INFLUENCE OF THIGH MUSCLE FORCES ON ANTERIOR CRUCIATE LIGAMENT FORCES DURING SINGLE-LEG LANDING FROM THREE DIFFERENT HEIGHTS

by

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Over 200,000 anterior cruciate ligament (ACL) injuries occur every year amounting to billions of dollars being spent on the ACL annually. While the quadriceps muscle produces an anterior shear force on the tibia that causes the ACL to strain, the hamstrings muscle can protect the ACL by producing a posterior shear force to the tibia reducing the strain. When the hamstrings contract simultaneously with the quadriceps, ACL strains are considerably less compared to isolated quadriceps forces, thus the balance of hamstring and quadriceps muscle forces play a critical role in determining the forces on the ACL. During dynamic landing tasks, quadriceps demands increase as the landing height increases, which may cause the ACL to be more susceptible to injury. The purpose of this study was to determine the relationship of the quadriceps and hamstring muscle forces on ACL forces during single-leg landing from three different heights. We hypothesized that the ratio between hamstrings and quadriceps muscle forces would be negatively correlated to peak ACL forces during landing from three different heights. We anticipated that the hamstring to quadriceps ratio would decrease as landing height increased primarily due to the increased quadriceps demands.

Three males with an average height of .75±0.07m with an average mass of 74.08±8.66kg and three females with an average height of 1.70±0.04m and an average mass of 55.93±6.83kg landed on their right leg from three different heights, 15cm, 30cm, and 45cm. Musculoskeletal modeling was used to estimate muscle forces. Regression analyses predicted the ACL forces from all three heights, and the heights pooled together.

The results showed that the quadriceps muscles forces were strongly positively correlated to the peak ACL force while the hamstrings muscle forces were not significantly correlated to peak ACL force. Linear analysis showed the hamstring to quadriceps ratio to be moderately negatively correlated with peak ACL force ($r^2 = 0.278$), but nonlinear curve analysis showed a stronger relationship between these variables ($r^2 = 0.425$). However, as the landing height increased, these linear and nonlinear relationships both decreased. This signifies that another factor was contributing to the peak ACL force especially at higher heights. The combined influence of ground reaction forces and the hamstring to quadriceps ratio revealed that as landing height increased the ground forces became more of a factor in predicting peak ACL forces compared to the hamstring to quadriceps ratio being the dominant predictor at the lowest heights.

In conclusion, the data support our hypothesis the hamstring to quadriceps ratio was inversely related to the peak ACL force although the strength of this relationship was height dependent. Further, as landing height increases, the ground reaction forces become the stronger predictor of peak ACL forces compared to the hamstring to quadriceps force ratio.

Influence of Thigh Muscle Forces on Anterior Cruciate Ligament Forces

during Single-Leg Landing from Three Different Heights

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CHAPTER 1: INTRODUCTION

Every year in the United States, there are over 200,000 injures to the anterior cruciate ligament (ACL) (Albright, Carpenter et al. 1999). The ACL can tear from a direct hit on the knee or it can tear during a non-contact deceleration motion such as landing or cutting. Once the ligament becomes injured, it may be surgically repaired. With every injury costing approximately $17,000 (Hewett, Lindenfeld et al. 1999), this amounts to billions of dollars spent every year on the tear in the ACL. One of the long-term consequences to having an injured ACL is that the individual may develop early onset knee osteoarthritis (OA). Even though the knee joint range of motion and knee ligament strength may be almost back to pre-injury status, physical changes in the knee joint occur and symptoms of stiffness, pain, and functional impairment increase as time progresses (Lohmander, Englund et al. 2007). Therefore, identifying factors contributing to the original ACL injury would be advantageous in order to ultimately prevent these injuries and the long-term effects of knee OA.

The quadriceps, acting as an antagonist to the anterior cruciate ligament, creates an anterior shear force on the knee joint, which strains the ACL. The highest strain placed on the ACL by the isometric quadriceps contraction occurs between knee flexion angles of 0° to approximately 40° (Beynnon and Fleming 1998). The hamstrings counteract the quadriceps by creating a posterior shear force on the knee joint. This unloads the ACL, decreasing strain on the ligament. The combined effects of the hamstrings and quadriceps forces therefore determine the extent of ACL forces.

With some knowledge of how hamstring, quadriceps, and the combined muscles affect the knee loads, we can now investigate how different landing heights can affect these muscle forces and the impending ACL forces. As landing height increases, the elevated ground reaction forces (Seegmiller and McCaw 2003) lead to an increase in knee extensor moments (Zhang, Bates et al. 2000) and presumably the hamstring-to-quadriceps force ratio decreases thereby placing elevated forces on the ACL.

In order to fully understand how the hamstrings and quadriceps affect knee loads and thus affect the ACL, one might want to examine the individual muscles forces. However, surface electromyography (EMG) is very limited and intramuscular EMG can be invasive. The use of musculoskeletal modeling can recreate the task performed, and the model and the individual muscles can be observed and even manipulated to the investigators specifications. Musculoskeletal modeling has been used to investigate walking and running and a value of r^2 = 0.94 was found to predict the knee joint moments from the musculoskeletal model (Lloyd and Besier 2003). Knee flexion-extension moments that are derived from individual muscle forces using an EMG-driven model have also been predicted with musculoskeletal modeling with a value of $r^2 = 0.81 \pm 0.09$ for walking and $r^2 = 0.89 \pm 0.07$ (Besier, Fredericson et al. 2009).

Hypothesis

No one knows with certainty how the anterior cruciate ligament is torn. The thigh muscles are involved but to what extent of this involvement are unknown. We do know that the quadriceps cause an anterior shear force that causes the tibia to move anteriorly causing the ACL to strain while the hamstrings create a posterior shear force decreasing that load. Therefore the hamstring to quadriceps ratio may be a valid indicator of ACL forces during landing. Because knee extensor moments (which can be primarily attributed to the net contributions from the quadriceps and hamstrings muscles) increase with landing height, we hypothesized that the hamstring to quadriceps force ratio would be inversely related to peak AC forces during landing from three different heights. We anticipated that the hamstring to quadriceps ratio would decrease as landing height increased primarily due to the increased quadriceps demands, thus more forces would be imparted on the ACL at these higher heights.

Statement of Purpose

The purpose of this study is to determine the relationship between the combined quadriceps and hamstrings muscle forces on ACL forces during single-leg landings from three different heights.

Delimitations

1) All subjects were healthy, with no history of lower extremity injuries and/or surgeries.

2) Subjects were young adults between the ages of 18 – 30.

3) All subjects had a Body Mass Index of less than 25 kg/m^2.

4) Subjects had a background of participating in jumping activities.

5) Subjects were able to complete the task of jumping from the three predetermined heights landing on the right leg.

6) Maximal Voluntary Isometric Contractions were tested using the right leg.

7) Recreational athletes were only used due to the lack of population size and effect of intercollegiate athletes.

8) Subjects landed with both arms crossed their chest. This was to reduce interference of the reflective markers and to control the task of landing

Limitations

1) Knee is a hinge joint and patella tendon is inelastic in the musculoskeletal model.

2) Fatigue may occur because of repeated dynamic task being performed.

3) Errors can occur between measured and simulated values.

4) Quality of dynamic simulation is dependent on the quality of kinematic data.

Assumptions

1) The markers and EMG setup do not interfere with the performed task.

2) All information obtained from the subjects is true.

3) Data from the force plate, motion analysis, and EMG systems are accurate.

4) If a forward dynamic movement is similar to the experimental motion, then the muscle forces that produced the forward dynamics motion are valid.

Operational Definitions

1) ACL Strain – The change in length of the ACL divided by the original length of the ACL and multiplied by 100. The elongation of the ligament which is changed by the knee flexion angle and the intrinsic measure of the ACL.

2) ACL Forces – The sum of all forces that cross the knee that affects the ACL (i.e. Anterior shear force, Posterior shear force, Patellar tendon Forces).

3) Knee Shear Forces - Summation of both knee muscle and knee joint reaction force components acting parallel to the surface of the tibia.

4) Musculotendon Unit – A group of muscle fibers and the connective tissue (tendon) attached to that muscle group.

5) Forward Dynamics Simulation - A simulation of human movement driven by the calculated muscle forces from the musculoskeletal model and using a computed muscle control (CMC) mathematical algorithm.

6) *In vivo* – The research study was conducted inside a living subject.

7) *In vitro/In situ* – The research study was conducted using cadaver subjects.

CHAPTER 2: REVIEW OF LITERATURE

The purpose of this study is to determine the influence of the quadriceps and hamstrings muscle forces on ACL forces during single-leg landing from three different heights. In this review of literature, the following issues will be covered: 1) Epidemiology and Economic Cost of ACL injury, 2) Contributions of the Quadriceps and Hamstrings on Knee Loads, 3) Biomechanics of Landing Height and the Effects of Landing Height and 4) Summary.

Epidemiology and Economic Costs of ACL Injury

The Anterior Cruciate Ligament (ACL) is the most common injured knee ligament. Over 200,000 ACL injuries occur in the United States every year (Albright, Carpenter et al. 1999) with a cost of approximately $17,000 per injury (Hewett, Lindenfeld et al. 1999). This totals to billions of dollars that are spent annually just on the surgery of the anterior cruciate ligament. Almost 80% of ACL injuries that occur are non-contact in nature (Renstrom, Ljungqvist et al. 2008). A non-contact injury is where there is no direct physical contact on the knee or body, from either a person or an object, and can be described as landing from a jump, cutting or decelerating during an activity. In the National Football League, 16% of knee injuries involved a tear in the ACL. This accounts for 2% of all injuries in the league (Bradley, Klimkiewicz et al. 2002). Even though ACL injuries are more highly publicized in the professional sports world, they are not relegated to just professional athletes. Among those that injure the ligament in the population, it is most common in recreationally active individuals. Once injured, the ACL requires surgery and extensive rehabilitation to get the ligament and knee joint back to normal function. One of the long-term effects of an injured anterior cruciate ligament is that it can cause early onset knee osteoarthritis (KOA) before the age of 30 (Lohmander, Ostenberg et al. 2004). This causes a decreased quality of life from age 30 to 50.

Osteoarthritis is usually associated with pain, stiffness and functional impairment of joints. As time goes by, changes in the joint, such as the bone marrow lesions, as well as synovial changes, capsule thickening, and meniscus maceration and extrusion can occur. These and other symptoms progress slowly over the years (Lohmander, Englund et al. 2007). Based on the short and long-term economic costs of ACL injuries, identifying factors that contribute to high ACL loads would be beneficial for improving ACL injury prevention programs and ultimately minimize the costs associated with ACL injury.

Contributions of the Quadriceps and Hamstrings on Knee Loads

Although some mechanisms of ACL injury are unclear, quadriceps and hamstring muscle function has been shown to affect loads imparted on the ACL. The quadriceps is known to cause an anterior shear force on the knee joint which causes the ACL to strain. The hamstrings, however, are the agonist to the ACL. They counteract the force produced by the quadriceps, producing a posterior shear force on the proximal tibia and unload the ACL. When the quadriceps produce a greater shear force versus the posterior shear created by the hamstrings, the anterior cruciate ligament is strained and can rupture the ACL (Boden, Dean et al. 2000). To get a better understanding of what causes the one group of muscle to have the ligament strained, while the other reduces strain, we have to look at them individually, as well as their combined forces.

