risk participants did not alter any knee abduction moments (G. D. Myer, Ford, Brent, & Hewett, 2007). Completing a resistance training program has the ability to reduce forces and moments experienced during athletic maneuvers (Irmischer et al., 2004). During a two year program where sport specific stretching, resistance and plyometric training was administered as a warm up, the incidence of ACL injuries were reduced between the experimental groups. It was found that the stretching, plyometric and resistance training resulted in a significant decrease in ACL tears for the intervention group, two ACL injuries within the first year, where the control group with no intervention sustain 32 ACL tears (Mandelbaum et al., 2005). Herman et al. (2008) found that after 9 weeks of just resistance training, lower extremity joint kinematics and kinetics were not altered. This information shows that the program of the resistance training, the type of resistance training, and the length of the resistance training play a role in the alterations of lower extremity mechanics during dynamic tasks.

TENDON AND LIGAMENTOUS CHANGE DUE TO EXERCISE

Living tissue has the ability to remodel itself, whether from inactivity, a decrease loading of the structure, or a loading of the tissue (Buckwalter & Grodzinsky, 1999; Hayashi, 1996). It has been shown that exercise can alter ACL strength and stiffness while maintaining functional ability during exercise in rats (Cabaud, Chatty, Gildengorin, & Feltman, 1980). It was also found that the more frequently the rats were exercised, the more improvement in the ACL there was (Cabaud et al., 1980). In humans, exercise, cardiovascular training or resistance training, has the ability to induce change, strength, thickness, and organization of the collagen within ligaments and tendons (Buckwalter & Grodzinsky, 1999; Butler, Grood, Noyes, & Zernicke, 1978; Hayashi, 1996), which may be explained through Roux's law of functional adaptation (Scheffler, Unterhauser, & Weiler, 2008). Through loading of the tendons and ligaments, alterations to collagen formation, proteoglycans and glycosaminoglycan production may be responsible for alterations observed within the tendons and ligaments (Buckwalter & Grodzinsky, 1999; Hayashi, 1996). It has been shown that ligament changes (length and thickness) occurs during growth and repeated stresses to the ligament in rabbits and rats (Cabaud et al., 1980; Dahners, Sykes, & Muller, 1989; Wood, Lester, & Dahners, 1998). Inversely, immobilization of the knee in rabbits and primates showed a reduction in net collagen in the ACL and tensile strength (Hayashi, 1996; Herpin, Raynier, Boutaud, Amiel, & Demange, 1990). Therefore, it is possible an increase in size can increase the tensile strength of the ACL. In humans it has been shown that tendon stiffness may be alter due to resistance training (Kubo, Kanehisa, & Fukunaga, 2002; Markovic & Mikulic, 2010). Increases in patellar tendon thickness have been observed after participants performed heavy knee extension exercises for 12 weeks (Kongsgaard et al., 2007). During the same intervention, it was found that as the tendon did alter in thickness, the changes along the patellar tendon were different (Kongsgaard et al., 2007). The Achilles tendon has also been shown to have adaptations to occur from resistance and plyometric training (Kongsgaard, Aagaard, Kjaer, & Magnusson, 2005; Kubo, Kanehisa, Ito, & Fukunaga, 2001; Lenskjold et al., 2015; Magnusson & Kjaer, 2003; Rosager et al., 2002). Research has shown that exercise has the ability to alter tendon thickness, but the exact relationship between tendon hypertrophy and muscle hypertrophy remains unknown. It has been suggested that as muscle hypertrophies, tendon will hypertrophy as well, the rate at which each will happen has yet to be determined. This increase in cross sectional area of the tendon may allow the tendon to resist more applied forces (Kongsgaard et al., 2007). It has been suggested that as continual strain has been applied to the ACL, it has the ability to hypertrophy as well (Simon et al., 2010).

ANALYZING LIGAMENT STRUCTURE AND FUNCTION

As technology advances new cost effective techniques are becoming available to examine soft tissue structures and diagnose injuries pertaining to joints (Chew, Stevens, Wang, Fredericson, & Lew, 2008; Larsen & Rasmussen, 2000; Lew, Chen, Wang, & Chew, 2007; C. S. Roberts et al., 2002). Although arthroscopy is the gold standard diagnostic tool to diagnose intra-capsular damage (Crawford et al., 2007; Friemert et al., 2004), other non-invasive options may be feasible. Using MRI to evaluate intra-capsular ligaments of the knee is a less invasive and more feasible option before utilizing arthroscopy (Crawford et al., 2007; Friemert et al., 2004). Although MRI is more feasible to diagnose the damage to structures of the knee than arthroscopy, cartilage damage within the joint is more reliably diagnosed through arthroscopy (Friemert et al., 2004). Again, as technology progressed, MRI started replacing arthroscopy to diagnose and examine ACL injuries. Now if may seem as though diagnostic ultrasound is progressing forward to become more readily available tool to diagnose ACL injuries (Chen et al., 2013; Dinnes, Loveman, McIntyre, & Waugh, 2003; Mahajan et al., 2015; C. S. Roberts et al., 2002), particularly ACL injuries (Chen et al., 2013; Friedl & Glaser, 1991; Z. Khan, Faruqui, Ogyunbiyi, Rosset, & Iqbal, 2006; Larsen & Rasmussen, 2000; Mahajan et al., 2015; Suzuki et al., 1991). Ultrasound is a viable, cheaper alternative method to locate and analyze the size or thickness of the ACL at the tibial insertions site (Chen et al., 2013; Mahajan et al., 2015). Although studies have shown the feasibility of diagnosing an ACL injury with ultrasound, research is lacking on the ability of ultrasound to measure and examine the ACL. Further research is needed to establish that ultrasound is a reliable means to examine and measure the ACL. It also needs to be determine what is inter and intra rater amongst various researchers when utilizing ultrasound to examine ACL diameter.

STATIC OPTIMIZATION AND OPTIMIZATION CRITERIA

During human activity there are multiple complex systems operating independently and in coordination with other systems. The coordinated actions of the central nervous, muscular and skeletal systems make dynamic motions possible. Examining human movement and the forces that drive this movement is critical for multiple disciplines; however, examining in vivo joint contact forces and muscle forces can be invasive and unnecessary, for some clinical evaluations (Erdemir, McLean, Herzog, & van den Bogert, 2007). The use of modeling and simulation can give researchers the ability to estimate forces acting within the body and guide conclusions to how forces influence human movement. The information from modeling and simulation can also provide valuable information to address poor mechanics or alter rehabilitation mechanisms to improve patient outcomes.

When examining human motion, external forces are utilized to estimate net joint moments through inverse dynamics (Delp et al., 1990; Erdemir et al., 2007; Lin, Dorn, Schache, & Pandy, 2011; Zajac, Neptune, & Kautz, 2002). Although inverse dynamics is a useful tool to evaluate joint moments, the internal muscle forces driving the movement remain unknown. However, static optimization can derive muscle forces from the joint motions and moments. Static optimization is performed using musculoskeletal models that incorporate physiologically relevant muscle properties and activation (F. C. Anderson & Pandy, 2001b; Delp et al., 2007; Erdemir et al., 2007). Opensim (Delp et al., 2007), an open source musculoskeletal modeling software, offers users multiple options to analyze human movement through various techniques from static optimization (Erdemir et al., 2007) to forward dynamics (Erdemir et al., 2007) and computed muscle control (Delp et al., 2007; Thelen & Anderson, 2006). With many options to utilize, multiple parameters must be taken into consideration to choose the appropriate optimization technique. For example, computational time varies between muscle activation estimation tools. Static optimization can be much more computationally efficient tool, while producing sufficient results in less dynamic tasks (F. C. Anderson & Pandy, 2001b) and during some more dynamic tasks such as running (Lin et al., 2011). Forward dynamics, computed muscle control and RRAs can be much more time consuming and may not produce significantly different results than static optimization.

To solve for muscle activations and forces using static optimization, the joint moments must be first solved with an inverse dynamics approach. Inverse dynamics is calculated from the experimental kinematic and force data, solving for the net joint moment about each joint of interest. The net joint moment is representative of all the muscle forces acting at that specific joint (Erdemir et al., 2007). From the net joint moment, muscle forces are estimated at each instance of time during the motion using a musculoskeletal model with predefined muscle lines of action and muscle force parameters. Individual muscle forces are solved using linear or non-linear algorithm based upon a chosen criterion. Since the conception of modeling and simulation, many optimization criteria have been proposed. An optimization criteria must be used because the known joint moments produce an infinite number of possible muscle force solutions (Penrod, Davy, & Singh, 1974; Seireg & Arvikar, 1973).

Estimating muscle forces through static optimization with different criteria was first proposed by Seireg and Arvikar (1973), who proposed that the criteria can used singularly or as a weighted combination. Seireg and Arvikar (1973) provided examples of possible optimization criterion, such as minimizing: muscular force, muscular work, vertical reaction force through all lower extremity joints, and moments through all lower extremity joints. It was concluded that since the potential optimization criterion are linear objectives, the solution to estimate muscular force can be solved with a linear solution. The estimated muscle force must also be related to their cross sectional area. Seireg and Arvikar (1973) first proposed this optimization for simple motions such as standing upright, stooping, and forward and backward leaning. Seireg and Arvikar (1975) continued their previous work by devising a model that could predict muscle forces during standing and walking.

Penrod et al. (1974) proposed a linear algorithm with what the authors propose as the optimal response principle, where the computation of the number of muscles attributing to the force need cannot be greater than the joints rotational degrees of freedom. Within this linear algorithm, muscle effort (force) is minimized. However, this proposed solution is limited in predicting muscle forces. Penrod et al. (1974) recognized that within the solution, only two of four possible muscles were being activated to complete most tasks, while the fourth muscle within the system wasn't activated in the completion of any task. The two muscle concept didn't remain constant with the proposed optimal response theory because previous research shows that multiple muscles, are active during the resistance of opposing moments. Furthermore, the model has infinite force potential from the muscles. Penrod et al. (1974) also stated that a limitation to the linear algorithm is that it may not be useful in a healthy population. proposed multiple problems that can arise from the previously defined? static optimization. The primary concern being that a linear algorithm approach with time independent calculations does not represent true human motion. Thus far, these models don't take into account various physiological demands.

Crowninshield (1978) suggested a different optimization criteria, minimizing muscle stress. By minimizing muscle stress and placing maximal muscle force constraints in the model, a more synergistic physiological approach may be utilized in predicting muscle forces. As the muscle force constraints increase, a more synergistic solution was theorized to take place. When minimizing muscle stress and placing force production constraints, the model recruited muscles with larger physiological cross sectional areas and the larger moment arms. However, since force production constraints are in place, the muscle with the next largest physiological cross sectional area will be recruited as well, which better represents muscle recruitment patterns.

Pedotti, Krishnan, and Stark (1978) estimated muscle forces from both previously derived and new optimization criteria, which include the minimization of: sum of muscle force, sum of squares of muscle forces, the sum of the ratio of actual force to maximal force capabilities, and the sum of squares of the ratio of actual force to maximal force capabilities.

When all four criteria are evaluated together over 11 lower extremity muscles used during walking, the differences in force estimation in the tibialis anterior, soleus, vastus, and iliacus were similar across optimization criterions. Within the Gluteus Maximus and Gastrocnemius, the criteria of minimizing the ratio of actual force to maximal force and the minimizing of the sum of squares of the ratio of the actual force to maximal force showed a larger estimation of force. The semimembranosus, semitendinosus, and biceps femoris showed a lower estimation of force from the criteria of minimizing the ratio of actual force to maximal force and the minimizing of the sum of squares of the ratio of the actual force to maximal force. Depending on the optimization criteria chosen, the estimation of muscle force can change.

Crowninshield and Brand (1981) state that linear optimizations may be utilized out of convenience rather than physiological properties of muscle. and that it may be more appropriate to use nonlinear optimization criteria to more closely match physiological properties. Crowninshield and Brand (1981) utilized an optimization criteria of muscle endurance, which is muscle force over cross sectional area (muscle stress), over a time period. This study examined the influence of the nonlinear solutions of the summation of muscle stress raised from powers of 1,2,3,4, and 100 around the elbow, and the summation of muscle stress cubed within a lower extremity model. Around the elbow, muscle stress raised to the 2,3, and 5 showed the ability to lower muscle stress and recruited more muscles to meet the demand placed on the system. The authors suggested that muscle stress cubed is a sufficient optimization criteria, but also that powers of 2 and 4 may be appropriate as there were slight variations in muscle force predictions between powers of 2, 3, and 4 but no changes within the number of muscles being activated. The use of musculoskeletal modeling can be an extremely useful tool to estimate joint moments, joint contact forces, and muscle forces. Depending on the optimization criteria, different outcomes can be given. Based on previous works by Crowninshield and Brand (1981), muscle stressed squared may be the most appropriate option to utilize.

MUSCULOSKELETAL MODELS

There are numerous musculoskeletal models that can be utilized when performing static optimization. A commonly utilized lower extremity model utilized in musculoskeletal modeling is Gait2392 (F. C. Anderson & Pandy, 1999, 2001a; Delp et al., 1990; Yamaguchi & Zajac, 1989). Gait2392 is an 8 segment lower extremity model that has 23 degree of freedom and 92 musculotendon actuators and utilizes the Thelen2003 muscle model (Thelen, Anderson, & Delp, 2003). The hip was designed as a ball and socket joint with 3 DOF. The knee is a one DOF joint that accounts for translation of the tibiofemoral joint as a function knee flexion and extension angles. The ankle was designed as a 1 DOF joint where the metatarsophalangeal and subtalar joints can be locked (Fox & Delp, 2010). The Thelen2003 muscle models are developed as a series of lines from origin to insertion points. Via points are used to give the muscle the ability to wrap around anatomical landmarks when necessary. Although Gait2392 is useful for less dynamic tasks such as walking, the muscle geometry within the model does not allow for higher flexion angles at both the knee and hip joints.

A more advanced model is the Lower Limb Model 2010, which is a 14 rigid body segment system that includes 44 Hill type musculotendon compartments in the Shutte1993 muscle model (Schutte, Rodgers, Zajac, & Glaser, 1993). The pelvis has 6 DOF (rotations and translations in each plane of motion), the hip was designed as a 3 DOF ball and socket joint, the knee is a 1 DOF joint flexion and extension joint with a maximum flexion angle of 100°, and the ankle is a 1 DOF joint. The standard Lower Limb Model 2010 has the metatarsophalangeal and subtalar joints locked. Within the model there are 17 ellipsoidal wrapping objects around the hip and knee to improve the accuracy of the muscle moment arms, physiology and muscle paths in the lower extremity. The adjustments made by Arnold, Ward, Lieber, and Delp (2010) in The Lower Limb Model 2010 were designed to provide a better representation of muscle parameters of force-length and muscle architecture compared to Gait2392. Although improvements were made within the model's ability to generate physiological conditions with wrapping objects, the model is still limited by the amount of hip and knee flexion that can be utilized during dynamic activity. Furthermore, the model utilizes muscle parameters from elderly cadavers, which may not be representative of younger populations that researchers tend to utilize.

The Rajagopal Model (Rajagopal et al., 2016), is comprised of 22 rigid body segments and 80 musculotendon Hill type compartments in the lower extremities within the Millard Equilibrium model (Millard, Uchida, Seth, & Delp, 2013). Muscle properties were based off of 21 cadaver specimens and 24 young healthy young individuals' MRIs. Like The Lower Limb Model 2010, The Rajagopal Model utilizes wrapping objects as well. However, all wrapping objects are cylindrical instead of ellipsoidal objects. The pelvis is designed to have 6 DOF of rotations and translations in each plane of motion, the hip was designed as a 3 DOF ball and socket joint, the knee is a 1 DOF joint, and the ankle, subtalar and metatarsophalangeal joints were designed as pin joints. During simulations, the authors set the subtalar and metatarsophalangeal joints at 0 degrees. The Rajagopal Model has increased joint range of motions. The hip allows for 10 degrees more extension, 30 degrees more flexion, 10 degrees more abduction and 20 degrees more adduction. At the knee the Rajagopal Model allows for 20 degrees more flexion and at the ankle allows for 10 degrees more dorsiflexion.

ESTIMATING MUSCLE, CONTACT AND LIGAMENTOUS FORCES

Musculoskeletal models can be utilized to calculate joint moments through inverse dynamics and use these moments to estimate muscle forces, followed by joint contact forces and ligamentous forces (F. C. Anderson & Pandy, 2001a; Delp et al., 2007; Erdemir et al., 2007; Lin et al., 2011; Thelen & Anderson, 2006; Wesseling et al., 2015). Importantly for investigations of ACL injuries and injury mechanisms, musculoskeletal modeling can be employed to estimate tibiofemoral and ACL loading including muscle activations and forces. The predicted muscle forces can be used to predict tibiofemoral contact forces (shear and compressive) and ACL loading. Modeling muscles as an ideal force generator, the force production would be the product of muscle activation and maximum isometric force of muscle. When physiological parameters are introduced into the model, the muscle force estimation becomes more complex, where muscle length, contractile velocity, and muscle pennation angle must also be taken into consideration.

To estimate three-dimensional ACL loading the previously estimated muscle excitations can be calculated to estimate muscle forces (Shelburne, Torry, & Pandy, 2005, 2006). The muscle forces of the three primary movers of the knee (the quadriceps, hamstrings and gastrocnemius) and the estimated knee joint reaction forces can be used to predict realistic ACL loads. McLean, Su, and van den Bogert (2003) estimated the forces transferred through the ACL by the anterior posterior directional forces produced from the quadriceps during side-step cutting. However, they did not include any other directional force within their calculation, which may underestimate ACL loading. Kernozek and Ragan (2008) proposed a solution to estimate ACL loading from sagittal plane moments, where the net moment of the knee is comprised to the net moments generated by the moments acting on the lower leg by the moments of the ankle, hamstrings, quadriceps via patellar tendon and soleus. Using:

$$M_{knee} = M_{pat} + M_{ham} + M_{soleus} - M_{ankle}$$

$$= M_{pat} + M_{ham} - M_{gast}.$$

Then, along with the sagittal plane loading, frontal and transverse plane loading can be estimated from the net joint moments about the knee (Markolf et al., 1995; Markolf, Gorek, Kabo, & Shapiro, 1990). Using these equations, ACL loading from all three planes can be estimated:

Adduction Moments (M_{AD}):

$$F_{ACL-FP} = -7.5003\ e^{(-0.041xM_{AD})}x\ \Theta_{knee}$$

Abduction Moments (M_{AD}):

$$F_{ACL-FP} = 3.8054\ e^{(-0.001xM_{AB})}x\ \Theta_{knee}$$

Internal Rotation Moments (M_{IR}):

$$F_{ACL-TP} = -24.57\ e^{(-0.045xM_{IR})}x\ \Theta_{knee}$$

External Rotation Moments (M_{ER}):

$$F_{ACL-TP} = 8.6485\ e^{(-0.032xM_{ER})}x\ \Theta_{knee}$$

F_{ACL}:

$$F_{ACL} = F_{ACL-SP} + F_{ACL-FP} + F_{ACL-TP}$$

There are numerous aspects of ACL injury risk mechanisms during landings that have been establish. As described above, anatomical influences, previous exercise or current exercise, landing mechanics, ground reaction forces, muscular strength imbalances, knee joint laxity, strength deficiencies, ligament size and properties and altered pre-activation EMG of lower extremity muscle influence the incidence of ACL injuries. Although there are numerous aspects of ACL injury mechanisms that are understood, there are aspects of the ACL injury paradigm that are still not understood in men. A more complete understanding of ACL injury mechanisms in men is needed. Although MRI has been establish as the gold standard for assessing the ACL, diagnostic ultrasound has emerged as a tool that can reliably see and measure the ACL compared to MRI. Although ultrasound is reliable to MRI, further research is needed to establish reliability of researchers when MRI comparisons are not included in the study.

CHAPTER 3: METHODS

PARTICIPANTS

For Aim #1, a previous assessment of sample size requirements for investigations of reliability (Walter, Eliasziw, and Donner (1998)) was consulted. According to Walter et al. (1998), a minimum of 20 participants are needed to achieve a minimal ICC of 0.6 and a desired ICC of 0.8 with an α of 0.05 and β of 0.20 when the study design includes two raters and two measurements per limb per participant. For aim #2, an a priori power analysis based on significant correlations between ACL size and height (r=0.70; (Fayad, Rosenthal, Morrison, & Carrino, 2008)) indicated a minimum of 13 participants would be needed to find similar strength of relationship with a power of 0.80. For aim #3, a priori power analysis using a significant correlation between knee extensor and ACL strain (Russell, Palmieri, Zinder, & Ingersoll, 2006) indicated a minimum of 15 participants would be needed.

Prior to participant recruitment, research approval from the Institutional Review Board of Old Dominion University was obtained. Upon arrival, each participant was informed of the study procedures, signed consent forms, and filled out an activity history questionnaire (see Appendix) to determine his level of fitness/activity based on ACSM guidelines: recreationally active (150 minutes of moderate intensity exercise, 5 days by 30 minutes) or collegiately active (participate in collegiate athletics).

EXPERIMENTAL PROTOCOL

Aim #1

To address Aim #1, ten female and ten male participants, ages 18-35, reported to the Neuromechanics Lab (ODU, Norfolk, VA, USA). After completion of consent forms, health history and activity level questionnaires, two ultrasound measurements of the ACL of each limb were performed. Researchers were be stationed at two different medical examination tables. Each researcher independently placed the participant in the desired position, targeting the knee to be flexed to at least 90°. The researcher then took a measurement of the ACL via ultrasound (Fujifilm Sonosite, Bothell, WA), at the tibial insertion site (Chen et al., 2013; Mahajan et al., 2015; Suzuki et al., 1991). The superior end of the transducer was placed at the distal portion of the patella and aligned with the patella tendon. The inferior portion of the transducer was rotated 30° to the medial aspect of the tibia so that the inferior portion is on the medial aspect of the patella tendon and the superior portion is slightly lateral to the patella tendon (Chen et al., 2013; Mahajan et al., 2015; Suzuki et al., 1991). Once an appropriate image of the ligament was taken, the participant moved to the next researcher, where the process was repeated. Once all researches took appropriate images of the ligament, the process was repeated for a second time. Once the first round of images were collected for the two researchers, all participants were re-examined by researcher one. The second round of image collection was implemented to examine intra-rater reliability. Once all images were collected, each researcher measured their own images via ImageJ (NIH). ImageJ was globally calibrated by dimensions set on each image. Once the ACL had been identified in the image, the perpendicular distance from the edge of the superior and inferior border of the ligament was used to measure the ligament diameter. Once all images were measured, two way random absolute agreement ICCs were conducted for inter-rater and intra-rater reliability using SPSS (v21.0, SPSS Inc., Chicago, IL).

Aim #2 and Aim #3

For Aims #2 and #3, 17 male participants, ages 18-35 were recruited for study. Participants reported to the Neuromechanics Lab (ODU, Norfolk, VA, USA) for all testing. Once all forms were completed, participants were asked to dress in spandex or tight fitting shorts and laboratory shoes. Participants' height and weight were measured with a medical weight scale with a height beam (Detecto, Webb City, MO, USA). Participants' anteromedial bundle diameter were measured at the tibial insertion using a diagnostic ultrasound (Fujifilm Sonosite, Bothell, WA) (Chen et al., 2013; Mahajan et al., 2015; Suzuki et al., 1991) guided by previously defined procedures. To measure ACL thickness the participant will have their knee flexed more than 90°. The superior end of the transducer was placed at the distal portion of the patella and aligned with the patella tendon. The inferior portion of the transducer was rotated 30° to the medial aspect of the tibia so that the inferior portion is on the medial aspect of the patella tendon and the superior portion is slightly lateral to the patella tendon (Chen et al., 2013; Mahajan et al., 2015; Suzuki et al., 1991). Once ACL thickness was measured, participants' knee joint laxity was measured using a KT 2000 (Medmetric, San Diego, CA) by a certified athletic trainer according in accordance with Snyder-Mackler, Fitzgerald, Bartolozzi, and Ciccotti (1997). Then, participant's maximal jumping ability was be assessed. Before maximal jumps are recorded, participants were allowed a warm-up per their discretion. Participants were asked to perform three maximal jumps, where the optimal performance will be recorded for maximal jumping ability.

Once ACL diameter, knee joint laxity, and jump height were determined, participants height and weight, the electromyography setup was performed. Before electrodes were placed on the muscle bellies, participant body hair were removed for the electrode placement, along with abrasions to remove epithelial cells. Once removed, the area was cleaned with alcohol. All electrodes were placed on the gluteus medius, rectus femoris, vastus medialis, vastus lateralis, medial hamstring, lateral hamstring, tibialis anterior and medial gastrocnemius of the participants' right leg, following the guidelines set by Cram, Kasman, and Holtz (1998). The placement for the gluteus medius was placed on the proximal third from the iliac crest to the greater trochanter. The rectus femoris electrode was placed halfway between the knee and the anterior inferior ridge of the iliac spine. The vastus lateralis electrode was placed on the muscle belly at an oblique angle laterally from the midline of the thigh. The vastus medialis electrode was placed on the muscle belly above the knee cap and placed at an oblique angle medially from the midline. The medial and lateral hamstrings electrodes was placed on each muscle belly half way between the gluteal fold and the back of the knee. The tibialis anterior electrode was placed on the muscle belly just lateral to the tibia approximately one third of the distance between the knee and the ankle. The medial gastrocnemius electrode was placed on the muscle belly at an oblique angle medial to the midline so that the electrode runs perpendicular to the fibers. Electrodes were secured with pre-wrap (Mueller Sports Medicine, Prairie du sac, WI) and athletic tape (Collins Sports Medicine, Raynham, MA). Once all electrodes are secure, maximal isometric contractions (MVIC) of the knee extensors, knee flexors, hip abductors, hip adductors, ankle plantarflexors, and ankle dorsiflexors were recorded. During the MVIC testing, isometric strength was assessed for the hip abductors and adductors and for the knee flexors and extensors. 10-second maximal voluntary isometric contractions (MVIC) were recorded for normalization of the electromyography data. MVICs of the knee extensors and flexors were completed with participants upright on a testing table and their knee flexed to 60 degrees (Alvares et al., 2015). Next hip abductor MVICs were collected, where participants were instructed to lay on their sides with the testing leg in a neutral alignment and abducted to 30 degrees (Fredericson et al., 2000). Participants were instructed not to rotate their body and to remain in the neutral alignment to maximize hip abductor MVIC. Lastly, ankle plantarflexor MVICs were collected with the ankle positioned at 0 degrees (Bojsen-Møller et al., 2004).

For motion capture assessments, 18 retro reflective markers were placed bilaterally on anatomical landmarks, including the: acromioclavicular joint, most superior aspect of the iliac crest, greater trochanter, and lateral and medial epicondyles of the femur, lateral and medial malleoli, and first and fifth metatarsals. Two elastic Velcro straps (McDavid, Woodridge, IL, USA) were placed on each leg, one on the middle of the thigh and the second on the middle of the shank. The Velcro straps will hold semi-rigid, molded Orthoplast plates (Johnson & Johnson, Raynham, MA, USA) with four tracking markers on each plate. There was a tracking plate placed on the heel of each shoe, in between the scapulae, on the thighs of each leg, shanks of both legs, and a pelvis marker plate that was placed on an elastic Velcro strap around the waist.

Once all markers, plates, and electrodes are placed on the body, participants will stand on a force plate while a static calibration trial is recorded. Once the calibration trial is collected, the anatomical markers were taken off. Participants will then be asked to complete five single leg landing with their dominate leg randomized from a 40 cm box and 5 landings from a box set equal to their maximal jumping ability. During landings, participants were instructed to land to their natural ability. If subject fell on both legs, and lost control during the landing, the trial was considered unsuccessful.

Instrumentation

The instrumentation that was used to collect data in the study was an eight camera motion analysis system, Bertec force plates, and dynamometer and Delsys surface electrodes. ACL diameter was measured via ultrasound (Fujifilm Sonosite, Bothell, WA) while knee joint laxity was measured using a KT 2000 (Medmetric, San Diego, CA). The MVIC measurements for each muscle/movement were recorded using a portable fixed dynamometer (BTE, Hanover, MD, USA). Three-dimensional marker coordinate data was collected at 200 Hz using an eight-camera motion analysis system (Vicon, Centennial, CO, USA). All, force data was collected at 2000 Hz using two Bertec force plates (Bertec Co., Columbus, OH). All, and EMG data was collected at

2000 Hz. using a Trigno 16 wireless surface electrode electromagnetic system (Delsys Inc., Boston, MA, USA).

Data Analysis

A kinetic model made of eight segments of a trunk, pelvis, right and left thighs, shanks and feet were created from the standing calibration trial (Weinhandl, Joshi, & O'Connor, 2010). Data reduction was performed using Visual 3D (Version 5, C-Motion, Inc.) for processing. The motion capture and GRF data were filtered using a low-pass fourth order Butterworth zero lag filter with a 15 Hz cut off frequency (Kristianslund, Krosshaug, & van den Bogert, 2012).

Three-dimensional ankle, knee, and hip angles were calculated using a joint coordinate system approach (Grood & Suntay, 1983). The hip joint center was defined as 25% of the distance from the ipsilateral to the contralateral greater trochanter markers (Weinhandl & O'Connor, 2010). The knee joint center was defined as the midpoint between the lateral and medial markers on the condyles of the femur (Grood & Suntay, 1983). The ankle joint center was defined as the midpoint between the medial and lateral malleoli.

Three-dimensional joint kinetics were calculated using a Newton-Euler approach (Bresler & Frankel, 1950) with body segment parameters estimated from (Dempster, 1955) and were reported in the distal segment reference frame. The EMG data were pre-amplified and high-pass filtered using fourth-order, zero lag, recursive Butterworth filter with a cutoff frequency of 20Hz to remove movement artifact. The signal was full-wave rectified and normalized to maximize recorded signal of each muscle over the trial. The full-wave rectified signal will then be low pass filtered with a cutoff frequency of 5 Hz to create a linear envelope. All EMG data will then be normalized against MVICs of the individual muscles. Filtered marker coordinate and GRF data were imported into OpenSim (Version 3.3, SimTK, Stanford, CA). Participant specific kinematic models were created from the base of The Rajagopal Model by scaling to each participant's height and mass. The inverse kinematics problem was solved using a least squares approach while accounting for constraint weights (Spoor & Veldpaus, 1980). Inverse dynamics was computed using the inverse kinematic and GRF data. Static Optimization (Steele, DeMers, Schwartz, & Delp, 2012) was used to estimate muscle forces at each time point using the sum of muscle stress squared . Next, the JointReaction analysis algorithm (Steele et al., 2012) was used with the muscle force and inverse dynamics data to estimate joint reaction forces during the landings. Lastly, three-dimensional ACL loading was calculated using previously established equations (Laughlin et al., 2011; Weinhandl et al., 2013), including the simulated sagittal plane forces, sagittal plane knee angles, and frontal and transverse plane moments from inverse dynamics.

Statistical Analysis

For Aim #1, after all images are analyzed in ImageJ, two-way random ICCs were conducted for intra-rater (between images and sessions) and inter-rater reliability for both the full ACL and the AM bundle diameters using SPSS (v21.0, SPSS Inc., Chicago, IL). When analyzing the reliability of measurements, moderate intra- and inter-rater reliability will have ICCs greater than 0.75 and excellent reliability will have ICCs greater than 0.90 (Gellhorn & Carlson, 2013).

For aim #2, Pearson correlations were performed to examine the relationship of the AM bundle diameter with time spent during high, moderate and low intensity exercise periods, knee flexor to knee extensor MVIC strength ratios, and hip abductor and hip adductor MVIC strength ratios. Three separate correlation analyses were conducted: 1) AM bundle diameter normalized to body weight (cm/BW) and strength variables in raw format, 2) diameter and strength variables normalized to body weight (cm/BW and N/BW (Hurd et al., 2011) respectively), and 3) diameter normalized to weight (cm/BW) and strength variables normalized to body size (N/m$^{0.67}$; (Jaric, Mirkov, & Markovic, 2005)). Relationships were considered weak, moderate, and strong relationships when the correlation coefficients (r) were < |0.30|, between |0.31| and |0.69|, and > |0.70|. The significance level was set a prior at $p<0.05$.

During aim #3, correlation analyses were performed using SPSS (Version 26, IBM). Correlations between peak ACL loading (40 cm (BW), relative landing (BW), relative landing (BW/√RLH)) and anteromedial ACL diameter, pre-activation integrated EMG levels of seven lower extremity muscles, and passive knee joint laxity were examined. Relationships were considered weak, moderate, and strong relationships when correlation coefficients (r) were < |0.30|, between |0.31| and |0.69|, and > |0.70|. The significance level was set a prior at $p<0.05$.

CHAPTER 4: INTRA AND INTER RATER RELIABILIY OF ULTRASOUND MEASURES OF THE ANTERIOR CRUCIATE LIGAMENT

ABSTRACT

Currently, magnetic resonance imagining (MRI) is the most common method of examining the anterior cruciate ligament (ACL). Diagnostic ultrasound, a cost effective alternative to MRI, can locate and measure the ACL at the tibial insertion site, specifically measuring the entire ACL diameter, the anteromedial bundle (AM) and the posterolateral bundle. The purpose of this study was to determine intra and inter-rater reliability of ultrasounding ACL diameter and anteromedial bundle diameter in researchers with low to moderate ultrasound experience. We hypothesized that intra and inter-rater reliability of ACL and AM diameters would reach acceptable levels, a minimal intraclass correlation (ICC) of 0.6 and a desired ICC of 0.8 with an α of 0.05 and β of 0.20. Twenty volunteers were recruited for this study. During the ACL ultrasound measures, participants were seated with their knee flexed to 90°. Each rater recorded two images of both the right and left ACL and AM bundles. Next, all participants were re-examined by rater one for intra-rater reliability analyses. Two-way random ICCs were conducted for intra-rater (between sessions) and inter-rater reliability for both the full ACL and the AM bundle diameters using SPSS. Standard errors between sessions for Rater 1's AM bundle and ACL diameters were less than 0.03 cm. Intra-rater reliability was higher in AM bundles compared to full ACL, 0.76 vs. 0.59, respectively. In addition, standard errors between Rater 1 and Rater 2 were less than 0.03 cm. Inter-rater reliability was higher in AM bundles compared to full ACL, 0.71 vs 0.41, respectively. The results of the study indicate researchers with low to moderate training with ultrasound measures can locate and measure the Anterior Cruciate Ligament.

INTRODUCTION

Anterior cruciate ligament injuries (ACL) are devastating to athletes, where the repercussion of those injuries may impact joint health and performance for weeks to years later (Ardern, Taylor, Feller, & Webster, 2012, 2014; Ardern, Webster, Taylor, & Feller, 2011a, 2011b; Caine, DiFiori, & Maffulli, 2006; Griffin et al., 2000; Griffin et al., 2006; Langford, Webster, & Feller, 2009; Maffulli, Longo, Gougoulias, Loppini, & Denaro, 2010). Current reports estimate 100-250,000 ACL injuries per year and suggest that surgical repairs cost between 1.5 billion dollars to 3 billion annually in the in the United States (Boden, Griffin, & Garrett, 2000; Griffin et al., 2006; Hewett, Myer, Ford, Paterno, & Quatman, 2016). A majority of ACL injury risk research focuses on assessing movement patterns and muscular imbalances that may perpetuate poor mechanical movement patterns associated with injury (Hewett, Ford, Hoogenboom, & Myer, 2010; Hewett & Johnson, 2010; Hewett et al., 1999; Hewett, Myer, Ford, et al., 2005; Hewett et al., 2016; Hewett, Zazulak, et al., 2005; Krosshaug et al., 2007; Laughlin et al., 2011; Levine et al., 2013; Myer et al., 2005; G. D. Myer et al., 2015; G. D. Myer, Ford, Barber Foss, et al., 2009). While assessing movement patterns are important, various anthropometric variables may also influence predisposition for ACL injuries. Identified anthropometric variables that may influence ACL injury risk include lower extremity alignment, Q-angle, pelvis width, femoral notch widths, and muscular imbalances (Deie et al., 2002; Eiling et al., 2007; Hewett et al., 2006; G. D. Myer, Ford, Divine, et al., 2009; G. D. Myer et al., 2005; Park et al., 2009). While these variables may be important evaluations of ACL injury risk, current research suggests that variation in ACL diameter (also referred to as width) may impact ACL injury risk (A. F. Anderson et al., 2001; Mahajan et al., 2015). As individuals perform dynamic tasks, a smaller diameter ACL may not be able to resist the amount of force or tibial translation that a larger ligament may be able to handle. If the cross sectional area or the slack length of the ligament is increased, the amount of stress that the ligament can handle prior to failure may be increased (David, Grood, Noyes, & Zernicke, 1978). Diameter of the ACL may be pertinent to the evaluation of ACL injury predisposition, and may be useful in a pre-athletic screening evaluation or in further research into ACL injury risk mechanisms.

Currently, magnetic resonance imagining (MRI) is the most accurate and reliable non-surgical option to examine healthy joints and asses structural damage within joints (Crawford et al., 2007; Friemert et al., 2004; J. Hall et al., 2016). Furthermore, MRI is the most commonly used imagining technique to diagnose or confirm ligamentous injuries (J. Hall et al., 2016; Kocabey, Tetik, Isbell, Atay, & Johnson, 2004; Hwan-Mo Lee et al., 2000). Although MRI is might be reliable and accurate measurement tool, utilizing MRI is time consuming and can be expensive (Pobozy & Kielar, 2016). When diagnosing ACL rupture, MRI has a reliable accuracy at 80-94% (Kostov, Stojmenski, & Kostova, 2014; Tung, Davis, Wiggins, & Fadale, 1993). Inter-rater reliability is reduced when utilizing MRI to measure the ACL and its bundles lengths and widths. Cohen et al. (2009), found that the anteromedial bundle had an average length and width in the sagittal plane of 36.9±2.8 and 5.1±0.7 mm. In the frontal plane the average width was 4.2±0.8 mm. Cohen also reported that the intraclass correlations between the two raters using MRI was 0.42 and 0.51 for the sagittal plane and frontal plane widths of the anteromedial bundle, respectively. A cost effective and mobile (available in the field) alternative to MRI is diagnostic ultrasound.