The antagonists of the anterior cruciate ligament are the quadriceps muscles. The highest strain that was achieved by the ACL by the this muscle group is shown to be produced when the knee is flexed from 0° to 40° (Beynnon and Fleming 1998). During isometric knee extension exercises, the isolated isometric contraction of the quadriceps produces the highest *in* *vivo* ACL strains between 15° and 30° of knee flexion (Beynnon, Fleming et al. 1995; Beynnon and Fleming 1998) .

Results of *in vivo* studies investigating the isometric thigh muscles on ACL force, the ACL is strained significantly by the quadriceps during knee flexion angles of 0° to 45° relative to the passive normal strain (Renstrom, Arms et al. 1986). In addition, with no tibial force applied to the leg, the quadriceps caused the ACL force to increase significantly with knee flexion angles of between 0° to around 70° of knee flexion (Durselen, Claes et al. 1995; Markolf, O'Neill et al. 2004). *In situ* forces in the ACL were found to be highest during 15° of knee flexion during isolated quadriceps forces and then the ACL force decreases past 60° of knee flexion (Li, Rudy et al. 1999). All of these results substantiate the findings by Markolf, Gorek et al. (1990), saying that the *in vitro* contractions of the quadriceps produce high ACL strains or forces in early knee flexion.

Similar results were found when researchers used musculoskeletal modeling to determine muscle forces on the ACL. During the simulated knee extension exercise, the ACL was loaded throughout the range of motion, 0° to 80° of knee flexion, with the ACL forces increasing as the quadriceps forces increase (Shelburne and Pandy 1997). Other musculoskeletal modeling studies have shown the ACL being loaded from full extension to 10° of knee flexion (Shelburne and Pandy 1998). This corroborates the data from the previous literature from *in vivo* and *in vitro* studies. During a more dynamic task of drop-landings, the musculoskeletal modeling resulted in the ACL being loaded during the first 25% of the landing phase, 33° to 48° of knee flexion (Pflum, Shelburne et al. 2004). This also is the knee angle at which the quadriceps forces, according to the model simulation, are at their peak.

The hamstrings, on the other hand, cause a posterior shear force on the knee which reduces stress or force on the ACL, acting as the agonist. The hamstring muscles prevent the tibia from moving anteriorly relative to the distal femur thus decreasing strain of the anterior cruciate ligament (Baratta, Solomonow et al. 1988). The question one has to ask is "in what position of knee flexion does the knee have to be in order for the hamstrings to be effective in reducing ACL loads?" *In vivo* studies by Beynnon (1995), which inserted a Hall effect transducer into subjects ACLs, showed that the reduced strain of the ACL by the hamstring is independent of the knee flexion angle. This was done during isometric knee extension where the resistance was applied by an anterior-posterior shear load on the tibia. The ACL strain remained low or unstrained throughout the range of motion that were tested (15°, 30°, 60°, and 90°, of knee flexion).

The results were similar with *in vitro* studies. The ACL strain was decreased relative to the ACL's passive normal strain throughout the range of motion due to the isometric hamstring activity. From full extension, 0°, to approximately 60° or 70° of knee flexion, the strain reduction due to the hamstring does not change significantly even though there is less ACL strain relative to the passive normal strain. Beyond 70° of knee flexion, however, isometric hamstring activity significantly decreases the ACL strain (Renstrom, Arms et al. 1986; Durselen, Claes et al. 1995). A study by Markolf, O'Neill et al. (2004) showed that the hamstrings not only decrease ACL strain past 70° of knee flexion but the strain significantly decreased much earlier starting at 10° of knee flexion, which is consistent with *in-vivo* research by (Beynnon, Fleming et al. 1995)**.**

During a knee extension simulation via musculoskeletal modeling, the ACL is only loaded during the 0 ° to 10° of knee flexion with the ACL load decreasing past 10° of knee flexion. This is agreeing with Markolf & O'Neill (2004) about the strain significantly decreasing past 10° of knee flexion. This is also found in Shelburne and Pandy (1998), where near full extension, 0° to 10°, the hamstrings have a small posterior shear component but at knee flexion angles of greater than 10°, the hamstrings applied a large posterior shear force on the tibia. This posterior force, as we know, significantly decreases the ACL strain. During a more dynamic task, such as drop-landings, the hamstrings were shown to also provide a posterior shear force to the tibia throughout the landing phase. This posterior force that is applied to the lower leg, significantly increased over time, causing the ACL to be loaded in the first 25% of the landing phase, when the knee is flexed from 33° to 48°, but then decreasing immediately following the first 25% of the landing phase (Pflum, Shelburne et al. 2004). This is due to the hamstrings counteracting the anterior shear force caused by the patellar tendon or the quadriceps activating.

While we know the isometric values the quadriceps and hamstrings have on the anterior cruciate ligament strain acting alone, our muscles move in concert with each other to provide movement, not just our quadriceps or hamstrings activating one at a time. The quadriceps and hamstrings work together to provide movement but an imbalance of the two muscles forces can produce significant knee loads and as a result, potentially injure the ACL.

In vivo studies investigating ACL strain indicate that the highest strains developed between 15° and 30° of knee flexion (Beynnon, Fleming et al. 1995). But when the quadriceps and hamstrings were co-contracted, the strain was significantly lower at these knee flexion angles compared to an isolated quadriceps contraction. The *in vitro* results were similar to the *in vivo*. The ACL strain when the quadriceps and hamstrings contract together, was significantly higher and at maximum strain during 0° to 15° of knee flexion during *in vivo* (Renstrom, Arms et al. 1986). But after 30° of knee flexion the combined muscles, hamstrings co-contraction significantly reduced the ACL strain. *In situ* forces of the ACL with combined quadriceps and hamstrings loads showed that the forces were greatest from full extension, 0° to 30° of knee flexion (Li, Rudy et al. 1999; Li, Zayontz et al. 2004). These forces, just like the *in vitro* results, decreased as the knee flexion increased and were significantly lower than the isolated quadriceps forces.

Musculoskeletal modeling also agrees with these results. As the hamstrings contract with the quadriceps, ACL forces are decreased compared to isolated quadriceps contractions during a simulated knee extension exercise (Shelburne and Pandy 1998). The ACL was only loaded during the first 30° of knee flexion in the musculoskeletal model. The reason for this is that the hamstrings produced a posterior shear force which decreases the net anterior force that is applied to the leg (Pandy and Shelburne 1997).

Biomechanics of Landing Height and the Effects of Landing Height

The thigh muscles have an effect on the knee loads and they can result in the ACL being strained however, a deeper look into this phenomenon and how it can be applied to practical situations such as landing from various heights must be explored. Many studies have found that the vertical ground reaction forces (GRF) increase as the landing heights increase (Bobbert, Huijing et al. 1987; Zhang, Bates et al. 2000; Seegmiller and McCaw 2003; Bisseling, Hof et al. 2007; Yeow, Lee et al. 2009). During single-leg landing, Yeow, Lee et al. (2010) saw that as the height increased from 30cm to 60m, the ground reaction force significantly increased from the lower height. With this increase in landing height, the knee becomes more flexed as well. This study also found that when the landing height increased from 30cm to 60cm, the knee flexion angle increased and this was for initial contact, and the peak ground reaction force.

Similar results were found by Bisseling, Hof et al. (2007) and Zhang, Bates et al. (2000) where the knee flexion increased as the landing height increased from 46°, 48°, and 53° for 30cm, 50cm and 70cm respectively, and 52°, 56°, and 63° for 32cm, 62cm and 103cm in height respectively. This knee flexion increase could be a common strategy to attenuate the ground reaction forces upon impact.

This landing strategy was investigated by DeVita and Skelly (1992), looking at landing stiffness. With a stiffer landing the final knee flexion position was less than 90° of knee flexion, while the softer landing had a final knee flexion angle of greater than 90°. The lesser knee flexion, stiff landing, created a greater ground reaction force than the softer landing with the more knee flexion. The knee flexor moments were also greater during the stiff landing when compared to the soft landing. These results are comparable to what Zhang, Bates et al. (2000) found when the landing heights increased. This increase in landing height caused the knee to become more flexed when landing, and subsequently the knee joint extensor moments increased. This result has been shown to prove previous studied correct (Bobbert, Huijing et al. 1987; McNitt-Gray 1993). With increases in knee extensor moments, the knee joint powers also increased as the landing heights increased indicating that the quadriceps muscles were the primary energy absorbers during landing.

With this data on how landing heights affect the knee biomechanics, there is very little research on ACL forces in landing from different heights. However, due to the increased knee extensor moments accompanying increased landing heights, it is suggested that hamstring/quadriceps force ratios decrease as landing height increases (primarily due to the increased quadriceps demands) resulting in elevated forces on the ACL.

Summary

A tear in the anterior cruciate ligament is one of the most common injuries in athletes, both professional and recreational. The injury usually occurs when the person lands or cuts. Increased forces in the quadriceps and not enough force by the hamstrings can cause anterior tibial translation in the knee and thus causing the ACL to tear. The anterior cruciate ligament is susceptible to undergo dangerously high forces during landing because the quadriceps muscle forces vs. the hamstring muscle forces are much higher in order to attenuate the ground reaction forces upon impact with the ground. This imbalance in quadriceps vs. hamstring muscle forces presumably is more exaggerated when landing from increased heights thereby placing potentially injurious forces on the ACL at these higher heights. This thesis will test the hypothesis that hamstring to quadriceps force ratio would be inversely related to peak ACL forces during landing from three different heights.

CHAPTER 3: METHODOLOGY

Design

This study is aimed to investigate the influence of the quadriceps and hamstrings forces on ACL forces when landing from different heights. This study followed a within subject design where subjects completed 8 single-leg landing trials each from heights of 15cm, 30cm, and 45cm. Since landing is a common mechanism for ACL injury, evaluation of how individuals adapt to increased landing heights through thigh muscle co-contraction is essential.

Subjects

The study involves young adults, who are between the ages 18 and 30 years. There were three male and three female participants. The males had an average height of 1.75±0.07m with an average mass of 74.08±8.66kg and the females had an average height of 1.70±0.04m and an average mass of 55.93±6.83kg. The individuals were in relatively good health and were recreationally active. All of the subjects had a background of participating in activities that involve jumping. The subjects also did not have any injuries or previous surgeries to their lower back, hip, knees, ankles, or any other lower limb injuries. They cannot be involved currently in any intercollegiate sports. All subjects will read and sign an informed consent form approved by the UMCIRB prior to participation.

Instrumentation

Kinematic data were obtained at 120 Hz using a 5-camera Motion Analysis/Falcon Analog System (Motion Analysis Corporation, Santa Rosa, CA) and the ground reaction forces were acquired using a Bertec force plate (Bertec Corporation, Columbus, OH).