Diagnostic ultrasound is a reliable measure of tendon and ligament thicknesses within the human body, including: rotator cuff, Achilles, Iliopsoas, Quadriceps, Patellar, Pes Anserine, and Biceps Femoris tendons and Medial Collateral, Lateral Collateral, and Posterior Cruciate ligaments (Cholewinski, Kusz, Wojciechowski, Cielinski, & Zoladz, 2008; Hodgson, O'Connor, & Grainger, 2014; K. Khan et al., 2003; Lalitha, Reddy, Reddy, & Kumari, 2016; Martinoli, 2010; Prickett et al., 2003; Tsai, Chiang, & Lew, 2015; Ying et al., 2003). Recent research has shown that ultrasound is a viable alternative to MRI in locating and analyzing the length and diameter of the ACL at the tibial insertion site (Chen et al., 2013; Mahajan et al., 2015). Diagnostic ultrasound has also been utilize to diagnose ACL injuries (Chen et al., 2013; Friedl & Glaser, 1991; Z. Khan et al., 2006; Larsen & Rasmussen, 2000; Mahajan et al., 2015; Suzuki et al., 1991). However, there are many important methodological considerations for diagnostic ultrasound.

When using diagnostic ultrasound several methodological aspects can impact accuracy and reliability. For instance, to obtain an appropriate measure of the ACL the plane of view must be considered. Previous research has utilized both posterior and off-planar views where the transducer is rotated in multiple planes to obtain the best-quality image of the ACL (Chen et al., 2013; Mahajan et al., 2015). Each one of these techniques provides potential obstacles to diagnoses. As the probe is rotated about multiple axes for one participant, the rotations about the axes may be slightly different for another participant. Furthermore, the off planar measurement may make it difficult to compare ultrasound measurements to MRI. Additionally, the level of expertise an individual has with using diagnostic ultrasound can affect their ability to accurately detect where the ACL and its boundaries lie . Although previous research has utilized MRI to validate ultrasound measures of the ACL, the presence of various structures or potential fibers within the image may make it difficult to identify the ACL for individuals with different levels of expertise. In support of ultrasound imaging, one study has indicated excellent inter rater reliability with an intraclass correlation (ICC) of 0.93 for experienced radiologists (Mahajan et al., 2015). However, many different practitioners within the realm of research use diagnostic ultrasound. Considering the infancy of imaging ACL anthropometrics using ultrasound, reliability should be assessed in practitioners with varying levels of expertise.

Therefore, the purpose of this experiment was to determine intra- and inter-rater reliability of ultrasound measures of ACL diameter in researchers with low (Rater 1) and moderate (Rater 2) ultrasound experience. It was hypothesized that the intra- (Rater 1) and intra-rater (Rater 1 & 2) reliability of ACL and AM bundle diameters would reach acceptable levels, a minimal ICC of 0.6 and a desired ICC of 0.8 with an α of 0.05 and β of 0.20.

METHODS

Participants

According to Walter et al. (1998), 40 ACL images are required to achieve a minimal ICC of 0.6 and a desired ICC of 0.8 with an α of 0.05 and β of 0.20 to compare inter-rater reliability. Seven female and seven male participants, ages 18-35, from the surrounding community were recruited for the study. Twenty-eight knees (14 participants) were included in assessments by Rater 1. Ten (5 participants) of the twenty-eight knees were also measured by Rater 2. After arrival, participants were informed of the study procedures and consent for participation was obtained. Participants were excluded from the experiment if they had any previous lower extremity orthopedic surgery or had experienced any lower extremity injury within the previous six months. Participants were not allowed any amount of physical activity before the ultrasound measures were taken the day they came to be examined.

Ultrasound Raters

Both researchers had different levels of experience with ultrasound measures. Researcher one (Rater 1) has minimal experience with ultrasound, but has been practicing ultrasound techniques for the previous two years. The second researcher (Rater 2) has utilized and published work with ultrasound measures on soft tissue (Bennett, Rider, Domire, DeVita, & Kulas, 2014). To confirm each researcher was able to locate and measure the ACL, numerous practice sessions were performed using the portable diagnostic ultrasound unit (Fujifilm Sonosite, Bothell, WA). During each practice session, each researcher was responsible for positioning each participant, locating the ACL using the ultrasound probe, and obtaining an image of the ACL. All images obtained during practice sessions were visually compared to previous reports (Chen et al., 2013; Mahajan et al., 2015).

Ultrasound Protocol

Each rater was stationed at a different medical examination table throughout the lab space, where observation of the other rater's ultrasound measurements were undetectable. Each participant attended an ultrasound measurement session with each rater where two images of both the participant’s right and left ACLs were obtained. Each rater independently placed the participant in the desired position of knee flexion to at least 90°. Each rater then obtained a measurement of ACL diameter near the tibial insertion site (Chen et al., 2013; Mahajan et al., 2015; Suzuki et al., 1991) using a portable diagnostic ultrasound unit (Fujifilm Sonosite, Bothell, WA). The superior end of the transducer was placed at the distal portion of the patella and aligned with the patella tendon. The inferior portion of the transducer was rotated 30° to the medial aspect of the tibia so that the inferior portion was on the medial aspect of the patella tendon and the superior portion was slightly lateral to the patella tendon (Chen et al., 2013; Mahajan et al., 2015; Suzuki et al., 1991). An image order of right, left, right, left was followed to increase time between scans of the same limb and reduce familiarization with exact placement of the transducer. Once all images of the ligaments were obtained, each participant returned for a subsequent session with Rater 1 to be used in an intra-rater reliability measures assessment.

Ultrasound Image Reduction

Once all images were collected, each researcher independently measured ACL diameter on their own images via an open-source imaging software, ImageJ (Abràmoff, Magalhães, & Ram, 2004). First, all images were calibrated from pixels to centimeters in ImageJ. Next, the anterior and posterior borders of the ACL and AM bundle were identified as the endpoints of the illuminated lines on the image (Figure 1a & 1b). Diameter of the ACL/AM bundle was determined on each image by measuring a perpendicular line relative to the direction of the ACL fibers and extending from the anterior to posterior border of the ACL (Chen et al., 2013).

Statistical Analyses

Once all images were measured, two-way random ICCs were conducted for intra-rater (between images and sessions) and inter-rater reliability for both the full ACL and the AM bundle diameters using SPSS (v21.0, SPSS Inc., Chicago, IL). When analyzing the reliability of measurements, "good" intra- and inter-rater reliability will have ICCs greater than 0.75 and excellent reliability will have ICCs greater than 0.90 (Gellhorn & Carlson, 2013). Bland Altman plots were constructed to present the amount of agreement between sessions (Rater 1) and between raters for both full ACL and AM bundle diameters (Giavarina, 2015).

RESULTS

Mean and one standard deviation of each rater's and each session's (Rater 1) ACL and AM bundle diameters are presented in Table 1. Mean differences between session 1 and 2 of

Rater 1's AM bundle and ACL diameters were less than 0.03 cm (0.68 vs 0.66 cm and 0.86 vs 0.84 cm, respectively). In addition, mean differences between Rater 1 and Rater 2 were less than 0.04 cm (0.66 vs. 0.63 cm and 0.86 vs. 0.84 cm, respectively).

Intra-rater and inter-rater reliability measures are presented in Table 2. Overall, intra-rater and inter-rater reliability were improved for AM bundle diameter measurements compared to full ACL diameter measurements. For AM bundles, ICC between sessions were 0.76 and between raters were 0.71. For the full ACL, ICCs only reached 0.59 and 0.41 between sessions and raters, respectively. Standard error of measurements were very small for all comparisons (<0.04 cm; Table 2).

Bland Altman plots of intra and inter-rater measurements are provided in Figure 2 (full ACL) and Figure 3 (AM bundle). Bias between Session 1 and Session 2 were +0.02 cm for both full ACL and AM bundle diameters (Figures 2a and 3a). Reported measurements of full ACL diameter were generally larger in Session 1 compared to Session 2 when average measures were 0.82 cm or higher (Figure 2a). No visual bias tendencies between sessions appear present for AM bundles (Figure 3a). Bias between raters (Rater 1 - Rater 2) were +0.03 and +0.01 cm for full ACL and AM bundle diameters, respectively (Figures 2b and 3b). For full ACL diameter, Rater 1's positive measurement bias appears to occur when average ACL diameters were 0.85 cm and greater (Figure 2b). However, Rater 1's positive measurement bias for the AM bundle appears to occur when measures were below 0.65 cm (Figure 3b).

DISCUSSION

Assessing the ability, accuracy, and reliability of different raters is crucial for an in-vivo technique such as diagnostic ultrasound, especially when using ultrasound in newer assessments such as ACL diameter. Therefore, the purpose of this research was to examine the intra- and inter-rater reliability in two low to moderately experienced ultrasound users on measurements of ACL diameter. We hypothesized that intra- and inter-rate reliabilities would reach acceptable levels, with a minimal ICC of 0.6 and a desired ICC of 0.8. Overall, intra-rater reliability reached levels of "good" agreement (ICC >0.75), while inter-rater reliability nearly reached "good" agreement (ICC = 0.71).

There are numerous anatomical, muscular and neurological factors that may alter an athletes susceptibility to injury (Besier et al., 2001; Griffin et al., 2006; Hewett, Myer, & Ford, 2005; Mendiguchia, Ford, Quatman, Alentorn-Geli, & Hewett, 2011). Diagnostic ultrasound can be used to examine intracapsulsar ligaments and has the potential for pre-screening athletes for overall ACL health and/or injury prevention measures. People that have sustained an ACL injury have smaller contralateral ACL size compared to healthy individuals (Mahajan et al., 2015), which can be detected using ultrasound. One of the major drawbacks to any screening measurement is the reliability within and between different raters. When utilizing ultrasound, small alterations to probe angle, shift, and rotation can lead to significant changes in the desired measurements (Gellhorn & Carlson, 2013). Although there may be the potential for multiple sources of error that could lead to intra- or inter-rater unreliability in the acquisition and measurement of the ACL through ultrasound, this study found that the reliability of Rater 1 (rater with minimal experience) was a very good between separate images (ICC=0.86) and good between separate sessions (0.76). In addition, inter-rater reliability across two separate training levels (Rater 2 was moderately trained) was moderate to good (ICC=0.71). Thus, the results of this study suggest that using ultrasound to measure ACL diameter in a typical research setting (laboratories without access to radiologists) is a reliable alternative to MRI for assessing ACL diameter.

Previous work using ultrasound to measure diameter of the full ACL found average ACL diameter of healthy persons to be 0.83 cm (Chen et al., 2013) and 0.81 cm (Mahajan et al., 2015). The current study's measurements agree with the previous literature, with average ACL diameter measured from both raters as from 0.84 to 0.86 cm. In addition to full ACL diameter, both raters' AM bundle diameters in this study agree with previous AM bundle diameters measured through intraoperative arthroscopy (Cohen et al., 2009). The current study's AM bundle diameters ranged from 0.63 to 0.68 cm, which are similar to the previous intraoperative measurements of 0.68 cm (Cohen et al., 2009). Interestingly, this study also found increased reliability with AM bundles compared to full ACL diameter when using ultrasound. Thus, future work incorporating ACL diameters should consider analyzing the AM bundle in addition to full ACL diameter.

Conclusion

This study shows that two researchers with low to moderate training with ultrasound measures can locate and measure ACL width, with a moderate inter rater ICC 0.70 and a moderate intra rater reliability ICC of 0.85. This study has comparable reliability when compared to Mahajan et al. (2015), which utilized two well experienced radiologists which had a strong interclass ICC of 0.93 when measuring ACL diameter at the tibial insertion site. This is a forward step in the potential to utilize ultrasound as a prescreening mechanism for lower extremity injuries. With training, it may be possible that medical professionals performing prescreening tests, may be able to locate and reliably measure the diameter of the ACL utilizing ultrasound. This potential utilization could save time and money compared to utilizing MRI in indicated the diameter of the ACL in patients.

TABLES AND FIGURES

Table 1. Anterior cruciate ligament diameter (cm) measures for each session and rater: mean±std.

	N	AM Diameter (Mean±Std)	Full ACL Diameter (Mean±Std)
Rater 1 (Session 1)	56	0.68±0.08	0.86±0.06
Rater 1 (Session 2)	56	0.66±0.07	0.84±0.05
Rater 1 (Subset)	20	0.66±0.09	0.86±0.08
Rater 2	20	0.63±0.07	0.84±0.05

Note: AM: Anteromedial bundle of ACL; Full ACL: ACL diameter including both AM and posterolateral bundles.

Table 2. Intra-rater and inter-rater reliability measures for AM bundle and full ACL diameters.

		Absolute Agreement (Single, Average)	Consistency (Single, Average)	F, P	SEM (cm)
AM Bundle	Intra-rater (Sessions)	0.76, 0.87	0.79, 0.89	8.7, <0.001	0.01
	Intra-rater (Measures)	0.86, 0.92	0.86, 0.92	13.1, <0.001	0.02
	Inter-rater	0.71, 0.83	0.70, 0.83	5.7, <0.001	0.03
Full ACL	Intra-rater (Sessions)	0.59, 0.74	0.63, 0.77	4.4, <0.001	0.01
	Intra-rater (Measures)	0.81, 0.89	0.81, 0.90	9.7, <0.001	0.01
	Inter-rater	0.41, 0.58	0.44, 0.61	2.5, 0.024	0.02

Note: AM: Anteromedial bundle of ACL; Full ACL: ACL diameter including both AM and posterolateral bundles; F, P: f-statistic and p-values for intraclass correlations; All intraclass correlations were two-way random; SEM: standard error of measurement.

Figure 1. Ultrasound image of full ACL (1a) and AM bundles (1b) for one subject.

Figure 1a.

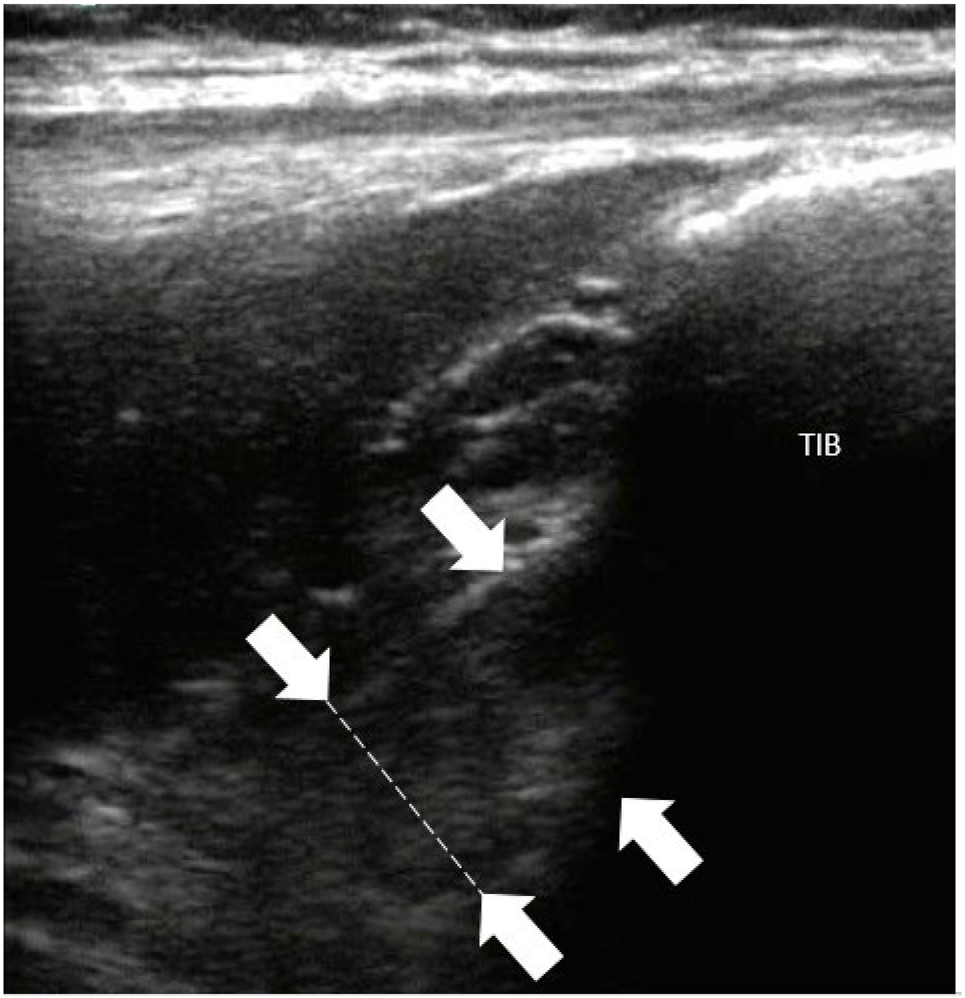

Figure 1b.

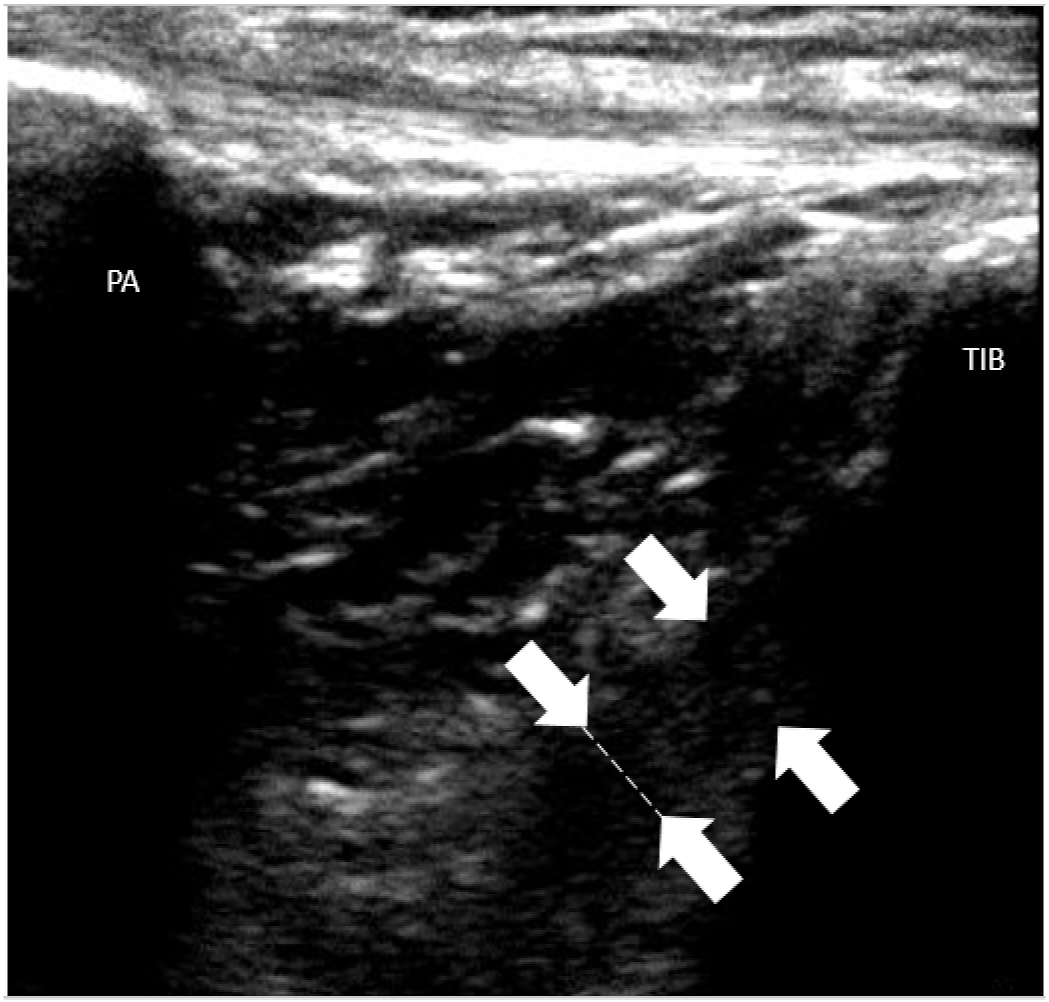

Figure 1 Caption. The full ACL (Figure 1a) and AM Bundle (Figure 1b) are outlined by white arrows on the figures. The dashed line between the arrows are the measurement that were taken for the diameter of the ACL and AM bundle. TIB is the tibia and PA is the Patella.