Procedures

The subject's age, height, weight, sex and leg dominance were recorded. The subject's pelvic inclination, pelvic depth, and trunk depth were collected using a PALM skeletal alignment and leg-length discrepancy instrument (Palpation Meter; Performance Attainment Associates, Lindstrom, MN). These measurements were used to construct the biomechanical model. All subjects wore a t-shirt, compression or spandex shorts, and their own athletic shoes. Prior to subject set-up, to make sure that the participate is familiar with the task of landing, we asked them to complete the landing task from heights of 15cm, 30cm, and 45 cm. Next, all subjects were instrumented for biomechanical analysis. Electrode pairs were placed in the following order: 1) vastus medialis (VMO), 2) vastus lateralis (VL), 3) rectus femoris (RF), 4) lateral hamstring (LHAM), 5) medial hamstring (MHAM), 6) lateral gastrocnemius (LGAS), 7) medial gastrocnemius (MGAS).

Cross-talk was checked manually by observing the EMG signal with a real-time scope during various contractions. Three maximum voluntary isometric contractions (MVIC) were taken for the quadriceps at 90°, the hamstrings at 45° of knee flexion and for the gastrocnemius. Reflective markers were placed on the first metatarsal head, heal, medial and lateral metatarsal heads, medial and lateral midfoot, and the medial and lateral ankle. This was done for both the left and right foot. Markers were placed on the left and right shank; a triad on the lateral side of the tibia and fibula, medial and lateral knee joint. Markers were placed on the left and right thigh using a triad. Reflective markers were also placed on the pelvis; the right and left trochanters; the right and left anterior superior iliac spine; the right and left iliac crest; and the sacrum. On the trunk, the markers were placed posteriorly on a fitted vest where three are located; on the bilateral acromions and the sternum. Markers were placed on the forehead, left and right lateral elbows, and left and right lateral wrists for calibration purposes only. Two calibration trials were done on the subject with all 48 markers in scarecrow stance and two trials with the arms across the chest that had 44 markers. The medial markers on the foot, medial markers on the both knees as well as the iliac crest markers were removed along with the forehead and arm markers, leaving 33 markers remaining. The dynamic marker set was loaded in Eva Real-time software (EVaRT; version 4.6; Motion Analysis Santa Rosa, CA) in the computer and one trial of landing was collected to make sure that all 33 markers are located to make a template. All subjects performed a set of 8 landing trials with four conditions; landing from heights of 1) 15cm, 2) 30cm, 3) 45cm heights and back to the 4) 15cm height. The order of the 30cm and 45cm landing height were counterbalanced across subjects. The second 15cm landing condition and the counterbalancing of the two highest heights were to try and negate any affect that fatigue may have. The subjects landed on their right leg with their arms folded across their chest. There was no set standard for landing style, only that the subject land "as naturally as possible".

Data Reduction

During the trials, the subject's data was tracked and collected in EVaRT 4.6. Eva Real-Time software (EVaRT) is used to set up, calibrate, capture motion in real-time, and capture motion for post processing. The trials were initially analyzed using Visual 3D software to determine unacceptable trials due to obvious marker tracking errors. It was then imported into OpenSim and the data was reduced. OpenSim is a freely available software package that enables you to build, exchange, and analyze computer models of the musculoskeletal system and dynamic simulations of movement. From the anthropometric measurements we took, height, weight, pelvic depth, etc., we scaled a model that was already in OpenSim. The model that was used is Gate 2392. From OpenSim, we obtained the inverse kinematics and forward dynamics and compared them to the experimental results. From a series of mathematical algorithms, we also obtained the muscle forces and muscle excitations. These muscles forces were required to produce human motion.

Muscle-Actuated Simulations of Landing

Muscle-actuated forward dynamics simulations of all landing trials will be conducted using OpenSim, a freely available software platform. OpenSim is an open-source platform for modeling, simulating, and analyzing the neuromusculoskeletal system. There is a five step process to create a muscle-driven simulation of a movement (Delp, Anderson et al. 2007) (Figure 1).

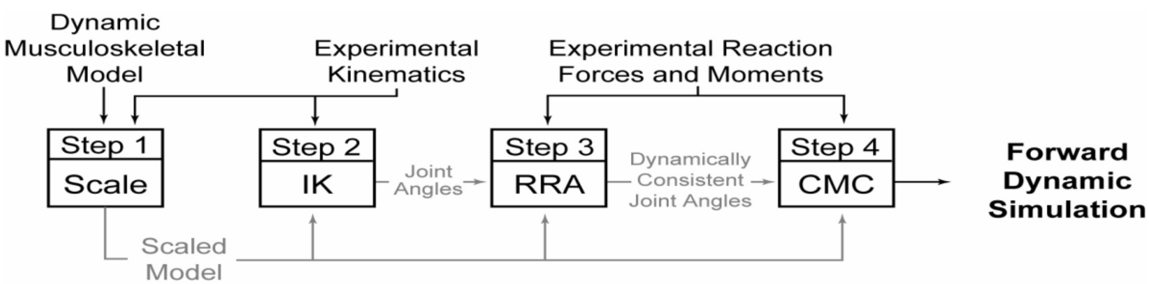

Figure 1. These are the steps for generating a muscle-driven simulation. Step 1 is to scale the musculoskeletal model; Step 2 is the inverse dynamics calculations; Step 3 is the residual reduction algorithm (RRA); Step 4 is computed muscle control (CMC); Step 5 is the forward dynamic simulation (Delp, Anderson et al. 2007)

Step 1: Scaling a Musculoskeletal Model

The generic musculoskeletal model, Gate 2392, has a height of 1.8 meters and a mass of approximately 72.6 kilograms. The model consists of 13 rigid body segments, 28 degrees of freedom and includes 86 muscles, with 43 per leg. A Hill-type model (Figure 2), with three components (Contractile Element, Series Elastic Element, and Parallel Elastic Element), was used to characterize musculotendon contraction dynamics.

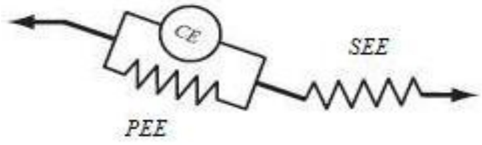

Figure 2. A Hill-type model is used to calculate musculotendon contraction dynamics, where CE is the contractile element; SEE is the series elastic element; and PEE is the parallel elastic element.

The musculoskeletal model was scaled to give us an accurate match to the actual subject that was being tested. This scaling was done by modifying the height, weight, and the size of the trunk and pelvis of the model so that it matches the anthropometry of the subject. The body segments of the model were scaled by calculating the average distance between specific marker pairs. The ratio between the averaged distances to the distance in the model was then used to scale each specific segment in the specific axis directions. Maximum peak isometric forces for all muscles in the model were scaled depending on the ratio of the musculotendon length of the unscaled model to the length of the generic model.

Step 2: Inverse Kinematics

Inverse kinematics is a mathematical problem that determines the joint angles of the model to best reproduce the experimental kinematics of the subject. It is formulated as a least-squares problem that minimizes the differences between the measured marker locations and the model's virtual marker locations. For each frame in the experimental kinematics, the inverse kinematics problem is to minimize the weighted squared error:

$$Squared\ Error = \sum_{i=1}^{\text{markers}} w_i(\bar{x}_i^{subject} - \bar{x}_i^{model})^2 + \sum_{j=1}^{\text{joint angles}} \omega_j(\theta_j^{subject} - \theta_j^{model})^2 \quad (1)$$

The weights (w) are operator determined so that the least squares algorithm will weigh more heavily markers and/or joint angles which you have more confidence in i.e. knee flexion angle versus kinematics that inherently have more error i.e. ankle inversion or hip rotation.

Step 3: Residual Reduction Algorithm (RRA)

In order for the muscles in the model to actually create the experimental motion, sources of error between the experimental kinematics and kinetics need to first be minimized. Newton's second law states:

$$\vec{F} = m \cdot \vec{a} \tag{2}$$

Errors in experimental data and the model itself result in a dynamic inconsistency where Newton's second law is violated. Sources of this error could be that the musculoskeletal model does not incorporate all of the model's segments and their masses or splitting by the muscle into separate segments. There is also, marker error, which is the distance between the experimental marker and the corresponding marker on the OpenSim model and coordinate error, which is the difference between the experimental coordinate value and the coordinate value that is computed during inverse kinematics step in OpenSim. The residual reduction algorithm (RRA) produces a model with the inverse kinematic that were previously computed match the measured ground reaction forces and torques. With the addition of an error term in Newton's second law, it allows us to have dynamic consistency between the forces and motion throughout the simulated task.

$$\vec{F}_{exp} + \vec{F}_{residual} = m \cdot \vec{a} \tag{3}$$

The RRA reduces the residual errors as much as possible primarily by making small adjustments in the accelerations of the segments in the model while tracking the inverse kinematics solution as much as possible. An example of the initial residual errors and final residual errors is presented in the following figure (Figure 3):

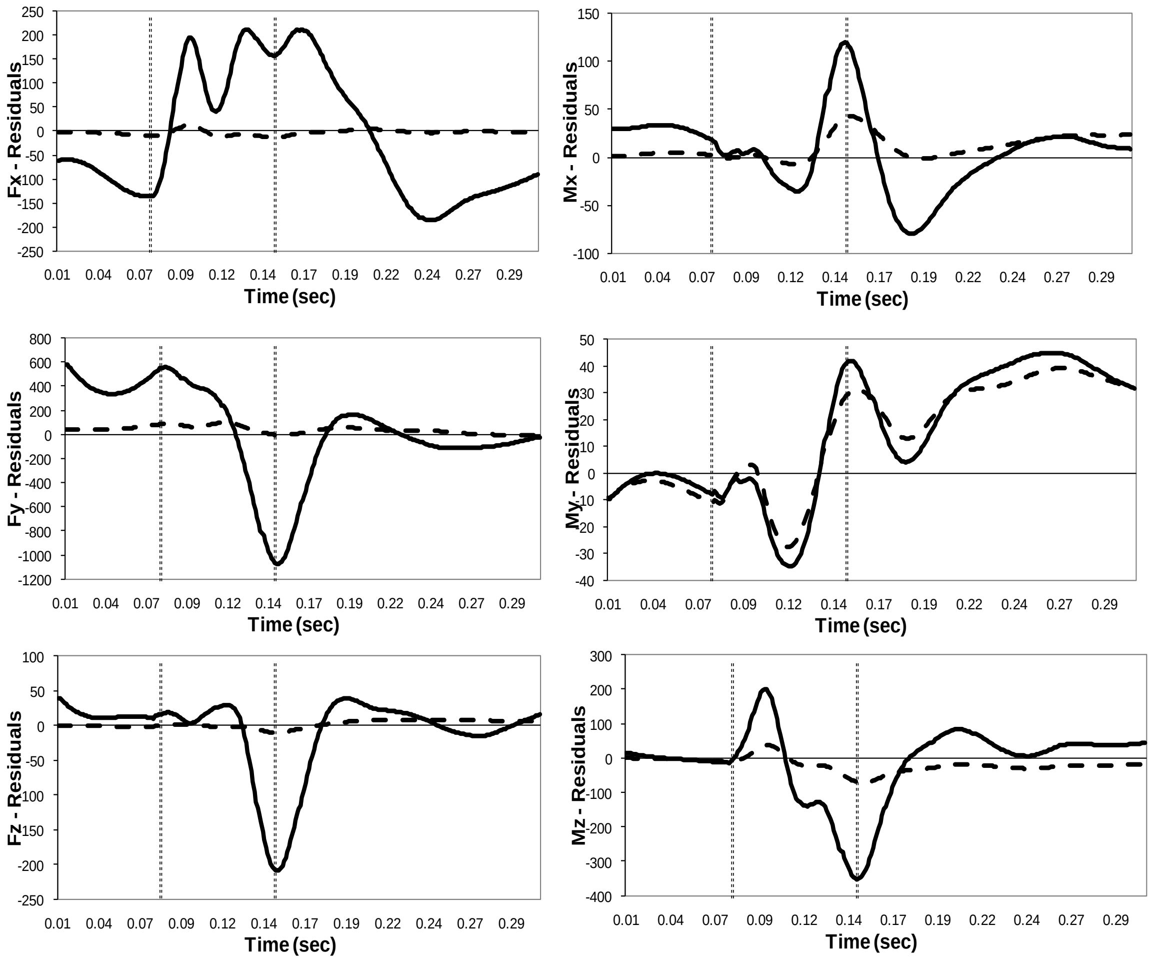

Figure 3. An example of how the RRA reduces the error in the ground reaction forces and moments is shown. The dashed line is after the RRA is completed i.e. residuals errors are reduced. The solid line is without the reduction in residuals error. First vertical dotted line reflects ground contact, and the second vertical dotted line is time of peak vertical ground reaction force.

Step 4: Computed Muscle Control (CMC)

The kinematic solution with reduced residual error determined by the residual reduction algorithm is then input in to the next step, computed muscle control (CMC). It generates a set of muscle excitations that will eventually be used to produce a coordinated muscle-driven simulation of the subject's movement. The proportional-derivative (PD) controller attempts to match the simulated vs. experimental kinematics with user specified joint velocity error and joint position error feedback gains. The simulated results are fed into a static optimization routine that computes a set of desired muscle forces that would then produce the simulated accelerations in the configuration, and also minimized a cost function to resolve muscle redundancy (Thelen and Anderson 2006). This cost function minimizes the sum of muscle activations squared. The static optimization results allow us to get the muscle activations that create the muscle forces. OpenSim uses static optimization to distribute forces across synergetic muscles the PD control to generate forward dynamics motion makes closely tracks the kinematics from the RRA. It is important to note that this method of computing muscle activations is mathematical and experimental muscle activation, EMG, is not used during this process. The muscle forces then produce the necessary accelerations via a forward dynamics analysis (Thelen, Anderson et al. 2003).

Step 5: Forward Dynamics

The final step of generating a muscle-driven simulation is forward dynamics. The muscles activations that were derived from the CMC are entered into the forward dynamics model. The rate of change of muscle activation, the change in muscle length, and the accelerations of the coordinates in response to the muscle forces are computed in forward dynamics. The equations, the muscle control algorithm, are then reran or fed back through the CMC and fed forward to completion again. The resulting activations and accelerations are compared to the experimental data. If they are not similar then the muscle control algorithm is rerun until the experimental and simulated results are comparable.

ACL Force Computations

In order to compute the forces on the ACL, knee shear forces will be computed followed by adjustment of the shear forces to the orientation of the ACL. First, the orientations of the patellar tendon and each of the four hamstring muscles relative to their insertion sites on the tibia will be taken from the musculoskeletal model and expressed as 3rd order polynomial functions to match the changing orientations as a function of knee flexion angle. The gastrocnemius angle, than angle between the gastrocnemius and the tibia, was held constant to approximately 3 degrees (DeVita and Hortobagyi 2001). Knee shear forces will then be calculated by adding each of the muscle and joint reaction force components acting parallel to the surface of the tibia (DeVita and Hortobagyi 2001; McLean, Huang et al. 2004; Winter 2009). Once the shear forces are solved, these forces will be adjusted to match the orientation of the ACL relative to the surface of the tibia. The changing orientation of the ACL relative to the tibia slope is expressed as a 3^{rd} order polynomial function (r^2 value of 0.99) and derived from *in vivo* studies measuring ACL kinematics during a forward lunge (Jordan, DeFrate et al. 2007).

Validation of peak ACL Forces and Muscle Force Estimation

Rupture to the anterior cruciate ligament has been found to occur within the first 75ms after initial ground contact (Krosshaug, Nakamae et al. 2007). None of the subjects that were tested had peak ACL forces that were within this range. Our results for the pattern of ACL force were similar to that of Cerulli, Benoit et al. (2003) *in vivo* study, and Pflum, Shelburne et al. (2004) musculoskeletal modeling study, where there was an increase in ACL force and then a slight decrease and an increase again in the ACL force (Figure 4). Cerulli, however measured the ACL strain and not the force but it has been seen that the strain is correlated to the force (Woo, Debski et al. 1999) and in that study, the ACL strain did not decrease but remained constant. This could be due to the task, a standing hop, which was performed.

It has been found that the magnitude for an ACL to rupture is ~ 2200N (Woo, Hollis et al. 1991). Fortunately, none of our subjects reached this threshold, with the highest ACL force magnitude measured ~1400N.

Figure 4. ACL Forces and Ground Reaction Force Curves from Three Different Landing Heights

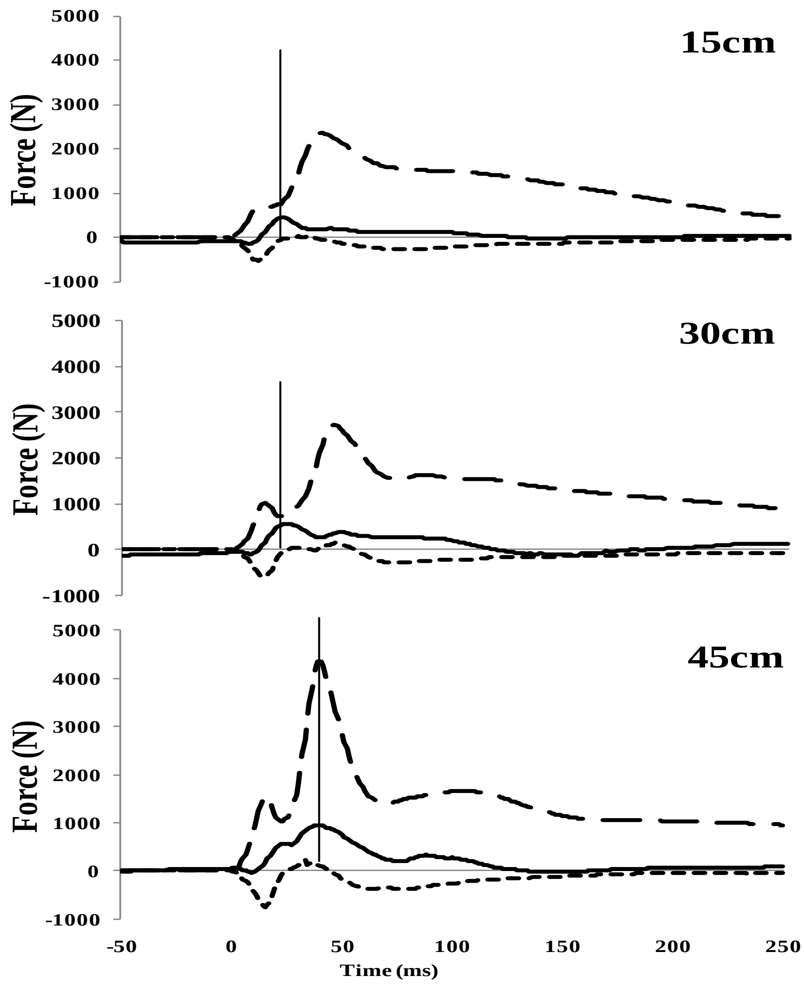

Figure 4. ACL and ground reaction force curves from three different heights. As landing height increased so did the ACL and ground reaction forces. The line represents the peak ACL force at that height. —— ACL Forces, — — — Vertical Ground Reaction Forces, - - - - Anterior-Posterior Ground Reaction Forces

Table 1. ACL, Quadriceps, and Hamstring Forces in Various Task

Literature	Task	ACL	Quadriceps	Hamstring
DeVita and Hortobágyi	Walking	—	2.70	1.30
Shelburne, Torry et al.	Walking	0.44	1.73	0.72
Besier, Fredericson et al.	Walking	—	2.23	1.02
Kulas, Hortobágyi et. al.	Double-leg Landing	—	7.80	3.50
Pflum, Shelburne et al.	Double-leg Landing	0.40	6.40	1.00
Our study	Single-Leg Landing	1.015	6.63	1.10

Table 1. All data are in units of force and reported in body weights.

Data Analysis

Our overall hypothesis stated that the hamstring to quadriceps ratio would be inversely related to peak ACL forces during landing with the primary contributions of this muscle force ratio coming from the quadriceps demands inherent in landing activities. We used bivariate correlations to describe the linear relationships between quadriceps force and peak ACL force, hamstrings force and ACL force, and the hamstring to quadriceps ratio and peak ACL force. The quadriceps force, hamstrings force, and the hamstring to quadriceps ratio were taken at the time of peak ACL force (Figure 4). Based on previous laboratory work with squatting, the hamstring/quadriceps ratio explained 72% of the variance in ACL forces. However, this relationship was nonlinear indicating that there is an optimal magnitude of hamstring/quadriceps force ratio which minimizes ACL forces. Based on this squatting work, curve analyses describing the non-linear relationship between the hamstring/quadriceps ratio and peak ACL force was conducted at each landing height and pooled across heights to best describe this actual relationship. Alpha levels of .05 were used for all analyses.

CHAPTER 4: RESULTS

It was hypothesized that the hamstring to quadriceps force ratio would be negatively related to peak ACL forces when landing from the three heights. A regression analysis was conducted to test this hypothesis by predicting ACL forces from the hamstring to quadriceps ratio at each landing height and pooled across heights. A curve analysis was also used to determine the hamstring to quadriceps force and the peak ACL force to best describe the actual relationship. This section is organized into the following manner: 1) analysis of the two 15cm conditions to test for a potential fatigue effect of the protocol, 2) correlation analyses of individual muscle forces and combined as the hamstring to quadriceps force ratio, 3) curve analysis with hamstring to quadriceps force ratio predicting peak ACL force, and 4) a supplemental analysis (stepwise linear regression) which utilized the hamstring to quadriceps force ratio and the vertical ground reaction forces as predictors of peak ACL force (pooled across height, and at each landing height). These last analyses were added because although the hamstring and quadriceps forces have been shown to affect ACL forces, ground reaction forces also increase with landing height and may also play a critical role in determining the total force on the ACL.

Effects of the Order Effect of the Protocol

Even though the protocol called for the order of landing heights to be counter balanced for the highest heights, 30cm and 45cm, for each subject, a fatigue effect might still occur. For this we compared the lowest heights since they were the first and last landing heights to be performed. Paired sample t-tests were performed and the peak ACL force differed by 7.1% between conditions 1 and 4 and this difference was not significant ($p<0.05$) (Figure 5). The thigh muscle force, quadriceps and hamstrings also were not seen as being significantly different between conditions, as they differed 2.5% and 8.9%, respectively ($p<0.05$). The ground reaction forces differed by 29.4% and it too was not significant ($p<0.05$), as well as the ground reaction forces at the time of peak ACL force, which differed 8.9% between conditions ($p<0.05$).