Figure 2. Bland-Altman plots of ACL diameters comparing sessions (2a) and raters (2b).

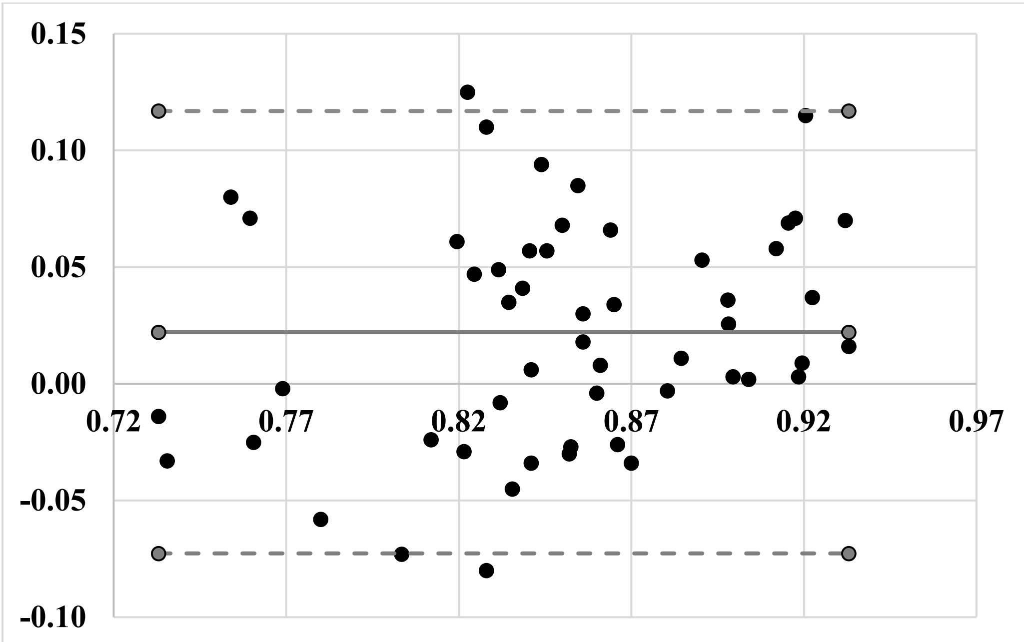

Figure 2a.

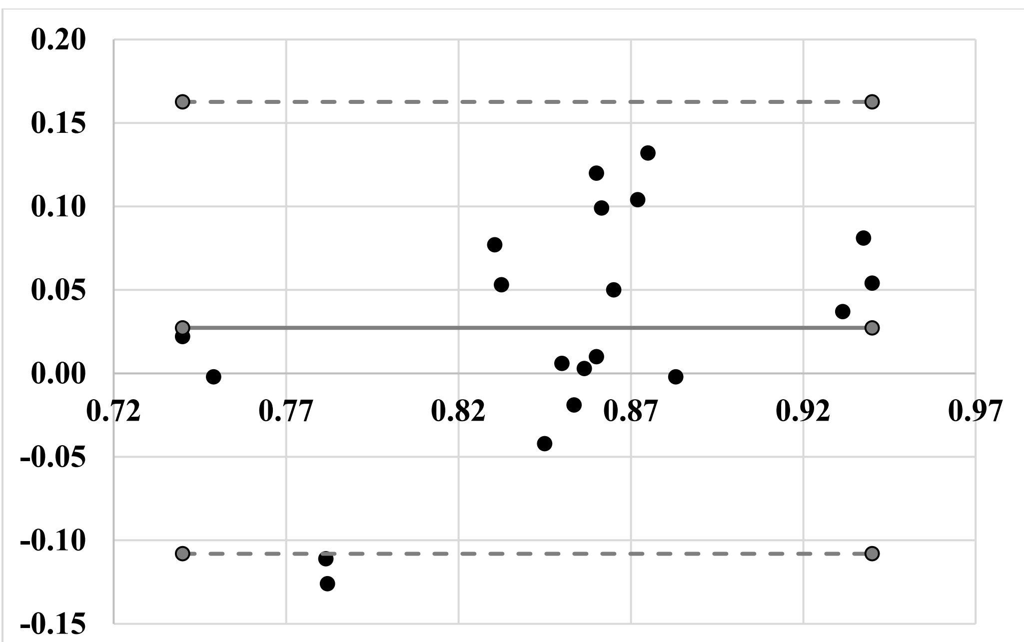

Figure 2b.

Figure 2 Caption: Intra-rater comparisons of full ACL diameter are provided in 2a. Inter-rater comparisons are provided in 2b. The X and Y-axes present the mean diameter of the two sessions and the differences in diameter measurements between the two sessions for Rater 1, respectfully. Dashed lines represent the 95% confidence intervals of agreement. Solid lines are the mean difference. Bias is the difference between the mean difference and the x-axis at zero.

Figure 3. Bland-Altman plots of AM bundle diameters comparing sessions (3a) and raters (3b).

Figure 3a.

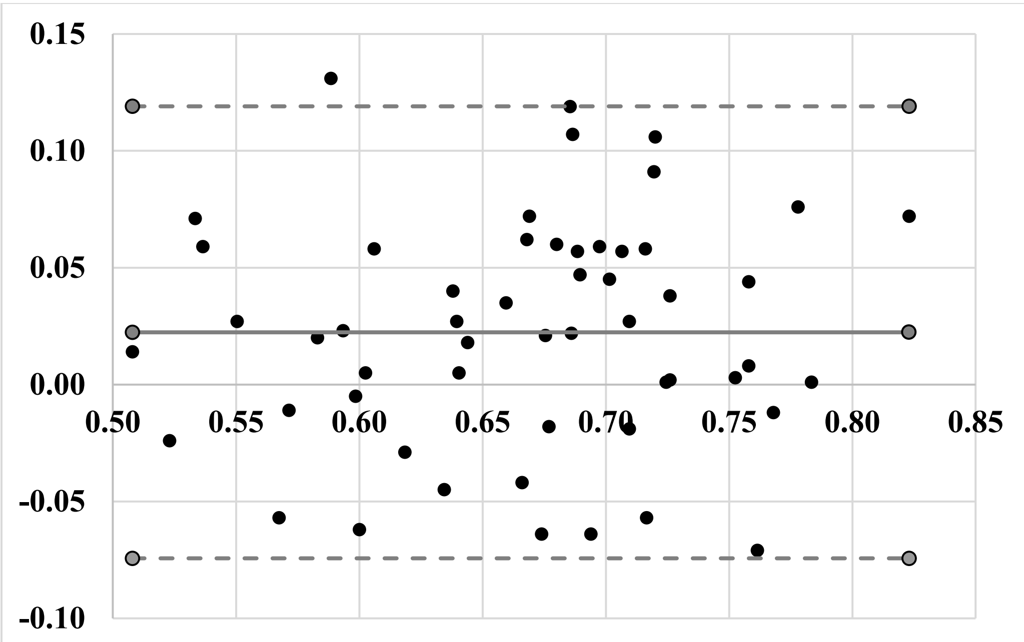

Figure 3b.

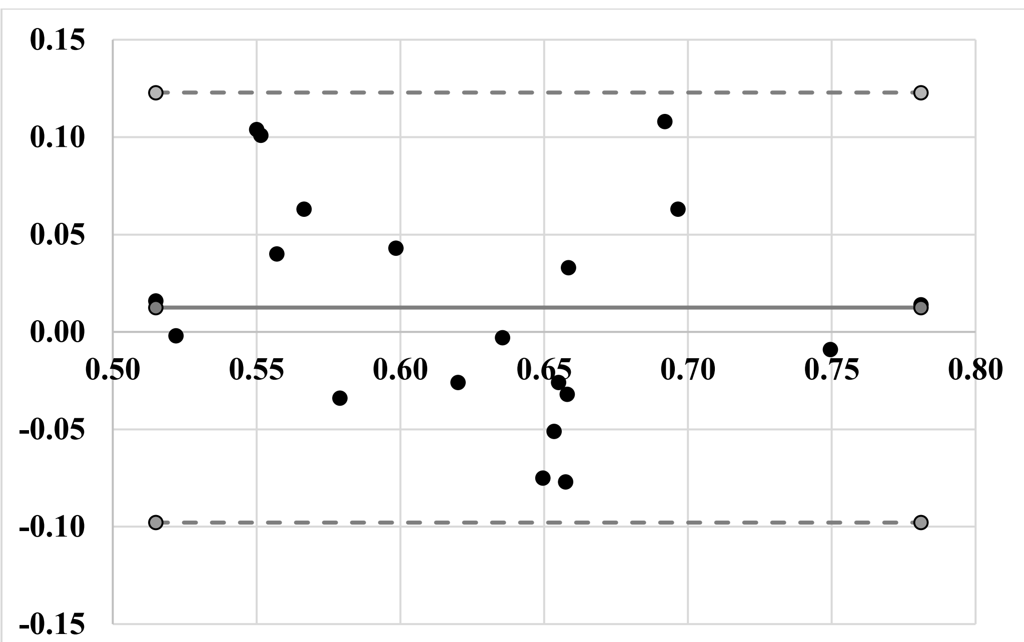

Figure 3 Caption: Intra-rater comparisons of AM bundle diameter are provided in 3a. Inter-rater comparisons are provided in 3b. The X and Y-axes present the mean diameter of the two sessions and the differences in diameter measurements between the two sessions for Rater 1, respectfully. Dashed lines represent the 95% confidence intervals of agreement. Solid lines are the mean difference. Bias is the difference between the mean difference and the x-axis at zero.

CHAPTER 5: AN EXAMINATION OF THE RELATIONSHIP BETWEEN ACL DIAMETER AND MUSCLE STRENGTH, ANTHROPOMETRICS, AND PHYSICAL ACTIVITY LEVELS

ABSTRACT

Anterior Cruciate Ligament (ACL) injuries are a common injury to the knee. The size of the ACL has been shown to influence how much load the ACL can handle. Various anthropometric variables, strength measurements, and physical activity or previous training has been shown to influence ACL loading. In accordance to Roux's Law of functional adaptation, increased healthy loading over time may increase the size of ligament structures. Therefore the purpose of this study was to examine the relationships of the knee flexors, extensors, hip abductors and hip adductors strengths, hamstring to quadriceps strength ratio, hip abductor to hip adductor strength ratio, anthropometric measurements of height and body mass, and previous physical activity levels to the diameter of the AM bundle of the ACL. It was hypothesized that there will be a strong relationship between strength measurements, strength ratios, height, weight and previous physical activity levels with the diameter of the AM bundle of the ACL. During data collection, ACL measurements were taken via ultrasound, while participants were seated with their examined knee was flexed to 90°. Next, muscular strength measurements were taken while performing a maximal voluntary isometric contraction measured by a portable fixed dynamometer. Once all data was collected, correlations were performed on each of the independent variables to ACL diameter. It was found that the main significant correlations to ACL diameter were the knee extensor and hip adductor strength values. There were no significant correlations between activity level and AM ACL diameter. As strength of the knee extensors and hip adductors increase, ACL size increases. As a person has greater strength in both muscle groups, the increased strength may cause continuous loading from these muscle groups which may lead to increases in ACL size.

INTRODUCTION

Physical activity is a crucial part of maintaining positive health outcomes throughout all stages of life (Hu et al., 2005; Hyland, Sodergren, & Singh, 1999; Laforge et al., 1999; Pate et al., 1995; Riise, Moen, & Nortvedt, 2003; Warburton, Nicol, & Bredin, 2006). In youth populations, 10-19 year olds (Cuff, Loud, & O'riordan, 2010), physical activity may include running, resistance training, plyometric exercises or any sport specific drills that may be utilized to increase performance or reduce injury risk (Hewett, Myer, Ford, et al., 2005; Hewett et al., 2009; G. D. Myer, Ford, Brent, & Hewett, 2006; G. D. Myer, Ford, McLean, et al., 2006; Snyder et al., 2009). With youth athletics, significant injuries may increase an individual's risk of degenerative diseases and could eventually lead to poor movement mechanics later in life (Schroeder et al., 2015). Anterior Cruciate Ligament (ACL) injuries comprise 45% of all injuries sustained to the internal structures of the knee (Majewski et al., 2006). Injuries and repairs of the ACL have lasting effects for years after competition has ceased (Daniel et al., 1994; L. Lohmander, Östenberg, Englund, & Roos, 2004; L Stefan Lohmander, Englund, Dahl, & Roos, 2007; Roos, 2005).

The incidence of ACL injuries is extremely high in athletic populations ranging from 15-25 years old (Dick et al., 2007; Griffin et al., 2006). ACL injuries may increase due to several mechanical factors, such as increased knee abduction angles, increased tibial translation, increased frontal plane loading, increased anteroposterior shear knee joint force, absolute muscular strength, muscular strength imbalances muscular fatigue, joint laxity, Q-angle, pelvis width, femoral notch widths and ACL size (Alentorn-Geli et al., 2009; Cochrane et al., 2007; Deie et al., 2002; Eiling et al., 2007; Griffin et al., 2006; Hewett et al., 2006; Hewett, Myer, Ford, et al., 2005; Myer et al., 2005; G. D. Myer, Ford, Divine, et al., 2009; G. D. Myer et al., 2005; Park et al., 2009). When examining muscular strength, many studies have indicated that muscular imbalances between the hamstring and quadriceps increase ACL injury risk (Ahmad et al., 2006; P. Kannus, 1988a; P. Kannus & Jarvinen, 1990; G. D. Myer, Ford, Barber Foss, et al., 2009). This decrease in strength reduces the ability of the hamstrings to resist anterior tibial translations during dynamics tasks. In addition, weakness of the hip abductors is associated with frontal plane motion of the knee, e.g. increased femoral adduction, which increases ACL injury risk (Claiborne, Armstrong, Gandhi, & Pincivero, 2006; Powers, 2010). Previous research indicates athletes with reduced ACL injury risk have hamstring to quadriceps strength ratios of 0.50 to 0.80 (P. Kannus, 1988b; P Kannus & Järvinen, 1990). Other researchers have predicted a more exact 0.60 hamstring to quadriceps strength ratio to prevent lower extremity injury (Dunnam, Hunter, Williams, & Dremsa, 1988b).

ACL size has also been associated with increased ACL injury risk (Mahajan et al., 2015). For example, a larger body weight with a smaller ACL size may increase risk of ACL failure (Mahajan et al., 2015). In addition, ACL size significantly affects its ability to withstand loading, with smaller ACL's having reduced capacity to dissipate/withstand forces (Mahajan et al., 2015). Increases in ACL size are associated with increased body weight (Chaudhari et al., 2009; Fayad et al., 2008). There are inconsistencies on the relationship between ACL size and height of participants. Participants with increased height may have larger ACL size (Naveen Chandrashekar, Mansouri, Slauterbeck, & Hashemi, 2006); however, other research has found there is no relationship between height and ACL size (Chaudhari et al., 2009; Mahajan et al., 2015).

ACL size is commonly analyzed through MRI due to its accuracy and reliability (Kostov et al., 2014; Rayan, Bhonsle, & Shukla, 2009); however, recent studies have demonstrated ultrasound reliability and feasibility of examining and measuring the ACL (Chen et al., 2013; Mahajan et al., 2015; Paczesny & Kruczyński, 2011). The ACL is comprised of two primary bundles, the anteromedial (AM) and posterolateral (PL)(Amis & Dawkins, 1991). Recent research using diagnostic ultrasound has also been able to accurately identify the AM and PL bundles of the ACL (Hsiao, Chang, & Özçakar, 2016). Recently, we found the AM bundle is a more reliable measure than full ACL diameter amongst researchers with various levels of training. For healthy adults, the size of the AM bundle is approximately 0.68 cm (Hsiao et al., 2016), whereas the full ACL diameter is approximately 0.80 cm (Chen et al., 2013; Mahajan et al., 2015).

Ligaments and tendons have the ability to undergo remodeling via changes in the production of collagen, proteoglycans and glycosaminoglycan in response to stress (Buckwalter & Grodzinsky, 1999; Hayashi, 1996). Physical activity can alter tendon properties over time (Kongsgaard et al., 2005; Kubo et al., 2002; Kubo et al., 2001; Lenskjold et al., 2015; Magnusson & Kjaer, 2003; Markovic & Mikulic, 2010; Rosager et al., 2002). The ability to show or confirm hypothesized alterations to ligament is more complicated. A lack of physical activity in animals results in decreased size of the ACL (Hayashi, 1996; Herpin et al., 1990). Inversely, an increase in stress applied to the ACL can show signs of hypertrophy in animals (Cabaud et al., 1980; Dahners et al., 1989; Wood et al., 1998). It has been suggested that a continual strain applied to the ACL in humans would cause a similar hypertrophic response to that seen in animals (Simon et al., 2010). The most feasible way of applying stress to muscles, ligaments, and tendons in the human body is through exercise. Grzelak, Podgorski, Stefanczyk, Krochmalski, and Domzalski (2012), found that high performance weightlifters had increased ACL and PCL size compared to healthy controls. Although the direct response of ACL size to physical activity/loading was not assessed, the previous study indicates that a link may exist between physical activity and ACL size. Furthermore, performing exercises that focuses on dynamic movements such as plyometrics, muscular strength and neuromuscular control may apply appropriate stress to the lower body and reduce injury risk (Enoka, 2008; Hamill et al., 2015; Kawakami et al., 1995; McGinnis, 2013). These training aspects may influence adaptations to musculature around the hip and knee, and potentially the ACL. However, the relationships of muscular strength, physical activity, and anthropometric measures to ACL size is unknown.

Therefore, the purpose of this study was to examine the relationships of the knee flexors, extensors, hip abductors and hip adductors strengths, hamstring to quadriceps strength ratio, hip abductor to hip adductor strength ratio, anthropometric measurements of height and body weight, and previous physical activity levels to the diameter of the AM bundle of the ACL measured via ultrasound. It was hypothesized that there will be a strong relationship ($r > 0.7$) for quadriceps strength, adductor strength, hamstring to quadriceps strength ratios, hip abductor to adductor strength ratios, height, weight and previous physical activity levels with the diameter of the AM bundle of the ACL. We also predicted that higher activity levels would be related to larger strength ratios.

METHODS

Seventeen male participants (age: 24.8±3.3 yrs., height: 1.78±0.07 m, weight: 806.49±107.04 N) ages 18-35 were recruited and participated in this study, which was approved by the university institution review board. After arrival, each participant was informed of the study's protocol, signed consent forms, and filled out health history and activity level questionnaires. A priori power analysis was conducted that showed 13 participants were needed to find significant correlations between ACL size and height with a power of 0.80 with an r of 0.70 (Fayad et al., 2008). Participant inclusionary criteria included: age 18-35 years, individuals who participate in exercise and had no prior lower extremity injuries within the last six months and no ACL injuries. All participants were required to have participated in at least the minimal requirements for physical activity stated by American College of Sports Medicine (ACSM) guidelines: 150 minutes of moderate intensity exercise weekly. Participants completed questionnaires to determine their level of physical activity per week for the previous six months. The answers to the questionnaires were self-reported about their exercise activity, intensity level and the amount of time completing exercise at either a low, moderate or high intensity exercise (Table 2.).

Next, the AM bundle diameter of each participant's right knee was measured at the tibial insertion using a diagnostic ultrasound (Fujifilm Sonosite, Bothell, WA) (Chen et al., 2013; Mahajan et al., 2015; Suzuki et al., 1991). To measure ACL diameter, participants' right knee was flexed to a position of 90° (Chen et al., 2013) or greater (Suzuki et al., 1991). The superior end of the transducer was placed at the distal portion of the patella and aligned with the patellar tendon. The inferior portion of the transducer was rotated 30° to the medial aspect of the tibia so that the inferior portion was on the medial aspect of the patella tendon and the superior portion was slightly lateral to the patella tendon (Chen et al., 2013; Mahajan et al., 2015; Suzuki et al., 1991). Once ACL images were captured using ultrasound, images were then analyzed in ImageJ (Abràmoff et al., 2004). First, all images were calibrated from pixels to centimeters in ImageJ. Next, the anterior and posterior borders of the AM bundles were identified as the endpoints of the illuminated lines on the image (Figure 4). The diameter of the AM bundle was measured on each image using a perpendicular line relative to the direction of the bundle fibers and extending from the anterior to posterior border of the ACL.

Maximal voluntary isometric contraction (MVIC) testing was completed on the right lower extremity (quadriceps, hamstrings, hip abductor, and hip adductors) of each participant using a portable fixed dynamometer (Portable Evaluator, BTE, Baltimore, MD) (Kollock Jr, Onate, & Van Lunen, 2010). All joint positions were measured using a goniometer. All muscle groups were tested three times. Participants were asked to maximally contract for 10 seconds and were given two minutes of rest between each trial. The peak force was recorded for each trial and then averaged and normalized. Specific protocols for each muscle tested are explained below.

For quadriceps and hamstring testing, participants were seated in an upright position and secured to the testing chair (Alvares et al., 2015). Participants were strapped to the testing chair to ensure little movement from the rest of the body as participants maximally contracted the desired muscle group. The portable fixed dynamometer was secured to the lower right leg directly above the participant's medial malleolus. The knee was flexed to 60 degrees (Alvares et al., 2015). During quadriceps testing, participants were instructed to extend the lower leg against the dynamometer as forcibly as possible. During hamstring testing, participants were instructed to flex the lower leg maximally against the dynamometer. During hip abductor and adductor testing, participants were instructed to lay on the left side, with legs, hips and knees in neutral positions and the right hip abducted to 30 degrees (Fredericson et al., 2000). The dynamometer was wrapped around the participant's lower right leg directly above the medial malleolus. Participants were instructed to remain on their side and not to rotate their upper body while keeping their hip in a neutral sagittal plane alignment as they maximally abducted or adducted against the dynamometer.

Statistical Analysis

Pearson correlations were performed to examine the relationship of the AM bundle diameter, which was normalized to body weight due to influence body weight has on ligament size (A. F. Anderson et al., 2001) with time spent during high, moderate and low intensity exercise periods, knee flexor to knee extensor MVIC strength ratios, and hip abductor and hip adductor MVIC strength ratios (all in N/BW). Because both body weight and "body size" can influence muscle forces (Jaric et al., 2005), we performed three separate correlation analyses: 1) AM bundle diameter normalized to body weight (cm/BW) and strength variables in raw format, 2) diameter and strength variables normalized to body weight cm/BW and N/BW (Hurd et al., 2011), respectively, and 3) diameter normalized to weight (cm/BW) and strength variables normalized to body size ($N/kg^{0.67}$; (Jaric et al., 2005)). Relationships were considered weak, moderate, and strong relationships when the correlation coefficients (r) were $< |0.30|$, between $|0.31|$ and $|0.69|$, and $> |0.70|$. The significance level was set a prior at $p<0.05$.

RESULTS

The mean and one standard deviation of AM Bundle diameters was 0.64±0.08 cm. Correlations of exercise times and strength variables with AM bundle diameter are presented in Tables 3 and 4. Statistically significant correlations are stated below based on the normalization factors that were utilized for the AM bundle and strength measurements.

AM Bundle Diameter per BW (cm/BWs)

A moderate negative correlation of height and normalized AM bundle diameter was found ($r=-0.494$, $p=0.044$). A moderate negative correlation was also found for hip abductor to adductor strength ratio and AM bundle diameter ($r=-0.521$, $p=0.032$, Table 4).

AM Bundle Diameter per BW (cm/BWs) and Strength per BW (BWs)

When the AM bundle diameter and dynamometer force were normalized to BW there were strong correlations between diameter and extensor (r=0.653, p=0.005; Table 4) and adductor (r=0.659, p=0.005; Table 4) force. In addition, a moderate correlation was found with flexor force and AM bundle diameter (r=0.527, p=0.030; Table 4).

AM Bundle Diameter per BW (cm/BWs) and Strength per Body Size (kg$^{0.67}$)

When dynamometer muscle force was normalized to body size, N/kg$^{0.67}$ (Jaric et al., 2005), there were significant moderate correlations between AM bundle diameter and extensor (r=0.577, p=0.015) and adductor dynamometer muscle forces (r= 0.585, p=0.014; Table 4).

DISCUSSION

The purpose of this study was to investigate the relationship between AM bundle diameter via ultrasound to knee extensors and flexors forces, hip abductor and adductor forces, hamstrings/quadriceps strength ratios, hip abductor/adductor strength ratios, height, weight and previous physical activity. We hypothesized AM bundle diameter would have a strong positive relationship with knee extensor and adductor strength, knee and hip strength ratios, height, weight and time spent exercising. Our hypotheses were partially rejected, as only two strong relationships between tested variables and AM bundle diameter were found (knee extensor and hip adductor forces normalized to body weight). Although we found multiple moderate relationships when strength variables were normalized, no significant relationships were found between physical activity and AM bundle diameter.

There are several known factors that can increase ACL injury risk during dynamic tasks: increased knee abduction angles, increased frontal plane loading, absolute muscular strength, muscular strength imbalances and ACL size (Alentorn-Geli et al., 2009; Cochrane et al., 2007;

Deie et al., 2002; Eiling et al., 2007; Griffin et al., 2006; Hewett et al., 2006; Hewett, Myer, Ford, et al., 2005; Myer et al., 2005; G. D. Myer, Ford, Divine, et al., 2009; G. D. Myer et al., 2005; Park et al., 2009). Muscular strength imbalances play a role in ACL injury mechanism, where reduced strength of the hamstrings cannot effectively counteract the force produced from the quadriceps during dynamics tasks, increasing the chance of ACL rupture (Hewett et al., 2006). This concept could also be true of the hip musculature. If individuals lack the strength needed in the hip abductors, the abductors may not be effective at counteracting any inwardly directed movement of the femur during dynamic tasks (Brent, Myer, Ford, Paterno, & Hewett, 2013; Homan, Norcross, Goerger, Prentice, & Blackburn, 2013; Jacobs, Uhl, Mattacola, Shapiro, & Rayens, 2007). This study found no significant correlation of knee flexor to extensor strength ratio to AM bundle diameter. However, we did find a significant correlation in hip abductor to adductor strength ratio and AM bundle diameter. These results are surprising as previous research indicates the balance between opposing muscle groups surrounding both the knee and frontal plane hip joints influence ACL loading (Arendt & Dick, 1995; Leetun, Ireland, Willson, Ballantyne, & Davis, 2004). It could be assumed that as individuals complete dynamic activities, i.e. sports or training, a larger disparity in muscular imbalances could place more stress on the ACL. Although ligaments may have the potential to hypertrophy, an increase in muscular imbalances may generate increased translations of the lower extremity stressing the ACL and potentially generate growth (Scheffler et al., 2008). , found a significant correlation between ACL volume and peak extensor torque at 60°/sec, where individuals with larger peak extensors torques had a positive relationship with ACL size. Interestingly, this study found positive relationships between extensor and flexor forces with AM bundle diameter. Thus, increased overall strength surrounding the knee joint appears to be correlated with larger ACL size.

We also found that individuals who had increased adductor strength had larger AM bundle diameter. It is commonly reported that decreased hip abductor strength may place more stress on the ACL during dynamic tasks (Brent, Myer, Ford, Paterno, & Hewett, 2013; Homan, Norcross, Goerger, Prentice, & Blackburn, 2013; Jacobs, Uhl, Mattacola, Shapiro, & Rayens, 2007). The lack of abduction strength and control against less desired hip adduction can place the knee in positions that may increase ACL loading. Whereas an increases in hip adductor strength may lend to increased hip adduction during dynamic tasks. This inwardly directed femur may lead to increased ACL stress during dynamic tasks, which may lead to growth of the AM bundle. Future work should consider investigating the relation between ACL loading, landing patterns and ACL diameter.

Since increases in collagen synthesis/adaptations increase as exercise intensity and mechanical loading increase (Kjær, 2004; Matyas, Anton, Shrive, & Frank, 1995; Wren, Beaupre, & Carter, 2000), it was assumed the more time spent performing intense exercise or the more time spent exercising in general would increase the AM bundle diameter. High performance weightlifters have been shown to have increased ACL and PCL size compared to healthy controls (Grzelak et al., 2012). Grzelak et al. (2012) commented that the weightlifters had begun weightlifting before and/or during the onset of puberty and it is assumed they continued to train through the completion of their study. The current study examined the relationship between time spent exercising at various intensities, including overall exercise time, and the AM bundle diameter of the ACL. We found no significant correlations between ACL diameter and previous physical activity. Thus, standard measures of minutes of physical activity likely are not specific enough detail differences ACL anthropometric measures. Future work should consider comparing ACL anthropometrics amongst several athletic/non-athletic populations to enhance our understanding of the influence of activity level on the ACL.

Conclusion

This study provides insight to the relationship between knee extensor, hip adductor, quadriceps to hamstring strength ratios, hip abductor to adductor strength ratios and physical activity levels with the diameter of the AM bundle of ACL. Overall, quadricep, hamstring, and hip adductor forces are positively related to AM bundle diameter. Physical activity level, measured in minutes, is not related to AM bundle diameter. Thus, surrounding knee and hip musculature appear to be important factors for ACL size differences amongst males.

TABLES AND FIGURES.

Table 3. Average Exercise Times in minutes per week and Correlation r and p values to AM Bundle Diameter Normalized to BW (cm/N).

	High Intensity	Moderate Intensity	Low Intensity	Total Exercise
Average ± 1 Std.	231.74 ± 123.62	70.88 ± 60.94	70.00 ± 70.36	372.65 ± 146.32
Correlation Coefficient Significance Level	r = 0.187 p = 0.473	r = 0.159 p = 0.543	r=-0.492 p=0.45	r = -0.012 p = 0.962

Note: Intensity measured in minutes of activity.

Table 4. Average Dynamometer Strength Measurements and Correlation r and p values to AM Bundle Diameter Normalized to BW (cm/N).

		Quadriceps	Hamstrings	Adductors	Abductors
Force (N)	Average ± 1 Std.	747.85 ± 167.22	364.79 ± 67.72	367.79 ± 87.80	130.19 ± 32.71
	Correlation Coefficient Significance Level	r = 0.317 p = 0.214	r = 0.061 p = 0.817	r = 0.328 p = 0.199	r = -0.277 p = 0.380
Force (BW)	Average ± 1 Std.	0.94 ± 0.28	0.46 ± 0.11	0.47 ± 0.13	0.16 ± 0.04
	Correlation Coefficient Significance Level	r = 0.653 p = 0.005	r = 0.527 p = 0.030	r = 0.659 p = 0.004	r = 0.311 p = 0.225
Force ($N/m^{0.67}$)	Average ± 1 Std.	39.51 ± 10.35	19.21 ± 4.16	19.42 ± 5.21	6.80 ± 1.56
	Correlation Coefficient Significance Level	r = 0.577 p = 0.015	r = 0.419 p = 0.094	r = 0.582 p = 0.014	r = 0.135 p = 0.606

Note: N: Newtons BW: body weight (N); $m^{0.67}$: equation from Jaric et al. 2005, m: height in meters.

Table 5. Average Dynamometer Strength Ratios and Correlation r and p values to AM Bundle Diameter Normalized to BW (cm/N).

		Hams/Quads	Ab/Add
Force (N)	Average ± 1 Std.	0.50 ± 0.10	0.37 ± 0.10
	Correlation Coefficient Significance Level	r = -0.314 p = 0.220	r = -0.521 p = 0.032
Force (BW)	Average ± 1 Std.	0.48 ± 0.41	0.34 ± 0.28
	Correlation Coefficient Significance Level	r = -0.417 p = 0.096	r = -0.304 p = 0.236
Force ($N/m^{0.67}$)	Average ± 1 Std.	0.50 ± 0.10	0.36 ± 0.10
	Correlation Coefficient Significance Level	r = -0.417 p = 0.096	r = -0.304 p = 0.236

Note: N: newtons BW: body weight (N); $m^{0.67}$: equation from Jaric et al. 2005, m: height in meters.

Figure 4. Ultrasound imaging of the AM Bundle of the ACL.

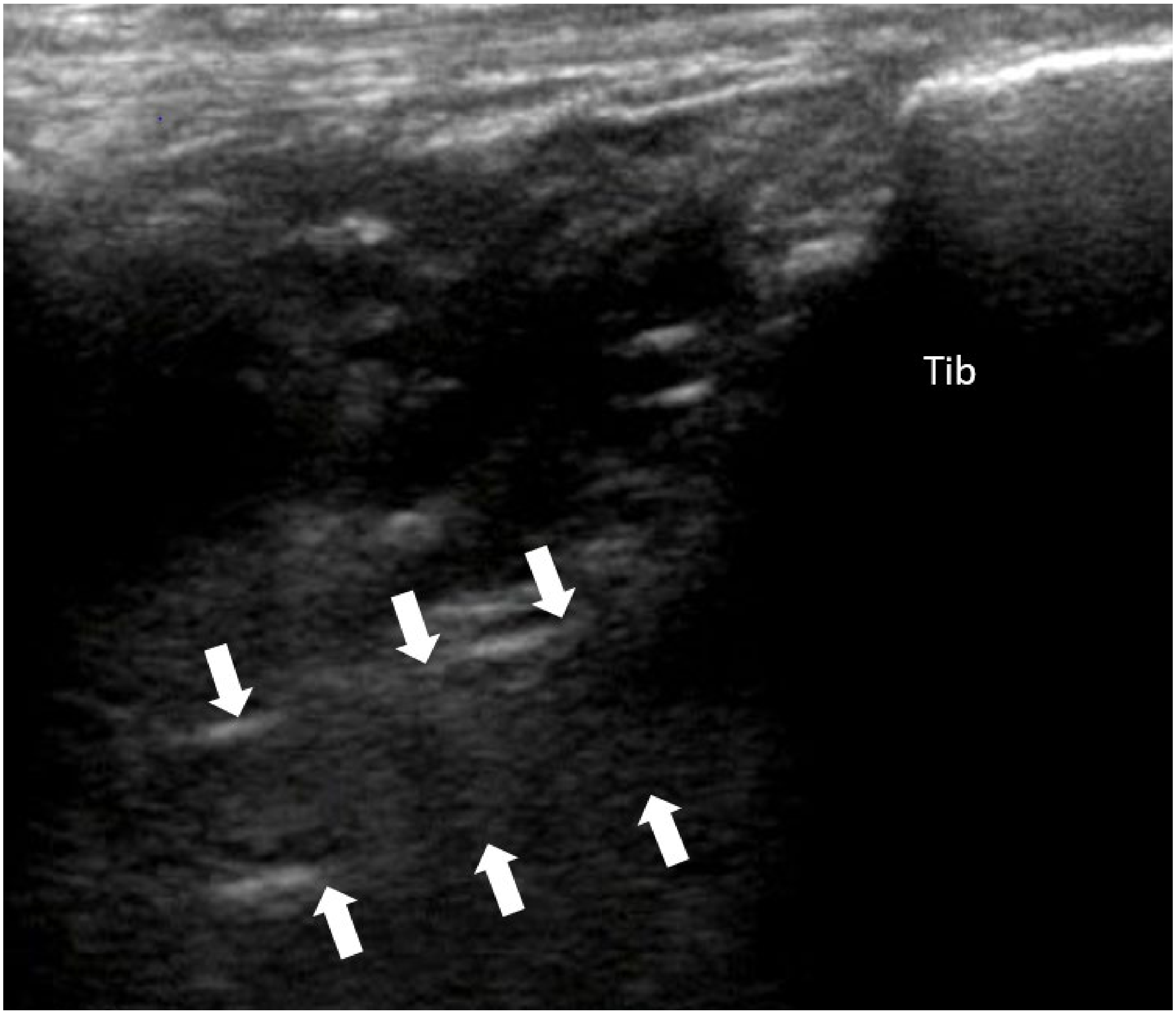

Figure 4 Caption: "Tib" is the tibial location. The white arrows outline the AM bundle.

CHAPTER 6: RELATIONSHIP BETWEEN ACL LOADING AND ACL DIAMETER, KNEE JOINT LAXITY, AND MUSCLE ACTIVATIONS DURING DROP LANDINGS

ABSTRACT

Research has demonstrated that there are numerous mechanisms that cause Anterior Cruciate Ligament injuries. However, research is inconclusive about the injury mechanisms that are directly correlated to ACL loading. Therefore the purpose of this study was to find the relationship between ACL loading and ACL diameter, knee joint laxity, and muscle activations during 40 cm and relative height drop landings. Participant passive knee joint laxity, anteromedial bundle diameter and EMG were collected. Visual 3D and Opensim were utilized to calculate inverse dynamics. Static optimization was used to solve for joint reaction forces. ACL loading was then calculated utilizing methods from previous establish literature. It was found that simulated ACL loading is moderately correlated to lateral hamstring pre-activation during 40 cm ($r = 0.53$) and relative landing heights ($r = 0.58$). It was also found that ACL loading is correlated to the ratio of medial hamstring to lateral hamstring pre-activation during 40 cm ($r = -0.57$) and relative landing heights ($r = -0.576$). The results indicate that lateral hamstring activation is significantly related to ACL loading and that many previously defined factors such as passive knee joint laxity, anteromedial bundle diameter and pre-activation EMG of the quadriceps and gastrocnemius are not correlated to simulated ACL loading.