Figure 5. Comparison of Conditions 1 and 4, 15cm

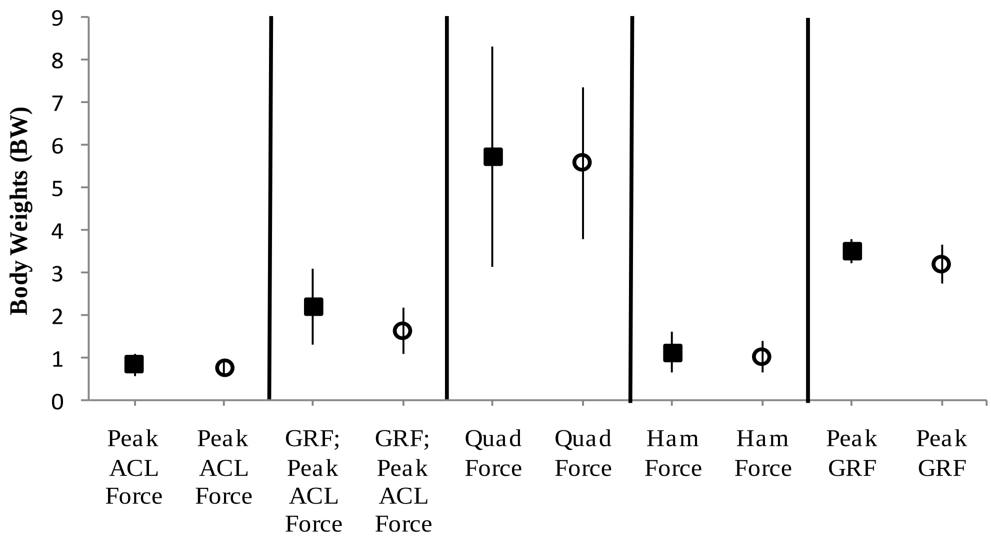

Figure 5**. Comparison of both 15cm conditions (first and last condition). There were no significant differences between conditions. ■ – Condition 1, ○ – Condition 4. The bars represent the 95% confidence interval.**

Evaluation of Landing Height on Muscle, Ligament, and Ground Reaction Forces

As landing height increased the quadriceps muscle forces, ACL forces, and the ground reaction forces increased. The quadriceps muscle forced increased by 40.0% from 15cm to 45cm, the ACL forces increased 56.2%, and the ground reaction forces increased 73.8% from the lowest height to the highest (Table 2). The hamstrings muscle forces remained relatively constant increasing 16.3%. With the hamstrings remaining constant and the quadriceps increasing, the hamstring to quadriceps ratio decreased by 46.8% from 15cm to 45cm.

Table 2. Effects of Landing Height on Thigh Muscle Force, Hamstring to Quadriceps Ratio, ACL Forces and Ground Reaction Forces

	Hamstring Force	Quadriceps Force	H/Q Ratio	ACL Force	GRF
15cm	1.082	5.655	0.253	0.813	1.909
30cm	0.974	6.735	0.177	0.984	2.586
45cm	1.274	8.484	0.157	1.449	4.143

Table 2. Mean values of the hamstring muscle forces, quadriceps muscle forces, the hamstring to quadriceps ratio, the ACL forces and the ground reaction forces at each landing height. As landing height increases, all of the variables increase, except for the hamstrings and hamstring to quadriceps ratio. All values are reported in Body Weights.

Effects of Quadriceps, Hamstrings, and Hamstring to Quadriceps ratio on Peak ACL Force

Quadriceps muscle forces at peak ACL force were strongly linearly correlated with peak ACL forces R = 0.781, $r^2 = 0.609$ (p<0.01) (Figure 6A). The hamstrings muscle force however, did not correlate with peak ACL forces (R = -0.071, $r^2 = 0.005$; $p = 0.46$, Figure 6B). The hamstring to quadriceps ratio was also negatively correlated with R = 528, $r^2 = 0.278$ (p<0.01) (Figure 6C).

Figure 6. Linear Correlations of Quadriceps, Hamstrings, and Hamstring/Quadriceps Ratio to Peak ACL Force

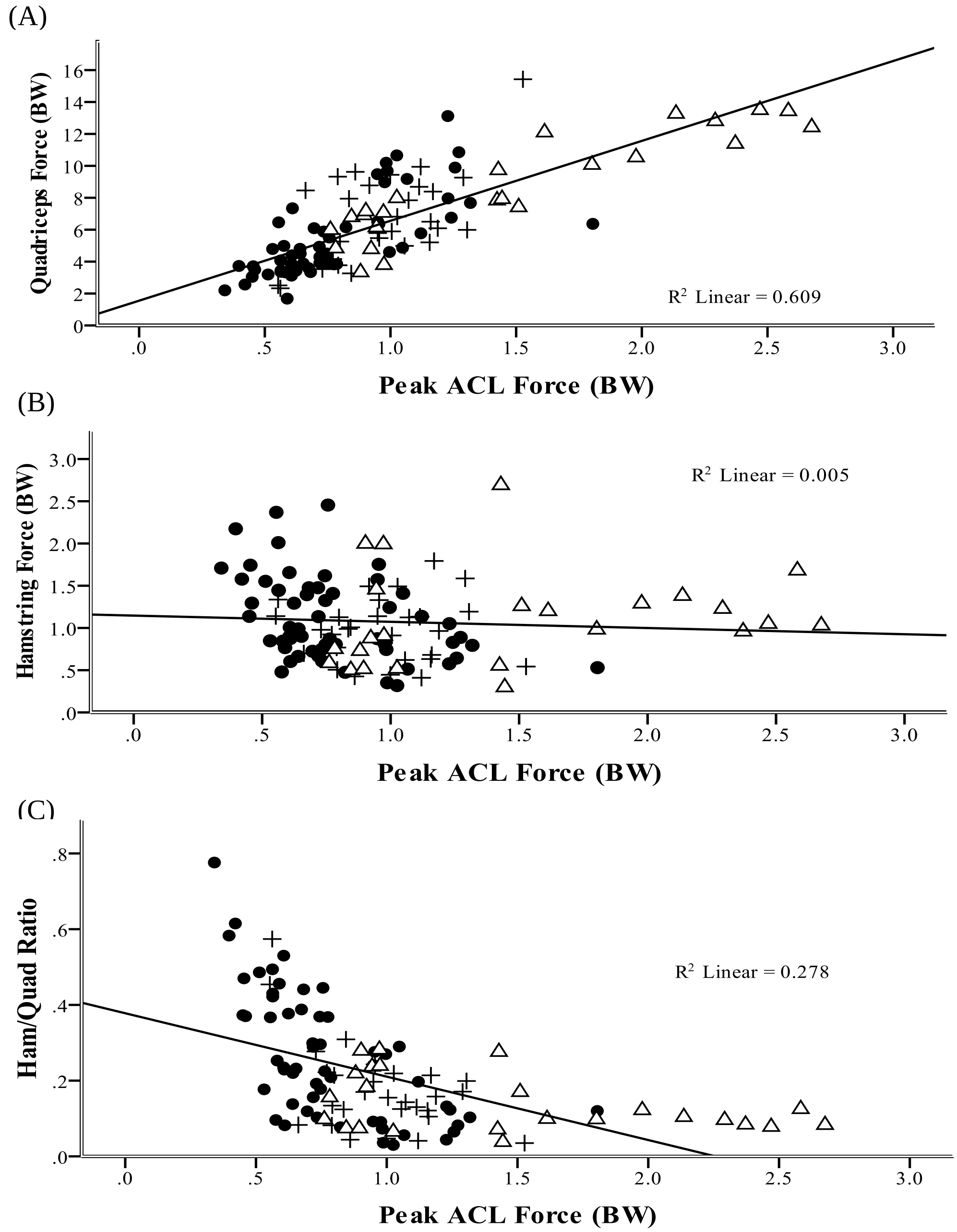

Figure 6. Linear regression correlation of the quadriceps (A), hamstrings (B) and the hamstring to quadriceps ratio (C) to the peak ACL force. The quadriceps muscles were seen to be highly correlated, while the hamstrings were not. ● - 15cm, + - 30cm, Δ - 45cm.

Nonlinear Analysis of Hamstring to Quadriceps Ratio Relationship with Peak ACL Forces

A nonlinear analysis of the hamstring to quadriceps ratio relationship showed the correlation was 0.652 with a total explained variance of $r^2 = 0.425$ ($p<0.01$) (Figure 7).

Figure 7. Nonlinear Relationship of Hamstring/Quadriceps Ratio to Peak ACL Force Pooled Across Heights

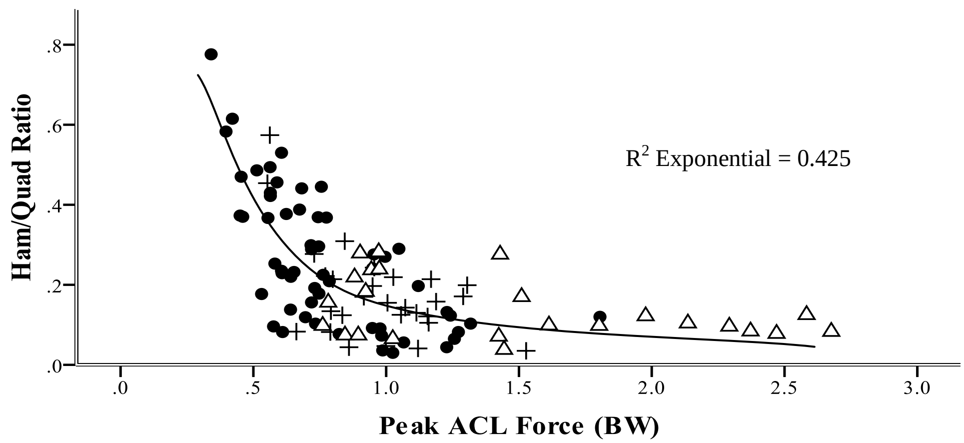

Figure 7. Nonlinear correlation relationship of the hamstring to quadriceps ratio to the peak ACL force pooled across heights. The Nonlinear (Inverse) correlation is a better fit and predictor than the linear correlation. ● - 15cm, + - 30cm, Δ - 45cm

A nonlinear correlation analysis of the hamstring to quadriceps ratio is a better predictor of the peak ACL forces than linear correlation. However, the correlation of all of the analyses showed that the strongest correlation is at the lowest heights regardless of whether the relationship was described as linear or non-linear. Regardless of analysis, the strength of the correlation decreased as height increased (Table 3).

Table 3. Total Explained Variance, R^2 for H/Q Ratio and Peak ACL Force

Landing Height	Linear	Inverse	Exponential	Power	Logarithmic
15cm	0.410	0.263	0.505	0.452	0.400
30cm	0.279	0.138*	0.343	0.213	0.198
45cm	0.172	0.053*	0.166	0.125*	0.123*
Pooled Heights	0.278	0.133	0.425	0.351	0.250

Table 3. Various correlations for the hamstring to quadicep ratio and peak ACL force, by landing height and all of the heights pooled together, where the H/Q ratio is the independent variable. The correlation for each height decreases regardless of analysis. * Regressions not signficant (*p*>0.05). All others are significant.