INTRODUCTION

Throughout the past decade, research has found many anthropometric, kinematic and kinetic variables may increase an individual's risk of anterior cruciate ligament (ACL) injury. Anatomical/anthropometric variables that can influence ACL injury risk include smaller ACL diameter, and increased joint laxity and various (Baratta et al., 1988; Chaudhari et al., 2009; Cowley et al., 2006; Fremerey et al., 2000; Gehring et al., 2009; Griffin et al., 2000; Hewett, Zazulak, et al., 2005; Mahajan et al., 2015; G. D. Myer et al., 2008; Shultz et al., 2004). In addition, research using motion capture and force data have found several kinematic and kinetic factors that may increase risk of ACL injury: increased knee abduction angles, increased frontal plane loading, increased anteroposterior shear knee joint force and increased ground reaction forces (Alentorn-Geli et al., 2009; Boden, Dean, et al., 2000; Boden, Sheehan, Torg, & Hewett, 2010; Cochrane et al., 2007; Griffin et al., 2006; Hewett, Myer, Ford, et al., 2005; Myer et al., 2005). Although numerous factors can influence ACL loading, it is also important to examine anthropometrics of the ACL to determine if any relationships exist with ACL loading.

The orientation of the ACL allows it to resist multi planar motion and forces. Although the largest contributor to ACL loading arises from the sagittal plane, adverse frontal and transverse plane moments also contribute to ACL loading (A. Kiapour & Murray, 2014; A. M. Kiapour et al., 2014). Markolf et al. (1995) found anterior tibial forces combined with internal or external rotation moments and adduction or abduction moments can alter ACL loading. Using in vitro testing, maximal load capacity of the ACL has been determined in both younger cadavers (≤30 years; 2160 N) and in older cadavers (≥60 years; 658 N) (Woo, Hollis, Adams, Lyon, & Takai, 1991). Recently, focus has also shifted to predicting ACL loading during dynamic tasks such as landing. Pflum et al. (2004), estimated ACL loading at be 253 N (0.4 BW) at 40 ms after initial contact during 60 cm height drop landings. Kernozek and Ragan (2008), also found 0.15 BW peak ACL loading during landings from a 0.60 m drop height. Drop landings from 37 cm resulted in an average peak ACL loading of 440 N (0.71 BW) during “soft landings” and 496 N (0.80 BW) during “stiff landings” (Laughlin et al., 2011). During dynamic movements where ACL injury risk may be increased due to increased loading, ACL size can play a role in the ability to resist excessive loading and translation of the tibia.

ACL size can determine the amount of force, strain, and loading that the ACL can handle (Chaudhari et al., 2009; Javad Hashemi et al., 2011). The number of collagen fibers can be increased in larger ACLs, which can allow the ligament to resist increased loading (Chaudhari et al., 2009; Javad Hashemi et al., 2011). Multiple studies have found that individuals who have suffered an ACL injury have smaller contralateral ACL size compared to healthy controls of equal body weight and height (Chaudhari et al., 2009; Mahajan et al., 2015). ACL size may be utilized as a pre-screening measurement to assess ACL injury potential in athletic populations (Mahajan et al., 2015). For instance, if an individual has a larger body weight but a smaller ACL diameter, they may be at increased ACL injury risk. Although a smaller ACL may increase injury risk, laxity in the knee may also negatively influence ACL loading and injury risk.

During landings, individuals with increased knee joint laxity may be more susceptible to ACL failure (Fremerey et al., 2000; Griffin et al., 2000; G. D. Myer et al., 2008; Shultz et al., 2004). Combined with increased peak ground reaction forces and tibial translation, increased joint laxity may allow for greater translation and rotation of the tibia during landings, increasing ACL injury risk. Potential influencers of joint laxity include reduced proprioception, muscular imbalances, decreased muscular strength, poor activation patterns during dynamic tasks, muscular fatigue and poor neurological control (Claiborne et al., 2006; Fremerey et al., 2000; M.

G. Hall et al., 1995; Hurley et al., 1998; Jerosch & Prymka, 1996; McNair & Marshall, 1994; D. Roberts et al., 2007; Rozzi, Lephart, & Fu, 1999; Rozzi, Lephart, Gear, et al., 1999).

The incidence of ACL failure occurs during a time period too fast for conscious response (Krosshaug et al., 2007). ACL failure, peak tibial translation, and peak GRFs occur from 17-50 ms after initial contact (Kernozek & Ragan, 2008; Krosshaug et al., 2007; Torry et al., 2011), whereas maximal muscle forces may require 300-400 ms (Savelberg, 2000; Zatsiorsky & Prilutsky, 2012). Thus, pre-activation of the surrounding musculature is required to assist in dissipating injurious forces to reduce the likelihood of ACL failure (Cowley et al., 2006; Gehring et al., 2009; Hewett, Zazulak, et al., 2005). Increased pre-activation and co-contraction around the knee (e.g. hamstring and quadricep musculature) can increase knee joint stiffness and stability, which can reduce load on passive structures of the knee such as the ACL (Ford, Van den Bogert, Myer, Shapiro, & Hewett, 2008; Palmieri-Smith, Wojtys, & Ashton-Miller, 2008). Increased co-contraction will also reduce unwanted frontal plane movements at the knee (Claiborne et al., 2006; Horita, Komi, Nicol, & Kyröläinen, 2002). Although co-contraction is desired to reduce unwanted movements around the knee, uneven activations can be produced among medial and lateral musculature, leading to increased ACL loading (Sell et al., 2007). In addition to knee musculature, increases in gluteal activation (controlling frontal plane movement of the femur) may increase frontal plane control of the knee (Claiborne et al., 2006), reducing knee valgus positions and the likelihood of knee injury or knee ligamentous damage (Palmieri-Smith et al., 2008; Russell et al., 2006). The activation of lower extremity muscles, joint laxity and the size of the ACL all cohesively contribute to knee movement while resisting unwanted forces and translation. However, no previous research has examined the relationships between ACL loading and the previously defined parameters within the same cohort.

Therefore, the purpose of this study was to examine the relationship between ACL size, knee joint laxity, and pre-activation levels of the lower extremity muscles to estimated ACL loading. We hypothesized that during drop landings, 1) there would be inverse relationships between ACL loading and ACL diameter and muscle pre-activation and 2) a direct relationship between ACL loading and knee joint laxity during drop landings.

METHODS

Participants

The Old Dominion University Institutional Review Board approved this study. Seventeen male participants (average height: 1.78 ± 0.07 m, average weight: 806.49 ± 107.04 N), aged 18-35 years old were recruited using flyers/advertisements and by word of mouth. An a priori power analysis was conducted, which determined that 15 participants were needed to achieve a power of 0.80 for a correlation of quadriceps pre-activation and ACL strain (Russell et al., 2006). Inclusion criteria for participants, determined by questionnaire, were: no lower extremity injuries within the last six months that impeded exercise, no lower body injuries that required orthopedic surgery, and must have met ACSM's guidelines for moderate intensity exercise (150 minutes a week). Upon arrival, all participants were informed of the study protocol and signed consent forms.

Protocol

First, ACL thickness was measured at the tibial insertion using a diagnostic ultrasound (Fujifilm Sonosite, Bothell, WA) (Chen et al., 2013; Mahajan et al., 2015; Suzuki et al., 1991). To measure ACL thickness, participants' right knee were flexed to a position of 90° or greater. The superior end of the transducer was placed at the distal portion of the patella and aligned with the patellar tendon. The inferior portion of the transducer was rotated 30° to the medial aspect of the tibia so that the inferior portion was on the medial aspect of the patella tendon and the superior portion was slightly lateral to the patella tendon (Chen et al., 2013; Mahajan et al., 2015; Suzuki et al., 1991). Next, the anteromedial bundle of the ACL was identified and was measured on each image using a perpendicular line relative to the direction of the anteromedial bundle fibers (Hsiao et al., 2016).

Next, passive knee joint laxity was measured using a KT-2000 arthrometer (MEDmetric Corp, San Diego, CA, USA). Participants were supine with their knee flexed to 30°. Testing and procedures were performed to the manufacturers guidelines (Rohman & Macalena, 2016; Snyder-Mackler, Fitzgerald, Bartolozzi III, & Ciccotti, 1997). Three tests were performed for the participant's dominant leg (determined as which leg was preferred to kick a ball with). First, the arthrometer was zeroed; then, an experienced certified athletic trainer performed an anterior pull/force. The average of the three tests were used for analyses. After knee joint laxity was measured, participants' bilateral maximal jumping ability was recorded. Participants were asked to perform three jumps, reaching upwards towards a target overhead. Jump height was calculated using impulse-momentum relationship. The optimal performance was recorded for maximal jumping height.

Once ACL diameter, knee joint laxity, and jump height were determined, a Delsys Trigno wireless electromyography (EMG) system (Delsys, Inc.) and eight electrodes were used to measure muscle activations. Electrodes placement sites were prepped by shaving hair, abrading hairless skin, and cleaning the skin with alcohol. All electrodes were placed on the muscle bellies of the right leg rectus femoris (RF), vastus medialis (VM), vastus lateralis (VL), medial hamstring (MH), lateral hamstring (LH), medial gastrocnemius (MG) and gluteus medius (GM). All electrodes were placed by the guidelines set by Cram et al. (1998). The rectus femoris electrode was placed midway between the patella and the anterior portion of the inferior ridge of the iliac spine. The vastus lateralis electrode was placed on the muscle belly at an oblique angle away from the midline of the anterior thigh. The vastus medialis electrode was placed on the muscle belly above the patella and at an oblique angle medially from the center of the thigh. The medial and lateral hamstrings electrodes were placed on each muscle belly half way between the gluteal fold and the back of the knee. The medial gastrocnemius electrode was placed on the muscle belly at an oblique angle medial to the midline so that the electrode is perpendicular to the fibers. Lastly, the gluteus medius electrode was placed in the middle of the superior third of the distance between the iliac crest and the greater trochanter.

After electrode placement, three 10-second maximal voluntary isometric contractions (MVIC) were recorded for normalization of the electromyography data. MVICs of the knee extensors and flexors were completed with participants upright on a testing table and their knee flexed to 60 degrees (Alvares et al., 2015). Next hip abductor MVICs were collected, where participants were instructed to lay on their sides with the testing leg in a neutral alignment and abducted to 30 degrees (Fredericson et al., 2000). Participants were instructed not to rotate their body and to remain in the neutral alignment to maximize hip abductor MVIC. Lastly, ankle plantarflexor MVICs were collected with the ankle positioned at 0 degrees (Bojsen-Møller et al., 2004).

Eighteen reflective markers were placed bilaterally on anatomical landmarks, including the acromioclavicular joint, the most superior aspect of the iliac crest, the greater trochanter, the lateral and medial epicondyles of the femur, the lateral and medial malleoli, and the first and fifth metatarsals. Four tracking markers attached to rigid shells were secured to the trunk, pelvis, thighs, and shanks using Velcro straps. Additional tracking plates were placed on the posterior heel of each shoe. After EMG and motion capture preparation, participants stood on a force plate while a static calibration trial was recorded. Next, participants were asked to complete five single leg landing trials each with their dominant leg for two testing conditions: from a 40 cm box and from a box set equal to their previously measured bilateral maximal jumping ability.

Data Analyses

Motion capture and GRF data were imported into Visual3D Biomechanical Software Suite (Version 5, C-Motion, Inc.) for processing. The motion capture and GRF data were filtered using a low-pass fourth order Butterworth zero lag filter with a 15 Hz cut off frequency (Kristianslund et al., 2012). An eight-segment kinematic model was created based on the standing calibration trial. Three-dimensional knee joint kinetics were calculated using a Newton-Euler approach (Bresler & Frankel, 1950) with body segment parameters estimated from Dempster (1955) and reported in the distal segment reference frame. Frontal and transverse plane waveforms for each trial and participant were exported for analyses.

EMG data were pre-amplified and high-pass filtered using fourth-order, zero lag, recursive Butterworth filter with a cutoff frequency of 20 Hz to remove movement artifact (De Luca, Gilmore, Kuznetsov, & Roy, 2010). The signal was full-wave rectified and then low pass filtered with a cutoff frequency of 5 Hz to create a linear envelope. EMG signals during landing were normalized against the peak value of each MVICs of the individual muscles. Integrated EMG was measured from 100 ms before initial contact to initial contact. The integrated signal was calculated for the VM, VL, RF, MH, LH, MG, GM. Then the ratios of MH/LH and VM/VL were analyzed. The ratios of medial muscle to lateral muscle were taken as a 1:1 ratio (da Fonseca, Vaz, de Aquino, & Brício, 2006; Hamstra-Wright et al., 2006).

Next, the filtered marker coordinate and GRF data were imported into an open source modeling software, OpenSim (Version 3.3, SimTK, Stanford, CA). Subject specific kinematic models were created from the base Rajagopal2015 model by scaling to each participant's height and mass. The inverse kinematics problem was solved using a least squares approach while accounting for constraint weights (Spoor & Veldpaus, 1980). Inverse dynamics were computed using the inverse kinematic and GRF data. Static Optimization (Steele et al., 2012) was used to estimate muscle forces at each time point using the sum of muscle stress squared . Next, the JointReaction analysis algorithm (Steele et al., 2012) was used with the muscle force and inverse dynamics data to estimate joint reaction forces during the landings. Lastly, three-dimensional ACL loading was calculated using previously established equations (Laughlin et al., 2011; Weinhandl et al., 2013), including the simulated sagittal plane forces, sagittal plane knee angles, and frontal and transverse plane moments from inverse dynamics. Peak ACL loading for the 40 cm and relative height landings were normalized to body weight (BW). In addition, ACL loading during relative height landings were normalized to the product of body weight and the square root of participants' relative landing height (RLH) (Hass et al., 2005).

Statistical Analyses

Correlation analyses were performed using SPSS (Version 26, IBM). Correlations between peak ACL loading (40 cm (BW), relative landing (BW), relative landing (BW/√RLH)) and anteromedial ACL diameter, pre-activation integrated EMG levels of seven lower extremity muscles, and passive knee joint laxity were examined. In total, three Pearson Product Correlations were conducted. Relationships were considered weak, moderate, and strong relationships when correlation coefficients (r) were $< |0.30|$, between $|0.31|$ and $|0.69|$, and $> |0.70|$. The significance level was set a prior at $p<0.05$.

RESULTS

During the 40 cm single leg drop landings, participants experienced an average peak ACL loading of 1758.89 ± 369.50 N and normalized peak ACL loading of 2.18 ± 0.37 BW. Relative height drop landings were from an average height of 43.37 ± 5.92 cm. During the relative height landing, participants experienced a peak ACL loading of 1604.94 ± 337.82 N and normalized peak ACL loadings of 2.00 ± 0.41 BW and 3.05 ± 0.63 BW/√RLH. The average anteromedial ACL diameter was 0.63 ± 0.08 cm. Participants had an average right knee joint laxity of 2.89 ± 1.47 mm. All integrated EMG signals for both 40 cm and relative landings are reported in Table 6. Simulated activations from the Rajgopal2015 model and measured EMG of the lower extremity muscles are in Figure 5 for the relative landings and Figure 6 for the 40 cm landings. The simulated activations of the lower extremity muscles show similar patterns compared to the measured EMG of the same lower extremity muscles during the relative and 40 cm landings.

No significant relationships were found between ACL loading and ACL diameter or knee joint laxity (all $p>0.05$; Table 7). There were significant moderate associations between lateral hamstring activation and ACL loading: 40 cm (BW; $r = 0.53$, $p = 0.04$), relative landing per unit body weight (BW; $r = 0.58$, $p = 0.02$), and relative landing per unit body weight and landing height (BW/√RLH; $r = 0.59$, $p = 0.02$; Table 6). No other significant relationships were found between ACL loading and individual muscle activations. However, significant moderate relationships were found between the ratio of medial hamstring to lateral hamstring integrated pre-activation EMG signal and ACL loading: 40 cm ($r = -0.570$, $p = 0.021$), relative landing per unit body weight (BW; $r = -0.576$, $p = 0.019$), and relative landings per unit body weight and landing height (BW/√RLH; $r = -0.560$, $p = 0.024$; Table 6).

DISCUSSION

The purpose of this study was to examine the relationship between ACL size, knee joint laxity, and pre-activation levels of the lower extremity muscles to estimated ACL loading. We hypothesized that during drop landings, 1) there would be inverse relationships between ACL loading and ACL diameter and muscle pre-activation and 2) a direct relationship between ACL loading and knee joint laxity during drop landings. We found that normalized ACL loading for the 40 cm (BW) and relative height landings (BW & BW/√RLH) had no significant relationship with anteromedial ACL diameter or passive knee joint laxity. Normalized peak ACL loadings had moderate significant relationships with lateral hamstring pre-activation and the ratio of medial to lateral hamstring activation. No other significant relationships between ACL loading and muscle activations were found.

A larger ACL can withstand increased mechanical loading, strain, shear forces, and tibial translations (Lipps, Oh, Ashton-Miller, & Wojtys, 2012). If ACL size is smaller across participants with equal mass, the smaller ACL may experience higher shear loads and have a greater risk of failure (Javad Hashemi et al., 2011). However, this study found no significant correlation between ACL size and ACL loading during drop landings from two heights. Similarly, Javad Hashemi et al. (2011) found that ACL volume and area had no significant influence on predictions of load at failure of the ACL. Although previous assessments of ACL size have found significantly smaller ligaments in those with a previous ACL injury (Chaudhari et al., 2009; Mahajan et al., 2015), ACL size alone does not appear to be a significant factor of ACL loading.

This study also aimed to determine the relationship between joint laxity and ACL loading. We found no correlation between passive knee joint laxity and peak ACL loading during drop landings from either relative or absolute drop landing heights. Similar to the current work, previous research by Sernert et al. (1999) found poor correlations between dynamic stability tests, such as the hop test, and passive knee joint laxity. Thus, athletes with increased joint laxity may not necessarily have increased ACL loading during dynamics task. It is possible that the influence of passive joint laxity may be negated as muscles contract, providing stability across the joint during dynamic task. Rozzi, Lephart, and Fu (1999), note that as knee joint laxity is increased, proprioceptive feedback was decreased during an open chain knee extension test. Although this proprioceptive feedback delay may be present, the time in which ACL failure happens during landings may be too fast for proprioceptive feedback to influence actions during high impact landing tasks (Burke, Dickson, & Skuse, 1991). Given the results of the current study and those of other published work (Hung-Maan Lee, Cheng, & Liau, 2009; Snyder-Mackler, Fitzgerald, Bartolozzi III, et al., 1997), passive joint laxity does not have a direct relationship with ACL loading during dynamic tasks.

From a neuromuscular perspective, pre-activation of lower extremity musculature surrounding the knee joint should be related to ACL loading during drop landings. However, we found that only the lateral hamstring had a significant association with normalized peak ACL loading. During knee flexion angles ≥60 degrees, hamstring forces can counteract quadriceps force and reduce ACL loading (Li et al., 1999). However, an increase in lateral hamstring activation can increase the joint space of the medial compartment, increasing shear forces of the knee and subsequently increasing ACL loading (Hewett, Zazulak, et al., 2005; Myer et al., 2005; Sell et al., 2007). Combined the previous reports, injury prevention and rehabilitation protocols targeting the ACL should monitor lateral hamstring activation during at-risk tasks.

When activation of the medial and lateral musculature of the thigh is disproportionate, ACL loading can increase (Sell et al., 2007). The current study found that the ratio of medial to lateral hamstring pre-activation had a moderate positive relationship with normalized ACL loading. Previous research has found women are at a greater risk of ACL injury when they have disproportionate medial to lateral hamstring activation compared to men (Rozzi, Lephart, Gear, et al., 1999). From the findings in this study, it appears that men are also more susceptible to increased ACL loading if the medial and lateral hamstrings are activated unevenly. Previous research has also found disproportionate VM to VL activations can increase ACL loading (Sell et al., 2007). However, the current study found no significant correlation to ratios of the VM to the VL integrated pre-activations and ACL loading.

Our hypothesis that GM pre-activation would have an inverse relationship with ACL loading was not supported. We found no correlation between normalized ACL loading and GM pre-activation levels. Palmieri-Smith et al. (2008) found that pre activation of the GM had no influence on participant's valgus knee joint angles during landings from single leg hops. Russell et al. (2006) also found that gluteus activation at initial contact did not influence frontal plane knee angles during drop landings at initial contact. The authors (Russell et al. (2006)) noted that during the landings, as knee flexion reached its maximum, GM activation increased, suggesting that the GM may play more of a hip and femur stabilizer. Since the knee is typically positioned at reduced knee flexion angles at initial contact or during maximal ACL loading, it appears as though the gluteus medius has little influence over knee positions, knee loading, or ACL loading at those positions.

Limitations

There are limitations to consider with this work. First, physical activity levels (time and intensities) were recorded using questionnaires, which are subject to participants' ability to recall this information accurately for the previous six months. Future work may consider monitoring activity levels using accelerometers and tracking participant exercise times and intensities. Second, the Rajagopal2015 model utilizes a single degree of freedom knee joint, where tibial translations and rotations are a function of knee flexion angles. Although we analyzed ACL loads using tri-planar loading (sagittal plane loading from the simulations coupled with frontal and transverse loads from inverse dynamics) from previously validated prediction equations (Laughlin et al., 2011; Weinhandl et al., 2013), future work may consider implementing more sophisticated musculoskeletal models to determine tri-planar knee loads. Lastly, we chose to use ultrasound to measure ACL diameter. Although this is a reliable and accurate measurement tool for ACL anthropometrics (Chen et al., 2013; Mahajan et al., 2015; Suzuki et al., 1991), future work may consider magnetic resonance imaging or other diagnostic systems.

Conclusion

Current research is inconclusive on the exact mechanisms of ACL injury or excessive loading in males. This study found that ACL injury risk mechanisms such as increased joint laxity, ACL diameter, pre-activation of the rectus femoris, vastus lateralis, vastus medialis, medial hamstring and the medial gastrocnemius, and current physical activity level are not strongly related to ACL loading. The only strong relationships found were between pre-activation level of the LH and activations levels of medial to lateral hamstring ratios and normalized ACL loading. Although mechanisms spanning pre-existing anthropometrics to landing mechanics have been previously shown to influence injury risk, many of these factors appear to not directly influence ACL loading during drop landings.

TABLES AND FIGURES

Table 6. Pearson Product Correlation and significance levels between ACL loading and muscle activations.

PACL	RF	VM	VL	MH	LH	MG	GM	MH/LH
40 cm (BW)	r=0.065 p=0.810	r=0.019 p=0.944	r=-0.103 p=0.703	r=0.181 p=0.503	r=0.526 p=0.036	r=-0.337 p=0.202	r=-0.33 p=0.212	r=-0.506 p=0.036
RH (BW)	r=0.179 p=0.507	r=0.112 p=0.678	r=0.003 p=0.99	r=0.213 p=0.429	r=0.582 p=0.018	r=-0.07 p=0.72	r=0.26 p=0.31	r=-0.576 p=0.021
RH (BW/√RLH)	r=0.033 p=0.905	r=-0.051 p=0.853	r=-0.046 p=0.865	r=0.241 p=0.368	r=0.587 p=0.017	r=-0.161 p=0.551	r=-0.042 p=0.878	r=-0.560 p=0.024

Note: Rectus Femoris (RF); Vastus Medialis (VM); Vastus Lateralis (VL); Medial Hamstring (MH); Lateral Hamstring (LH); Medial Gastrocnemius (MG); Gluteus Medius (GM); BW: bodyweight; RH: relative height; RLH: relative landing height.

Table 7. Pearson product correlation and significance levels between ACL loading and joint laxity and AM bundle diameter.

PACL	Joint Laxity	AM Diameter
40 cm (BW)	r=-0.252 p=0.347	r=0.001 p=0.969
RH (BW)	r=-0.247 p=0.357	r=0.034 p=0.897
RH (BW/√RLH)	r=-0.196 p=0.468	r=0.043 p=0.870

Note: BW: Body Weight; RH: relative landing height.

Figure 5. Ensemble muscle activations measured using electromyography compared to simulations during relative height drop landings.

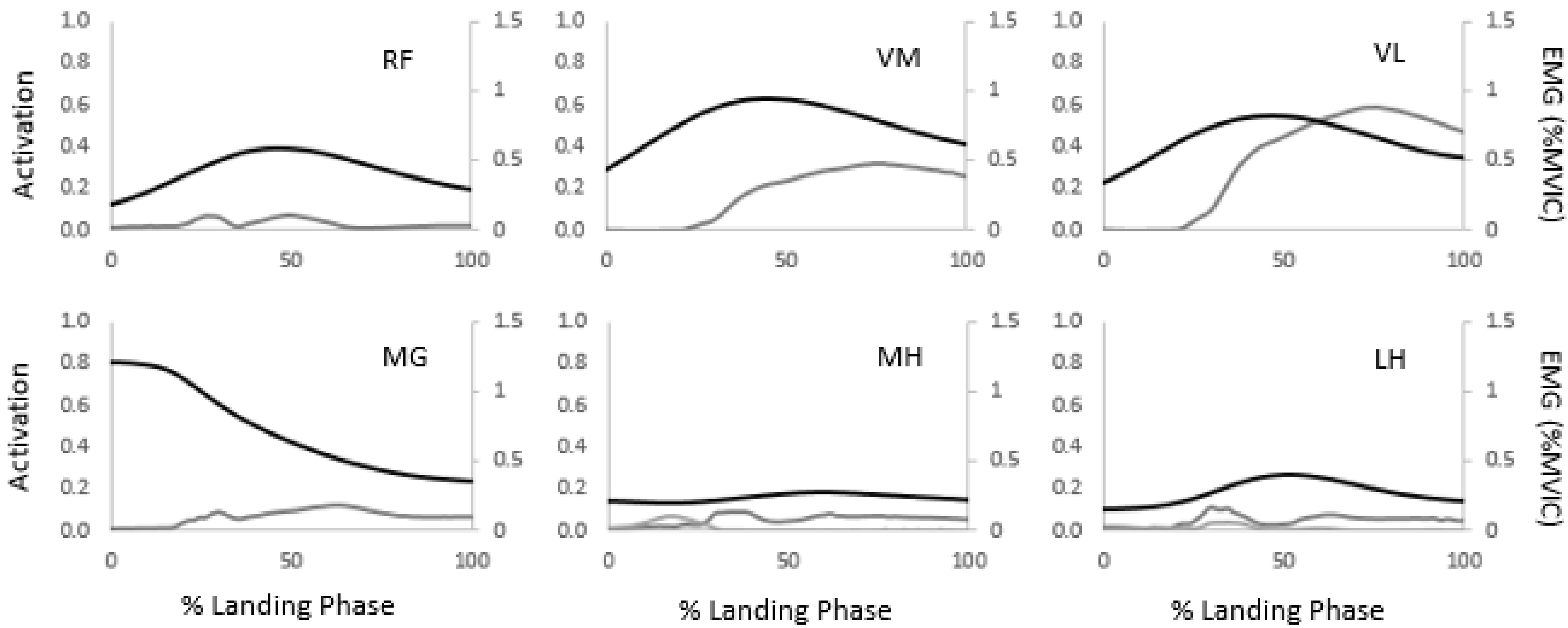

Figure 5 Caption. Muscles included in figure are: Rectus Femoris (RF), Vastus Medialis (VM), Vastus Lateralis (VL), Medial Gastrocnemius (MG), Medial Hamstring (MH), and Lateral Hamstring (LH). Black lines are electromyographic recordings of each muscle and grey lines are the simulated activations.

Figure 6. Ensemble muscle activations measured using electromyography compared to simulations during 40 cm drop landings.

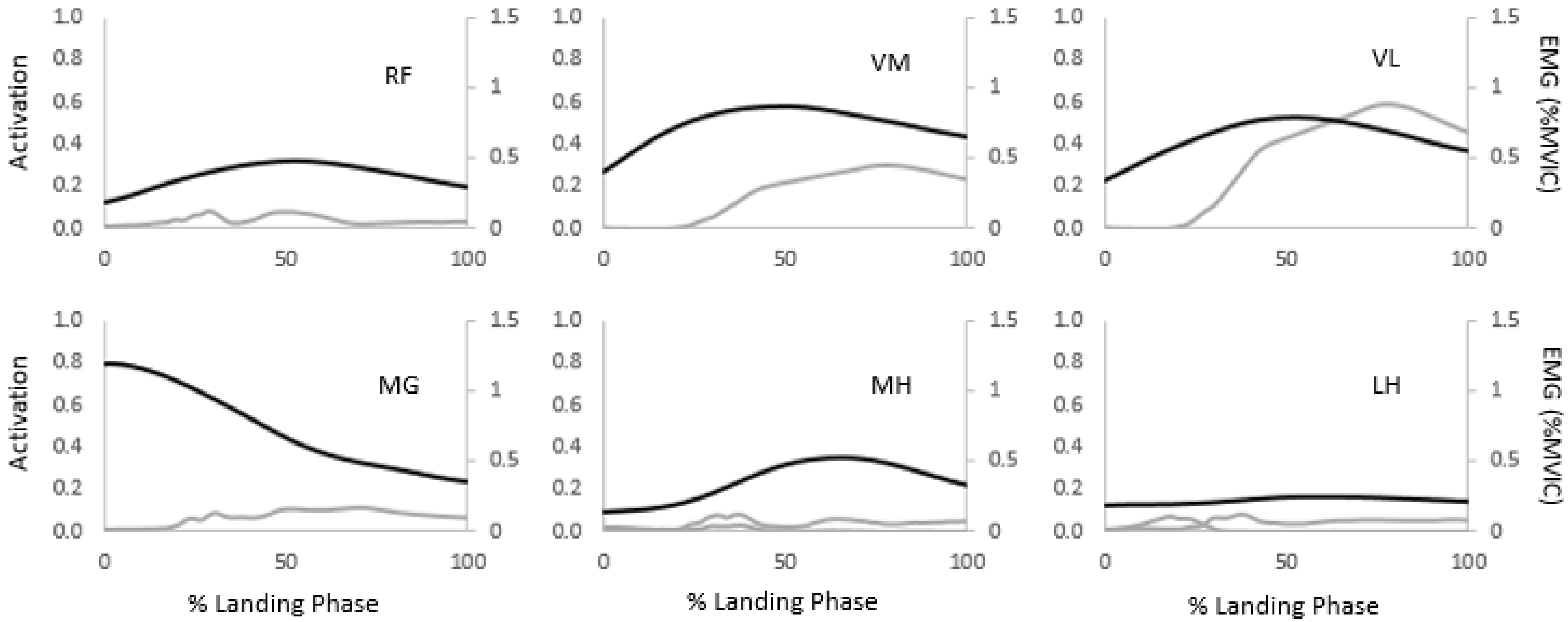

Figure 6 Caption. Muscles included in figure are: Rectus Femoris (RF), Vastus Medialis (VM), Vastus Lateralis (VL), Medial Gastrocnemius (MG), Medial Hamstring (MH), and Lateral Hamstring (LH). Black lines are electromyographic recordings of each muscle and grey lines are the simulated activations.

CHAPTER 7: CONCLUSION

Numerous aspects of ACL injury mechanisms during landing and cutting tasks exist: mechanics, muscle strength, and activation patterns. Morphological parameters of the ACL, such as ACL diameter, could also be important factors for ACL loading and injury. Thus, this dissertation set out to determine the relationships of these known factors of ACL injury with ACL loading during drop landing tasks. This dissertation also sought to determine the reliability of using ultrasound to measure ACL diameter in a research setting.

Aim #1 examined the intra- and inter-rater reliability of ultrasound of ACL diameter. The results of Aim #1 show that researchers with varying ultrasound experience can measure diameter of the full ACL with low to moderate reliability. During examination of the ACL, the distal portion of the ligament was notably difficult to identify, which could explain the lower agreement of the intra and inter-rater reliabilities. As the ACL is made of two separate bundles, of which the anteromedial (AM) bundle has been previously examined using diagnostic imaging, this work also examined reliability of measuring the more superficial AM bundle. The AM bundle had more clearly defined superior and inferior borders, which resulted in increased the intra and inter-rater reliabilities (reaching "good" levels of reliability). This work is a first step in providing support for the usage of diagnostic ultrasound for imaging of ACL anthropomorphic measurements such as ACL diameter in common research settings (e.g. raters with moderate to low experience).

Aim #2 focused on the relationship of strength measurements and physical activity levels with ACL diameter measured via ultrasound. Moderate positive associations were found between ACL size and quadriceps strength and adductor strength. ACL size was also significantly negatively correlated to hip abductor to adductor strength ratios. No significant correlations were found with any time spent exercising at high, moderate, or low exercise intensities. Previous literature has established that an increase in quadriceps strength or diminished hamstring strength can result in increased ACL loading. The same has also been found with hip abductor and adductor musculature. With an imbalance of hip adductor strength to hip abductor strength, the knee may be placed in a position where the ACL is stressed more. It is also speculated that this increase in quadriceps and adductor strength may cause more healthy stress in the ACL. This healthy stress may cause ACL hypertrophy. Thus, this work appears to agree with the previous literature in that increased quadriceps and hip adductor are related to increased ACL size, possibly due to healthy ACL loading. Further research is needed to examine if other influences of increased ACL loading or increased ACL stress is related to increase in ACL size.

The third and final aim of this dissertation examined the relationship between ACL loading with ACL diameter, passive knee joint laxity and pre-activation of the Rectus Femoris, Vastus Lateralis, Vastus Medialis, Lateral Hamstring, Medial Hamstring, Medial Gastrocnemius, and Gluteus Medius. This study found that only pre-activation of the Lateral Hamstring and the pre-activation ratio of the Medial to Lateral Hamstring were related to ACL loading. The increase in pre-activation of the Lateral Hamstring has the ability to open the medial compartment of the knee joint during landing, placing the knee and ACL at increased risk of injury. The lack of significant relationships found between ACL loading, quadriceps activation and knee joint laxity are surprising as numerous previous studies have found ACL injuries are related to increased quadriceps activation, reduced hamstrings activation and increased knee joint laxity.

The current study shows that ultrasound can be utilized and reliable between multiple researchers that are not trained radiographers. It also shows that ultrasound is more reliable when measuring and assessing the anteromedial bundle of the ACL compared to the full ACL. Further research may need to examine ways to improve the reliability of measuring the full ACL when utilizing ultrasound when MRI or arthroscopic means are unavailable. ACL size also needs to be further examined. More evidence is needed to examine if the ACL of humans can be hypertrophied and if any other ACL injury mechanisms or activities are correlated to ACL size. The current study sheds more light on injury mechanisms directly correlated to ACL injuries. As this study has found, numerous ACL injury risk mechanism were not directly correlated to ACL loading or ACL. Further work may need to examine if other ACL injury risk factors are correlated to ACL loading.

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APPENDIX

Appendix A: Informed Consent Document

INFORMED CONSENT DOCUMENT
OLD DOMINION UNIVERSITY

PROJECT TITLE: The reliability of ultrasound measure on ACL thickness

INTRODUCTION

The purposes of this form are to give you information that may affect your decision whether to say YES or NO to participation in this research, and to record the consent of those who say YES. This project is being carried out to determine the reliability of ultrasound measures on ACL thickness. The research will be conducted at Old Dominion University in the Neuromechanics Laboratory.

RESEARCHERS

Johanna Hoch, PhD, Responsible Project Investigator, Assistant Professor, College of Health Sciences, School of Physical Therapy and Athletic Training, Old Dominion University

Hunter Bennett, PhD, Assistant Professor, Darden College of Education, Department of Human Movement Sciences, Old Dominion University

Zach Sievert, Graduate Student, Department of Human Movement Sciences, Old Dominion University

Cortney Armitano, Graduate Student, Department of Human Movement Sciences, Old Dominion University

DESCRIPTION OF RESEARCH STUDY

The purpose of this study is to test the reliability of ultrasound measure on ACL thickness. A total of 40 participants will participate in this study. You will report to the Neuromechanics Laboratory, SRC 1007, for one, 30 minute testing session.

At the beginning of the testing session you will be informed of the study procedures and you will provide written consent in accordance with institutional guidelines. During the visit to the lab, you will be asked to fill out a Medical History and Physical Activity Questionnaire. This includes questions pertaining to age, physical activity, lower body injury(ies), recent head injury(ies), or any medications that may cause dizziness, etc.

Once you have completed the medical history and physical activity questionnaire to determine eligibility, researchers will measure your height and weight. Once completed you will be asked to sit in an upright position on an examining table. Next, you will be asked to bring one foot up on the table so that you are flexed at the hip and have knee flexion between 90 degrees and maximal knee flexion. Once in the proper testing position, researchers will apply ultrasound gel to your knee and researchers will use the ultrasound probe to take images of the intracapsular space.

EXCLUSIONARY CRITERIA

To be eligible to participate you must be physically active three days a week for at least 30 minutes.

You will not be able to participate in the study if you:

- have suffered any injuries to the lower extremities within the last six months.

- have ever had surgery to your knees.
- are pregnant or think you may be pregnant.

RISKS AND BENEFITS

RISKS:

If you decide to participate in this study, then you may experience general and/or minor skin irritation due to the ultrasound gel. It is also possible, although unlikely, that you may experience musculoskeletal soreness or injury such as a strain through constant high knee flexion.

To reduce the above risks, care will be taken when applying and removing the ultrasound gel. If you feel any soreness or irritation while participating in this study, please tell the investigators as soon as possible. The investigators will provide you with the appropriate information for treating these problems, based on our level of expertise. If you are injured while participating in this research study, you will initially be provided care by the investigator(s) and will then be referred to the Student Health Services (students) or your personal physician (non-students) for follow-up care.

There is a small risk of loss of confidentiality. To minimize this risk all information gathered from you will be confidential in nature and stored in the laboratory of the principle investigator (SRC 1007). Only the principal investigator and research personal will have access to these files.