Combined Effects of Ground Reaction Forces & H/Q Ratio on Peak ACL Force

As supplemental analyses, we analyzed, through a stepwise linear regression, the combined influence of the hamstring to quadriceps ratio and ground reaction forces on peak ACL forces. This was based on previous research showing the ground reaction forces do factor into the loading of the ACL (Pflum, Shelburne et al. 2004). Therefore, the ground reaction forces' influence on the ACL with and without the combined effect of the hamstring to quadriceps ratio was examined. There was a moderately strong correlation between ground reaction forces and the peak ACL force with R = 0.690, $r^2 = 0.477$ ($p<0.01$) (Figure 8).

Figure 8. Linear Correlation of Ground Reaction Forces and the Peak ACL Force

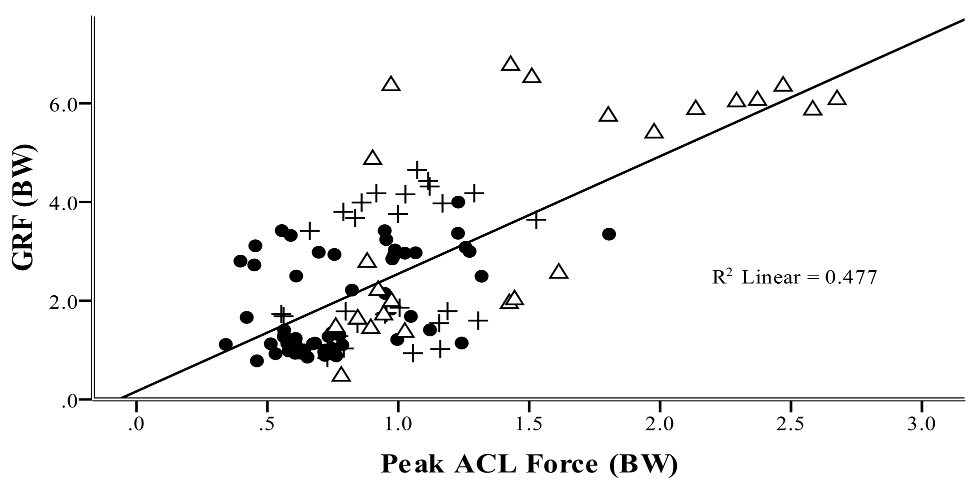

Figure 8. Linear correlation of ground reaction forces to the peak ACL force. The ground reaction forces are positively correlated with the ACL forces; R = 0.690, $r^2 = 0.477$ ($p<0.01$). ● - 15cm, + - 30cm, Δ - 45cm.

As explained in the previous section, as the landing height increases, the correlation of the hamstring to quadriceps ratio and peak ACL forces decreases. There must be another factor that is driving the correlation down. A stepwise regression was used to investigate the combined effects for the ground reaction forces and hamstring to quadriceps ratio on peak ACL forces. At the lowest height, 15cm, the hamstring to quadriceps ratio was inserted first, and had R = 0.640 and accounted for 41.0% of the variance ($r^2 = .410$; Table 4) in peak ACL forces. The ground reaction forces had an R-value of 0.706 and added 8.9% of the explained variance (r^2 total = 0.499). At 30cm, the ground reaction forces were the only component to be entered into the model (R = 0.528, $r^2 = 0.279$). However, at the highest height, 45cm, the ground reaction forces entered first in the model with R = 0.715 and explaining 51.1% ($r^2 = 0.511$) of the variance. The hamstring to quadriceps ratio came next with R = 0.894, and explained an additional 28.8% of the variance (r^2 total = 0.799). When the landing heights were pooled together, the ground reaction force entered the model first with R = 0.690 and explaining 47.7% of the variance ($r^2 = 0.477$) (Table 4). The hamstring to quadriceps ratio entered second explaining 10.9% of the variance (r^2 total = 0.586). All stepwise regression analyses were $p<0.05$.

Table 4. Stepwise Linear Regression for the Ground Reaction Force, H/Q Ratio and Peak ACL Force

Landing Height		R-value	R-squared	Std. Error
15cm	H/Q Ratio	0.640	0.410	0.218
	GRF	0.706	0.499	0.203
30cm	GRF	0.528	0.279	0.199
45cm	GRF	0.715	0.511	0.463
	H/Q Ratio	0.894	0.799	0.304
Pooled Heights	GRF	0.690	0.477	0.340
	H/Q Ratio	0.766	0.586	0.304

Table 4. A stepwise linear regression analysis for the ground reaction force, hamstring to quadriceps ratio and peak ACL force show that during the lowest height, 15cm, that the H/Q ratio played a bigger role than the GRF. At the highest height, 45cm, the GRF was the main contributor to the increase in ACL force.

The total regression model predicted peak ACL force with an R = 0.756, r^2 = 0.586 and is graphically illustrated in Figure 9.

Figure 9. Predicted Peak ACL Forces vs. Peak ACL forces.

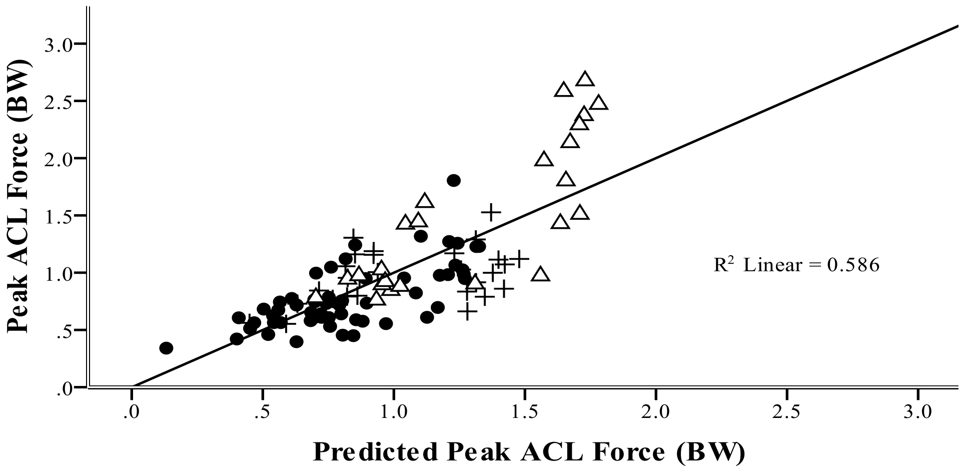

Figure 9. Predicted peak ACL forces with the addition of the GRF against the peak ACL forces with only the H/Q ratio. Based on the ground reaction forces playing a role in the peak ACL forces at the higher heights, the ACL forces were computed with the ground reaction forces and the hamstring to quadriceps ratio and correlated with the ACL forces with only the hamstring to quadriceps ratio. ● - 15cm, + - 30cm, ∆ - 45cm (Predicted ACL force = 0.794+GRF*0.169 – H/Q Ratio*1.096)

Summary of Results

The results of this study investigating the relationship of the thigh muscle forces and the peak ACL force, found that the quadriceps muscle forces were highly and signficantly correlated to the peak ACL force. The hamstring muscle forces, on the other hand, were not correlated to the peak ACL force. The combined forces of the hamstrings and quadriceps muscle were negatively correlated to the peak ACL force when a linear analysis was conducted. When performing a nonlinear correlation analysis, the correlation of the hamstring to quadriceps ratio became stronger. However, as the landing height increased, the correlation of the hamstrings to quadriceps ratio and peak ACL force decreased regardless of the correlation analysis that was completed. When the ground reaction forces were analyzed, it was found they were highly correlated to the peak ACL force. A stepwise regression analysis was performed to determine the variance of the peak ACL force is correlated with the hamstring to quadriceps ratio and the ground reaction force. At the lowest height, 15cm, the hamstring to quadriceps ratio entered the model first. As the landing heights increased, the ground reaction forces entered the model first and at the highest height, it explained 51% of the variance. When the landing heights were pooled together, the ground reaction force explained 47.7% of the variance while the hamstrings to quadriceps forces explained an additional 11%.

CHAPTER 5: DISCUSSION

This study was conducted to determine the relationship of the hamstring and quadriceps muscle forces on the anterior cruciate ligament. The hypothesis that led us to this purpose was that the hamstring to quadriceps ratio would be inversely correlated with the peak ACL forces. This hypothesis was derived from previous research showing increased quadriceps demands with increased landing height (Zhang et al 2000). However, as landing height increased, the effect of the ground reaction forces on peak ACL forces could not be ignored. This chapter will discuss the results related to the literature, overall hypothesis, and our supplemental analyses (including the effect of ground reaction forces on peak ACL force) and is organized in the following approach: 1) Development of Hypothesis, 2) Thigh Muscle forces, Hamstring/Quadriceps Ratio, and the peak ACL forces, 3) Clinical Manipulation of Knee Loads, 4) Summary, 5) Conclusion, and 6) Future Considerations.

Development of Hypothesis

There are over 200,000 occurrences of injury the anterior cruciate ligament every year (Albright, Carpenter et al. 1999). This accounts for billions of dollars every year spent to repair, reconstruct, and rehab the ACL. Injuries can occur via direct physical contact or non-contact activity. Non-contact can be constituted as running, cutting, or landing. Although there is no definitive cause for the ACL to be injured, factors have been identified and possibly within these factors, a better understanding of this injury can be known.

The data from this investigation show that the quadriceps muscle forces were strongly correlated to peak ACL forces regardless of landing height. During early knee flexion, the quadriceps muscles are seen as causing strain or exerting an anterior shear force that creates higher knee loads, and thus causes increased ACL loads. For this reason, the quadriceps is described as the antagonist of the ACL. This event has also been seen in both cadaver studies, *in vitro* (Markolf, Gorek et al. 1990; Durselen, Claes et al. 1995; Markolf, O'Neill et al. 2004), as well as in living subjects, *in vivo* (Renstrom, Arms et al. 1986; Beynnon and Fleming 1998; Li, Rudy et al. 1999; Li, Zayontz et al. 2004), where the ACL force is recorded by a force transducer. These studies show that the ACL is the most strained or force is applied on the ACL the most during the early knee flexion angles when the quadriceps are acting alone. Musculoskeletal modeling studies have been confirmed, as similar results have been recorded where no physical interaction with the knee occurs (Shelburne and Pandy 1997; Shelburne and Pandy 1998; Pflum, Shelburne et al. 2004).

The hamstrings have been shown to counter act the quadriceps muscle forces by providing a posterior shear force on the tibia relieving the stain or force on the ACL. *In vivo* research (Beynnon, Fleming et al. 1995) as well as *in vitro* research (Renstrom, Arms et al. 1986; Durselen, Claes et al. 1995; Markolf, O'Neill et al. 2004) have shown that the hamstrings reduce the ACL strain or force applied on the ACL throughout the range of motion of knee flexion. As with the quadriceps, the results from musculoskeletal modeling showed comparable findings (Shelburne and Pandy 1998; Pflum, Shelburne et al. 2004).

Previous research has shown that when the hamstrings and quadriceps are combined, the force applied to the ACL or the ACL strain is lower than when the quadriceps muscles are acting alone. However, the ACL is still strained and loaded during early knee flexion but still not as much with the quadriceps acting alone. Near mid knee flexion, the hamstrings provide a posterior shear force, and like with the hamstrings acting alone, the ACL force or strain begins to decrease (Beynnon and Fleming 1998). This combined quadriceps and hamstrings effect on the ACL has been found by means of *in vivo* research (Beynnon and Fleming 1998), *in vitro* research (Renstrom, Arms et al. 1986; Li, Rudy et al. 1999; Li, Zayontz et al. 2004) and musculoskeletal modeling (Shelburne and Pandy 1997; Shelburne and Pandy 1998).