BENEFITS:

There are no direct benefits from participating in this study. The information obtained in this study will expand our knowledge base leading to beneficial changes in the future

COSTS AND PAYMENTS

The researchers are unable to give you any payment for participating in this study. Students in Exercise Science courses may be offered extra credit for participation per individual course policy. Total point value for participation will be determined by course instructor. Students in courses offering extra credit for research participation may receive extra credit of equal point value by completing an alternative assignment

NEW INFORMATION

If the researchers find new information during this study that would reasonably change your decision about participating, then they will give it to you.

CONFIDENTIALITY

The researchers will provide confidentiality to all documents regarding patient information, including questionnaires and test results, by storing information in a safe, locked location. The results of this study may be used within reports, presentations or publication; but all personal identifiers will be disregarded. However, your records may be subpoenaed by court order or inspected by government bodies with oversight authority. Following the completion of the study, all subject information will be destroyed.

WITHDRAWAL PRIVILEGE

In the event that you no longer wish to participate, you have the right to discontinue participation for this study. Even if you initially wish to participate you can withdraw at any time. Your decision will not affect your relationship with Old Dominion University or cause a loss of benefits to which you might otherwise be entitled. In addition, the researcher has the right to withdraw your participation if they find potential problems with your continued participation.

COMPENSATION FOR ILLNESS AND INJURY

If you participate, then your consent in this document does not waive any of your legal rights. In the event that you are harmed or injured from participating in this study, neither Old Dominion University nor the researchers are able to give you any compensation, including money, insurance coverage, or free medical care. In the event that you suffer any injury from participation in this study, you may contact Dr. Johanna Hoch at (757) 683-6131, Zach Sievert at (757)-683-3048, Dr. George Maihafer, the current IRB chair, at (757) 683-4520, or Office of Research (757) 683-3460 at Old Dominion University.

VOLUNTARY CONSENT

By signing this form, you are saying several things. You are saying that you have read this form or have had it read to you, that you are satisfied that you understand this form, the research study, and its risks and benefits. The researchers should have answered any questions you may have had about the research. If you have any questions later on, then the researchers should be able to answer them:

Dr. Johanna Hoch (757) 683-6131

Zach Sievert (757)-683-3048

If at any time you feel pressured to participate, or if you have any questions about your rights or this form, then you should call Dr. George Maihafer, the current IRB chair, at 757-683-4520, or the Old Dominion University Office of Research, at 757-683-3460.

And importantly, by signing below, you are telling the researcher YES, that you agree to participate in this study. The researcher should give you a copy of this form for your records.

Subject's Printed Name & Signature	**Date**

INVESTIGATOR'S STATEMENT

I certify that I have explained to this subject the nature and purpose of this research, including benefits, risks, costs, and any experimental procedures. I have described the rights and protections afforded to human subjects and have done nothing to pressure, coerce, or falsely entice this subject into participating. I am aware of my obligations under state and federal laws, and promise compliance. I have answered the subject's questions and have encouraged him/her to ask additional questions at any time during the course of this study. I have witnessed the above signature(s) on this consent form.

Investigator's Printed Name & Signature	**Date**

INFORMED CONSENT DOCUMENT
OLD DOMINION UNIVERSITY

PROJECT TITLE: The Effects of Physical Activity on ACL Loading and Size.

INTRODUCTION

The purposes of this form are to give you information that may affect your decision whether to say YES or NO to participation in this research, and to record the consent of those who say YES. This project is being carried out to determine the effects of several currently available hip joint center prediction methods on assessing walking, running, and jump landing tasks. The research will be conducted at Old Dominion University in the Neuromechanics Laboratory.

RESEARCHERS

Hunter Bennett, PhD, Assistant Professor, Responsible Project Investigator, Department of Human Movement Sciences, Old Dominion University

Zachary Sievert, Graduate Student, Department of Human Movement Sciences, Old Dominion University

Cortney Armitano, Graduate Student, Department of Human Movement Sciences, Old Dominion University

DESCRIPTION OF RESEARCH STUDY

The purpose of this study is to determine the effects of physical activity levels on ACL loading and size. A total of 100 male participants will participate in this study. You will report to the Neuromechanics Laboratory, SRC 1007, for one, 1 hour and 30 minute testing session.

At the beginning of the testing session you will be informed of the study procedures and you will provide written consent in accordance with institutional guidelines. During the visit to the lab, you will be asked to fill out a Medical History and Physical Activity Questionnaire. This includes questions pertaining to age, physical activity, lower body injury(ies), recent head injury(ies), or any medications that may cause dizziness, etc.

Once you have completed the medical history and physical activity questionnaire to determine eligibility, researchers will measure your height, weight, and shoe size. You will be instructed the put on spandex shorts and the provided laboratory shoes. All participants will be asked to sit in a position with their knee flexed between 90 degrees and maximal knee flexion as researchers measure ACL thickness with diagnostic ultrasound. Passive knee joint laxity measures will also be performed by a licensed athletic trainer, where participants will be asked to lay in a supine position with the knee flexed to 30 degrees as the athletic training measures joint laxity with a KT-2000 arthometer.. Researchers will then palpate the thickest portion of your quadriceps, hamstrings, calf, and glute muscles. Next, the skin above the muscle will be abraded, cleaned with alcohol wipes, and have passive muscle electrodes attached. Isometric strength testing of the knee extensors and flexors as well as the hip abductors and adductors will be recorded. Retro reflective motion capture markers will be attached to Velcro straps placed around your trunk, pelvis, thighs, lower legs, and shoes so that the motion capture system can track movements of your trunk and lower extremity during several tasks. You will then be asked to perform landings from a box with height equal to your jumping ability, from a height of 40 cm, as well as a jump land jump cut maneuver.

EXCLUSIONARY CRITERIA

You will not be able to participate in the study if you:

- have suffered any injuries to the lower extremities within the last six months.
- have diagnosed joint disease
- have ever had major lower extremity surgery or joint replacement.

RISKS AND BENEFITS

RISKS:

If you decide to participate in this study, risks are considered minimal. The jumping and landing to be tested in this study are no different from what you would perform in normal daily movements or moderate exercises. It is possible the skin above each muscle tested using electromyography may be irritated due to abrasions or cleaning with alcohol wipes during the skin preparation process.

In consideration of all risks, the researchers will be present and interactive during all tasks. Should any injury occur during the course of testing, standard first aid procedures will be administered as necessary. At least one researcher with a basic knowledge of athletic training and/or first aid procedures will be present at each test session. If you feel any injury while participating in this study, please tell the investigators as soon as possible. If you are injured while participating in this research study, you will initially be provided care by the investigator(s) and will then be referred to the Student Health Services (students) or your personal physician (non-students) for follow-up care.

There is a small risk of loss of confidentiality. To minimize this risk all information gathered from you will be confidential in nature and stored in the laboratory of the principle investigator (SRC 1007). Only the principal investigator and research personal will have access to these files.

BENEFITS:

There are no direct benefits from participating in this study. The information obtained in this study will expand our knowledge base leading to beneficial changes in the future.

COSTS AND PAYMENTS

The researchers are unable to give you any payment for participating in this study. Students in Exercise Science courses may be offered extra credit for participation per individual course policy. Total point value for participation will be determined by course instructor. Students in courses offering extra credit for research participation may receive extra credit of equal point value by completing an alternative assignment

NEW INFORMATION

If the researchers find new information during this study that would reasonably change your decision about participating, then they will give it to you.

CONFIDENTIALITY

The researchers will provide confidentiality to all documents regarding patient information, including questionnaires and test results, by storing information in a safe, locked location. The results of this study may be used within reports, presentations or publication; but all personal identifiers will be disregarded. However, your records may be subpoenaed by court order or inspected by government bodies with oversight authority. Following the completion of the study, all subject information will be destroyed.

WITHDRAWAL PRIVILEGE

In the event that you no longer wish to participate, you have the right to discontinue participation for this study. Even if you initially wish to participate you can withdraw at any time. Your decision will not affect your relationship with Old Dominion University or cause a loss of benefits to which you might otherwise be entitled. In addition, the researcher has the right to withdraw your participation if they find potential problems with your continued participation.

COMPENSATION FOR ILLNESS AND INJURY

If you participate, then your consent in this document does not waive any of your legal rights. In the event that you are harmed or injured from participating in this study, neither Old Dominion University nor the researchers are able to give you any compensation, including money, insurance coverage, or free medical care. In the event that you suffer any injury from participation in this study, you may contact Dr. Hunter Bennett at (757) 683-4387, Zach Sievert at (757)-683-3048, Dr. Tancy Vandecar-Burdin, the current IRB chair, at (757) 683-3802 (tvandeca@odu.edu), or Office of Research (757) 683-3460 at Old Dominion University.

VOLUNTARY CONSENT

By signing this form, you are saying several things. You are saying that you have read this form or have had it read to you, that you are satisfied that you understand this form, the research study, and its risks and benefits. The researchers should have answered any questions you may have had about the research. If you have any questions later on, then the researchers should be able to answer them:

Dr. Hunter Bennett (757) 683-4387

Zachary Sievert (757)-683-5676

If at any time you feel pressured to participate, or if you have any questions about your rights or this form, then you should call Dr. Tancy Vandecar-Burdin, the current IRB chair, at 757-683-3802 (tvandeca@odu.edu), or the Old Dominion University Office of Research, at 757-683-3460.

And importantly, by signing below, you are telling the researcher YES, that you agree to participate in this study. The researcher should give you a copy of this form for your records.

Subject's Printed Name & Signature	**Date**

INVESTIGATOR'S STATEMENT

I certify that I have explained to this subject the nature and purpose of this research, including benefits, risks, costs, and any experimental procedures. I have described the rights and protections afforded to human subjects and have done nothing to pressure, coerce, or falsely entice this subject into participating. I am aware of my obligations under state and federal laws, and promise compliance. I have answered the subject's questions and have encouraged him/her to ask additional questions at any time during the course of this study. I have witnessed the above signature(s) on this consent form.

Investigator's Printed Name & Signature	**Date**

Appendix B: Medical History & Physical Activity Questionnaire

<u>Medical History & Physical Activity Questionnaire</u>

Please answer the following questions to the best of your ability:

Gender:
☐Male ☐Female

Race/ethnicity *(please check all that apply)*:
☐American Indian/Alaska Native
☐Asian
☐Native Hawaiian or other Pacific Islander
☐Hispanic or Latino
☐Black or African American
☐White

Which leg would you use to kick a ball?
☐Right ☐Left

To be completed by investigator:

Age:________ yr Height: ________ m Mass: ________ kg

<u>Medical History Questionnaire</u>
For your safety, a list of conditions that would make you unable to participate in this study has been prepared. Please read this list carefully and consider whether any of the conditions apply to you. If any of these conditions are true for you, you will not be able to participate in this study. For each condition, please indicate "yes" or "no" if this is true or not for you.

☐Yes ☐No Are you currently physically active at a moderate level for at least 30 minutes/day, at least 3 days of the week?

☐Yes ☐No Do you have a medical condition that may impair your balance performance (i.e. concussion, neurological impairments, etc)?

☐Yes ☐No Are you taking medications/drugs that may make you dizzy or make you tired (i.e. cold medications, sleeping medications, muscle relaxants)?

☐Yes ☐No Have you <u>ever</u> had a lower extremity injury that caused you to decrease the amount of physical activity you undertake? If yes, please complete the following:

☐Yes ☐No Hip injury(ies)
If yes, approximately how many injuries?__________________________
☐Yes ☐No Knee injury(ies)
If yes, approximately how many injuries? __________________________
☐Yes ☐No Ankle/foot injury(ies)
If yes, approximately how many injuries? __________________________

☐Yes ☐No Have you had, in the last 6 months, a lower extremity injury that caused you to decrease the amount of physical activity you undertake?

☐Yes ☐No Do you currently have any lower extremity pain or injury(ies)?

☐Yes ☐No Have you ever had major orthopedic surgery on your lower extremities?

☐Yes ☐No Are you pregnant or do you have reason to believe that you may be pregnant?

Physical Activity Questionnaire

1. On average, how many days a week have you exercised over the past six months?

   _____ days per week

   a. During your exercise periods, how many days did you spend performing strenuous or vigorous activity (heart beating rapidly) such as heavy lifting, running, jogging, hockey, football, soccer, basketball, or vigorous long distance bicycling? How much time is spent performing vigorous activity?

      _____ days per week _____ minutes per week

   b. During your exercise periods, how many days did you spend performing moderate activity (Not Exhausting) such as fast walking, baseball, tennis, easy bicycling, volleyball, badminton, or easy swimming? How much time is spent performing moderate activity?

      _____ days per week _____ minutes per week

   c. During your exercise periods, how many days did you spend performing mild activity (Minimal Effort) such as leisurely walking, bowling, or golf? How much time is spent performing mild activity?

      _____ days per week _____ minutes per week

2. How many times per week do you participate in a resistance training?

   _____ days per week

   a. How many times per week do you perform lower extremity resistance training?

      _____ days per week

   b. How many exercises do you perform that specifically utilize knee flexors and extensors?

      Flexor exercises _________ Extensor exercises_________

3. How many times per week do you perform cardiovascular exercises?

_____ days per week

a. How many times per week while performing cardiovascular activities are you performing cutting, jumping, and landing maneuvers?

   _____ days per week

VITA

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<u>Education</u>:

Old Dominion University

Ph.D. (Applied Kinesiology) – August 2019

Concentration: Biomechanics, ACL Prevention and Injury Mechanisms

Dissertation: "The Effects of Physical Activity on ACL thickness and ACL Loading."

Old Dominion University

M.S. (Exercise Science) –August 2014

Thesis: "The Effects of Squat Depth on ACL, PCL, and Tibiofemoral Contact Forces."

College of Mount St. Joseph

B.S. (General Studies) – May 2012

<u>Professional Experience</u>:

Teaching:

Instructor, Virginia Commonwealth University — Fall 2017-
Musculoskeletal Structure and Movement (HPEX 374)
Resistance Training for Health and Performance (HPEX 380)
Professional Certification Seminar (HPEX 480)

Graduate Teaching Assistant, Old Dominion University — Spring 2017
Prevention and Care of Injuries Lecture (EXSC 240)

Graduate Teaching Assistant, Old Dominion University — 2015-2017
Biomechanics Lecture (EXSC 417)

Graduate Teaching Assistant, Old Dominion University — 2014-2017
Anatomical Kinesiology Lecture (EXSC 322)

Graduate Teaching Assistant, Old Dominion University 2013-2014
Biomechanics Lab (EXSC 417)

Adjunct Instructor, Old Dominion University 2013-2016
Anatomical Kinesiology Lab (EXSC 322)
Biomechanics Lab (EXSC 417)

Research:

Neuromechanics Laboratory, Old Dominion University 2012-2017
Research Assistant
Administrate the day-to-day technical and scientific operations of the lab. Oversee funded research, undergraduate and graduate student research projects.

Publications:

Weinhandl JT, Irmischer BS, & **Sievert ZA**. (*2015*). Sex differences in unilateral landing mechanics from absolute and relative drop heights. *The Knee. 22(4):298-303.*

Weinhandl JT, Irmischer BS, **Sievert ZA** & Fontenot K. (2016). Influence of sex and limb dominance on lower extremity joint mechanics during unilateral land-and-cut maneuvers. *Journal of Sports Sciences*.

Weinhandl JT, Irmischer BS & **Sievert ZA**. (2017). Effects of gait speed on femoroacetabular joint forces. Applied Bionics and Biomechanics. *Applied Bionics and Biomechanics*.

Works in Review:

Peel SA, Schroeder LE, **Sievert ZA**, Weinhandl JT. (2018). Comparing Anterior Cruciate Ligament Injury Risk Variables between Unanticipated Cutting and Stopping Tasks. *Journal of Applied Biomechanics*

Works in Progress:

Sievert ZA, Hill L, Ringleb SI & Weinhandl JT. The effects of a 90° and deep squat on kinematic and kinetic variables of the lower extremity.

Sievert ZA, Hill L, Ringleb SI & Weinhandl JT. The effects of squat depth on electromyography of eight lower extremity muscles.

Sievert ZA, Armitano C, Ringleb SI, Bennet HJ, Weinhandl JT. Are theoretical models appropriate to predict gait speeds?

Sievert ZA, Hill L, Ringleb SI & Weinhandl JT. The effects of squat depth on ACL, PCL, and tibiofemoral contact forces.

Sievert ZA, Bennett HJ & Weinhandl JT. Male and female differences in lower extremity mechanics during the barbell backsquat.

Sievert ZA, Bennett HJ, Armitano C, Ringleb SI, Hoch J, Weinhandl JT. The reliability of Ultrasound Measures on ACL thickness.

Presentations:

Sievert ZA, Ringleb SI, Hill LC & Weinhandl JT. The effects of squat depth on electromyography of eight lower extremity muscles. Medicine & Science in Sports & Exercise. 48(5S):881. May 2016; Boston, MA.

Pitman E, **Sievert ZA**, Fontenot KC, Mutchler JA & Weinhandl JT. 2015. The effect of posterior foot placements on sit-to-stand kinetics, energetics and muscle activity. Proceedings of the 39th Annual Meeting of the American Society of Biomechanics. August 2015; Columbus, OH.

Sievert ZA, Ringleb SI, Hill LC & Weinhandl JT. The effects of squat depth on right leg kinematics and kinetics. Proceedings of the 39th Annual Meeting of the American Society of Biomechanics. August 2015; Columbus, OH.

Fontenot KC, Pitman E, **Sievert ZA** & Weinhandl JT. The influence of gender and limb dominance on lower extremity joint mechanics during land-and-cut maneuvers. Proceedings of the 7th World Congress of Biomechanics. July 2014; Boston, MA.

Irmischer BS, **Sievert ZA** & Weinhandl JT. Effects of walking speed on hip joint forces. Proceedings of the 7th World Congress of Biomechanics. July 2014; Boston, MA.

Pitman E, Fontenot KC, **Sievert ZA** & Weinhandl JT. Comparison of unilateral drop landings and land-and-cut maneuvers. Proceedings of the 7th World Congress of Biomechanics. July 2014; Boston, MA.

Sievert ZA, Irmischer BS, & Weinhandl JT. Gender differences in unilateral landing mechanics from absolute and relative drop heights. American Society of Biomechanics 37th Annual Meeting. September, 2013. Omaha, NE.

Current Research Projects

"The Reliability of Ultrasound Measures on Anterior Cruciate Ligament Thickness."

"Effects of Hip Joint Center Method on Hip Kinematics and Kinetics."

"The Effects of Segmental Parameters on Gait."

"The effects of a 12 week resistance training program on ACL loading and ACL injury risk in 18-25 year olds."

"The influence of squat depth on lower extremity kinematics, kinetics, and muscle activations."

Professional Meetings Attended:

2016 – 63rd Annual Meeting of the American College of Sports Medicine. Boston, MA.

2013 – 37th Annual Meeting of the American Society of Biomechanics. Omaha, NE.

Professional Memberships:

American Society of Biomechanics 2013-Present

American College of Sports Medicine 2014-present

Computer Skills:

Microsoft Word, Excel, and PowerPoint
Vicon Motion Capture System
Visual 3D
Opensim
Matlab