Due to the ACL being injured during more dynamic tasks, it would be beneficial to examine other factors that may have an effect on the ACL. This study chose the dynamic task of landing. As the landing heights increases for a task, the ground reaction forces also increase. This has been found true for double-leg landing, as well as single-leg landing (Bobbert, Huijing et al. 1987; Zhang, Bates et al. 2000; Seegmiller and McCaw 2003; Bisseling, Hof et al. 2007; Yeow, Lee et al. 2010). These increased ground reaction forces can be reduced by changing knee flexion angle. It has been found that the more the knee is flexed, the lower ground reaction force is during landing (Devita and Skelly 1992). Also, as the landing heights increase, the knee flexion angle also increases (Zhang, Bates et al. 2000; Bisseling, Hof et al. 2007; Yeow, Lee et al. 2010).

With this increase in ground reaction force consistent with increased landing heights, the joint power and joint work increases (Zhang, Bates et al. 2000; Yeow, Lee et al. 2010) suggesting that the quadriceps muscles are the main absorbers during landing and this could be a strategy for landing, along with the change in knee flexion angle. This landing strategy led us to hypothesize that as landing height increases, quadriceps muscles forces increase which causes the hamstring to quadriceps ratio to decrease, thus causing the ACL forces to increase which may lead to injury of the anterior cruciate ligament.

Thigh Muscle Force, Hamstring/Quadriceps Ratio and the Peak ACL Forces

Knee ligament loading in landing is determined by the balance of both muscle forces and ground reaction forces. Originally, we hypothesized the influence of muscle forces on the ACL that as the hamstring to quadriceps ratio would decrease the ACL forces would increase.

Previously, most ACL studies and prediction models have been done on non- dynamic tasks such as isolated isometric knee flexion angles with no or minimal ground reaction forces included. However, the present study is consistent with the current model of predicting ACL forces which is similar with the study by Cerulli, Benoit et al. (2003). Our findings were similar with Ceruilli, Benoit et al., where the ACL force increased as the ground reaction force increased (Figure 4). What was not comparable the Ceruilli, Benoit et al. study was that after peak ACL force, the force or strain, where strain and force are assumed to be correlated, remained almost constant, whereas, in our present study, after the peak ACL force, the force decreased.

Although the hamstrings are protective of the ACL, the data from this investigation showed that the hamstrings muscle forces were not correlated with peak ACL forces and suggests the existence of a global hamstring force strategy in landing that is independent of landing height. Another assumption could be that at the time of peak ACL force, the knee flexion angle was the same for all landing heights and hamstring muscle forces were the same since they are not a factor in the increased ground reactions forces unlike the quadriceps muscles. The quadriceps alone were seen as having a higher influence on peak ACL forces, as 67% of the peak ACL force were explained by the quadriceps. Moreover, the hamstrings and the hamstring to quadriceps ratio, related to 4.1% and 33.2% of the ACL force, respectively. This agrees with Li, Rudy et al., (1999), Li, Zayontz et al., (2004), and Markolf, O'Neill et al., (2004) in which they found that the quadriceps muscle forces alone produced higher forces in the ACL then either the hamstrings muscles acting alone or the combined hamstrings and quadriceps. The force produced by the hamstrings on the ACL was also found to be lower than the combined hamstring and quadriceps muscle forces (Li, Zayontz et al. 2004). Renstrom, Arms et al., (1986) also found that the ACL strain is higher when the quadriceps are activated isometrically, whereas the isometric hamstrings and combined hamstrings and quadriceps caused the ACL to strain less. Even though the quadriceps produced higher strain and forces and have stronger influences on the ACL, the quadriceps when working jointly with the hamstrings (i.e. hamstring to quadriceps ratio), ACL forces decrease. However, the correlations between the hamstring to quadriceps force ratio were stronger at lower heights compared to higher heights, suggesting that the balance between the hamstrings and quadriceps muscles played a stronger role in predicting ACL forces when the ground reaction forces were low compared to the highest heights.

A reason for this occurrence could be the range of motion of the knee. Since we are only investigating the knee angle at peak ACL force and not the full range of motion of the knee as in a squatting or knee extension study, we only get a snapshot of the knee. The knee flexion angle at ground contact was -25.2 ± 9.14°. This knee angle was pooled together from all three landing heights. At peak ACL force and with all the landing heights pooled together, the knee flexion angle was -39.7 ± 10.2°. In addition, the knee angle at the time of peak ACL force did not correlate with peak ACL force ($R = -0.125$, $p = 0.192$). The knee angle during peak ACL force is within the same range that Pflum, Shelburne et al, (2004) found when they the modeled ACL and the ACL was loaded when the knee was flexed 33° to 48°. This knee angle and the correlation of the thigh muscle forces and the peak ACL force at this knee angle, we can insinuate that our results additionally support the Renstrom, Arms et al. (1986), findings where the quadriceps muscles increase ACL strain from 0° to 45° of knee flexion, as well as the Li, Zayontz (2004) where the combined thigh muscle forces caused the ACL force to peak around 30° of knee flexion. Although the current results show a low to moderate influence of hamstring to quadriceps ratio compared to that suggested through other *in-vivo* and *in-vitro* studies (Beynnon and Fleming 1998; Renstrom, Arms et al. 1986; Li, Rudy et al. 1999), a primary difference was that vertical ground reaction forces were incorporated into the current study while these other research isolated the effect of muscle forces on ACL forces using minimal or no vertical ground reaction forces.

The ground reaction forces were more highly correlated to ACL forces than the hamstring to quadriceps ratio. The stepwise regression analysis showed the ground reaction forces explained 48% of the variance in peak ACL forces while the hamstring to quadriceps ratio added an additional 11% to the total variance of 59% ($p<0.01$). The contribution of the ground reaction force with the hamstring to quadriceps ratio increased also as height increased. This increase in ground reaction forces is comparable to other studies (Zhang, Bates et al. 2000; Yeow, Lee et al. 2010), which they also found that with this increase in ground reaction force, the knee joint moments and power also increases (Zhang, Bates et al. 2000; Yeow, Lee et al. 2010). The quadriceps are seen as the main contributors to producing this increase in joint moments and powers and this increase in quadriceps muscle demands may explain why the ground reaction forces are nearly as correlated to the peak ACL forces as are the quadriceps muscle alone.

Since a mixed gender subject pool was used, the question may come up that one gender may skew the data. Using a paired sample t-test, it resulted in finding that there were no significant differences between male and females subjects for the dependent variables; peak ACL force differed between male and females 22.5% ($p = 0.156$), the quadriceps forces differed by 11.7% ($p = 0.407$), the hamstrings force differed by 5.9% ($p = 0.748$), the hamstring to quadriceps ratio 18.6% ($p = 0.425$), and the ground reaction forces differed by 7.2% ($p = 0.765$)

(Table 5). Therefore, with the current sample of subjects, it is unlikely that gender had an effect on these results, although, the number of subjects were low and that may have affected the p-values.

Table 5. Comparison of the Sexes

		Mean	Std. Dev.	95% CI	Std. Error	P-value
Peak ACL Force	Male	0.900	0.220	(0.760, 1.040)	0.064	**0.156**
	Female	1.129	0.493	(0.816, 1.442)	0.142	
Quad Force	Male	6.243	1.801	(5.099, 7.388)	0.520	**0.407**
	Female	7.021	2.603	(5.350, 8.691)	0.759	
Ham Force	Male	1.070	0.580	(0.702, 1.439)	0.167	**0.748**
	Female	1.136	0.389	(0.889, 1.383)	0.112	
H/Q Ratio	Male	0.190	0.083	(0.138, 0.243)	0.024	**0.425**
	Female	0.230	0.145	(0.137,0.322)	0.042	
GRF	Male	2.541	1.695	(1.464, 3.618)	0.489	**0.765**
	Female	2.732	1.376	(1.858, 3.607)	0.397	

Table 5. Comparison of the male and female subjects. There was no significant difference between the male and female subject (*p*>0.05). All values are reported in body weights.

Figure 10. Comparison of the Sexes

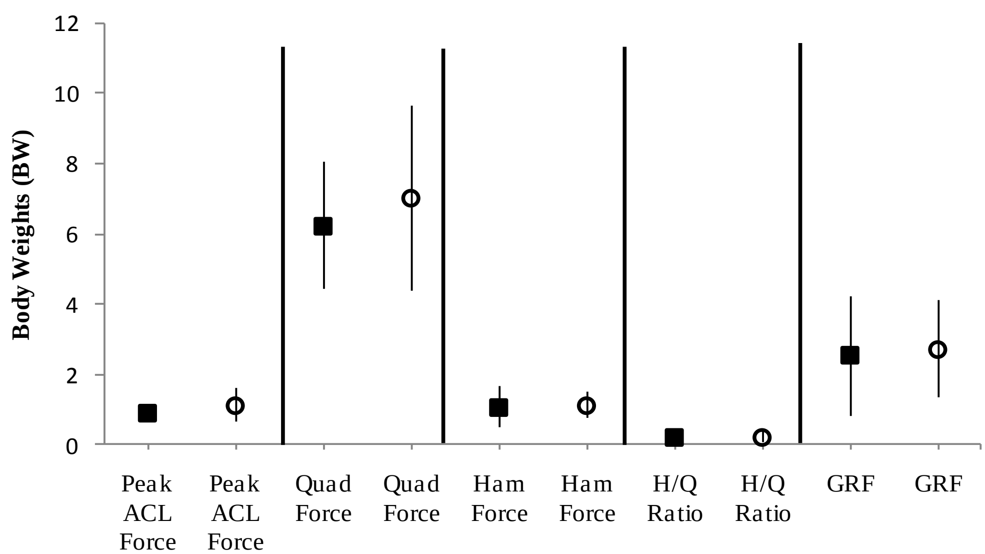

Figure 10. The comparison of the main variables between the male and female subjects. There was no significant difference between the male and female subject (*p*>0.05). ■ – Males, ○ – Females. The bars represent the 95% confidence interval.

Clinical Manipulation of Knee Loads

Since our findings show that the ACL force is greatly influenced by the quadriceps and that the quadriceps forces increase as landing height increases, how do we get recreational or professional athletes to try and decrease their risk of ACL injury? We could ask them to land with less quadriceps activation or to not try and land from such high heights and only jumping and landing from smaller heights but that is not possible. One solution is to land with less ground reaction force which inherently lowers the quadriceps demands as well. The ground reaction forces that are applied to the knee cause the ligaments in the knee to become more loaded since the loads that are applied to the knee ligaments are determined by the muscle forces that cross the knee and ground reaction forces. With our results having the ground reaction forces highly correlated via regression analysis, $R = 0.690$, to the peak ACL force, it would be advantageous for the participant in any athletic activity to find a way to try and lower their ground reaction forces and quadriceps muscle forces which could then decrease the ACL forces that are imparted on the knee. In an athletic activity that involves jumping and landing, the landing heights are not uniform. With an increase in landing height, the ground reaction forces increase (Zhang, Bates et al. 2000; Seegmiller and McCaw 2003). Also, when the landing height increases, range of motion of the joint increases which could be a strategy to overcome the force that is produced by the increase in landing height.

DeVita and Skelly's (1992) investigation of landing stiffness showed that when one lands with a more flexed knee that the ground reaction forces decrease. This decrease in ground reaction forces reduces the quadriceps muscles demands and loads produced around the knee, and based on this study's findings, would decrease the ACL. From previous studies, the more flexed the knee is, the more the hamstrings are incorporated with the quadriceps and this lowers the force and ACL strain (Renstrom, Arms et al. 1986; Durselen, Claes et al. 1995; Beynnon and Fleming 1998). The nonlinear correlation of the hamstring to quadriceps ratio to peak ACL of our study showed when ground reaction forces were lower (i.e., 15cm landing height), the hamstring to quadriceps ratio was a stronger predictor of ACL forces. The higher the hamstring to quadriceps ratio, meaning that when more of the hamstrings are activated during a task relative to the quadriceps, the peak ACL force decreases. When one uses a more flexed knee strategy, decreasing the quadriceps muscle forces, it increases the hamstring to quadriceps ratio, and the peak ACL force decreases, enabling one to decrease their risk of injury to their anterior cruciate ligament.

Summary of Discussion

With linear correlation analysis, the quadriceps was found to highly influence the peak ACL force, with the hamstrings being weakly influential to the peak ACL force. At the peak ACL force, the knee angle was $39 \pm 10°$, which is the knee angle where the hamstrings begin to activate along with the quadriceps to decrease the load on the ACL. The peak ACL force is found to be low to moderately negatively correlate with the hamstring to quadriceps ratio. With these findings, our hypothesis, that the hamstring to quadriceps ratio would be inversely related to the peak ACL force as landing height increases, was found to be correct, although, not as strongly as anticipated. However, as the landing height increases, the ground reaction forces become more of an influential factor in determining the peak magnitude of ACL forces than the hamstring to quadriceps ratio. The hamstring to quadriceps ratio becomes less of a factor at the higher heights then they do at the lower heights. This phenomenon could be due to that the quadriceps and ground reaction forces could be highly correlated. From, the present study, when the quadriceps muscle forces increase, the peak ACL forces increase, and that increase in quadriceps muscle force causes the hamstring to quadriceps ratio decreases. Also, as the ground reaction forces increase, which happens when the landing height increases, it requires a high demand of the quadriceps muscles. This then puts more emphasis on the ground reaction force and less on the hamstring to quadriceps ratio as landing height increases.

Conclusion

This study examined the relationship between the hamstring to quadriceps ratio and the peak ACL force and was tested by landing on a single leg from three different heights. It was hypothesized that the hamstring to quadriceps ratio would be negatively correlated to peak ACL forces. The study not only inspected the relationship between the hamstring-to-quadriceps ratio and the peak ACL force but also the relationship of the quadriceps muscle forces and the peak ACL force as well as the hamstrings muscle force and peak ACL relationship. The ground reactions forces were also included in a supplementary analysis predicting peak ACL force. The results showed the quadriceps forces were highly and significantly correlated with the peak ACL force, while the hamstring forces were not. Thus the hamstring to quadriceps ratio did not have a strong linear correlation to peak ACL force compared to the quadriceps alone. However, the hamstring to quadriceps ratio relationship with peak ACL forces was stronger at lower heights compared to higher heights regardless of the type of curve analysis employed.

In conclusion, the overall hypothesis of this study was that the hamstring to quadriceps ratio would be negatively correlated with peak ACL forces, which was found but the correlation was low to moderate. In addition, as landing heights increase, the contribution of the ground reaction forces predicting peak ACL forces increases while the contribution of the hamstring to quadriceps ratio decreases.

Future Recommendations

Studies investigating the mechanics and causes of injury to the anterior cruciate ligament have been used *in vivo*, and cadavers via *in vitro*. However, those studies typically use cadavers from older adults and a force transducer implanted in the knee can be cumbersome. Musculoskeletal modeling, like the one used in this study, have been used as well, to further research on the ACL. The limitation with musculoskeletal modeling is that it is assumed that each subject has the same physiological make up. Nevertheless, each person is made different and some models do not take in account for the contact forces or have all of the individual joints degrees of freedom. A model that incorporates these functions and behaves as a human being does will progress research and accurately predict ACL forces more than they are today.

With the present study, we grouped the quadriceps and hamstrings muscles together. If we were to isolate each quadriceps muscle or hamstring muscle and manipulate it, then we could better understand which quadriceps muscle or hamstrings muscle or a combination of muscles could potentially harm the ACL.

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APPENDIX A: INSTITUTIONAL REVIEW BOARD APPROVAL

University and Medical Center Institutional Review Board
East Carolina University, 600 Moye Boulevard
1L-09 Brody Medical Sciences Bldg. • Greenville, NC 27834
Office **252-744-2914** • Fax **252-744-2284** • **www.ecu.edu/irb**
Chair and Director of Biomedical IRB: L. Wiley Nifong, MD
Chair and Director of Behavioral and Social Science IRB: Susan L. McCammon, PhD

1907-2007 CENTENNIAL

TO: Anthony Kulas, PhD, Department of Health Ed. & Promotion, ECU, 249 Ward Sports Medicine Building

FROM: UMCIRB

DATE: February 3, 2010

RE: Expedited Continuing Review of a Research Study

TITLE: "Effects of Sex and Landing Height on Trunk and Lower Extremity Biomechanical Control in Landing"

UMCIRB #08-0006

The above referenced research study was initially reviewed and approved by expedited review on 1/14/08. This research study has undergone a subsequent continuing review using expedited review on 2/1/10. This research study is eligible for expedited review because it is a collection of data through noninvasive procedures (not involving general anesthesia or sedation) routinely employed in clinical practice, excluding procedures involving x-rays or microwaves. Where medical devices are employed, they must be cleared/approved for marketing. (Studies intended to evaluate the safety and effectiveness of the medical device are not generally eligible for expedited review, including studies of cleared medical devices for new indications.) Examples: (a) physical sensors that are applied either to the surface of the body or at a distance and do not involve input of significant amounts of energy into the subject or an invasion of the subject's privacy; (b) weighing or testing sensory acuity; (c) magnetic resonance imaging; (d) electrocardiography, electroencephalography, thermography, detection of naturally occurring radioactivity, electroretinography, ultrasound, diagnostic infrared imaging, doppler blood flow, and echocardiography; (e) moderate exercise, muscular strength testing, body composition assessment, and flexibility testing where appropriate given the age, weight, and health of the individual. Also, it is a collection of data from voice, video, digital, or image recordings made for research purposes. The Chairperson (or designee) deemed this **unfunded** study **no more than minimal risk** requiring a continuing review in **12 months**. Changes to this approved research may not be initiated without UMCIRB review except when necessary to eliminate an apparent immediate hazard to the participant. All unanticipated problems involving risks to participants and others must be promptly reported to the UMCIRB. The investigator must submit a continuing review/closure application to the UMCIRB prior to the date of study expiration. The investigator must adhere to all reporting requirements for this study.

The above referenced research study has been given approval for the period of **2/1/10** to **1/31/11**. The approval includes the following items:

- Continuing Review Form (dated 1/26/10)
- Informed consent (dated 10/16/07)
- Content of Preliminary Interview Prior to Participation
- Advertisement

The Chairperson (or designee) does not have a conflict of interest on this study.

The UMCIRB applies 45 CFR 46, Subparts A-D, to all research reviewed by the UMCIRB regardless of the funding source. 21 CFR 50 and 21 CFR 56 are applied to all research studies under the Food and Drug Administration regulation. The UMCIRB follows applicable International Conference on Harmonisation Good Clinical Practice guidelines.

APPENDIX B: SUBJECT CONSENT FORM

CONSENT DOCUMENT

Title of Research Study: Effects of Sex and Landing Height on Trunk and Lower Extremity Biomechanical Control In Landing
Principal Investigator: Anthony Kulas PhD, ATC, LAT
Institution: East Carolina University
Address:249 Ward Sports Medicine Building
Telephone #: 252-737-2884

INTRODUCTION

You have been asked to participate in a research study being conducted by Anthony Kulas PhD, ATC. This research study is being conducted to examine how performance of single-leg landing from different heights affect trunk and lower extremity control in males and females.

PLAN AND PROCEDURES

1) I will participate in an interview to determine whether I am eligible for the study. I understand that I cannot participate if I: 1) have a recent (last 6 months) injury to the lower extremity or low back; 2) have a history of recurrent injury to the lower extremity or low back; 3) have any history of low back and/or lower extremity surgeries. In order to be included in this study, I attest that I have participated in jumping and landing activities at the recreational level (i.e. intramurals, activity classes, or regularly scheduled pickup games) but not at the inter-collegiate level.
2) I will be tested once. All data will be collected and analyzed in the Athletic Training Research Laboratory located in Sports Medicine Room in Scales Fieldhouse.
3) I will be tested while wearing athletic shoes, a t-shirt, shorts and an athletic bra (if applicable). My height and weight will be measured. Pairs of non-invasive surface electrodes will be placed on my abdomen, back, and lower extremity to monitor the electrical activity of the muscles. I will then perform several maximal contractions to ensure correct electrode placement and record representative maximal efforts by each of the muscles. Reflective markers will then be placed on my shoulders, back, pelvis, and right hip, knee, ankle, and foot. Following setup, I will perform four sets of eight single-leg drop landings each from different heights in the following predetermined order: 15cm, 30cm, 45cm, 15cm.
4) I understand that the instruments used in this study include a six camera motion analysis system to record how I land, a force platform to measure the forces between my foot and the floor, and an electromyography unit which measures the electrical activity of my muscles. The force platform is a metal plate firmly mounted to the floor and landing on it will feel like a regular landing.
5) I will be allowed to rest any time I feel tired or have pain during the testing session.
6) I understand that if I am bothered by any pain I can quit the testing session without any penalty or repercussions.

Participant's initials

POTENTIAL RISKS AND DISCOMFORTS

A few possible risks are involved in this study. Because this project involves physical activity, there is always the risk of physical injury. I will be required to land from 15, 30, 45, and 15 centimeter heights, during which I could potentially be injured. Although these landings may be from a slightly higher height than I am used to, it will be performed in a controlled environment and possesses less risk than during a typical game situation. I may feel some fatigue or discomfort within 24-48 hours after the test. However, my experience in jumping and landing activities such as volleyball, basketball, or gymnastics may eliminate any discomfort. I also understand that if I feel pain, I will be allowed to rest until the pain has subsided. If the pain does not subside I can stop the testing until I feel better. One or more research assistants will always be present during all testing.

POTENTIAL BENEFITS

The benefits of this project far outweigh the risks. We expect to gain a better understanding of males and females adapt to increasing height by assessing trunk and lower extremity control. Landing from different heights are common situations encountered in jumping and landing activities in sports. Other researchers and future research performed by the PI will build on these results. It is expected that the research will be disseminated in the biomedical research literature.

SUBJECT PRIVACY AND CONFIDENTIALITY OF RECORDS

I understand that my personal data will be held in strict confidence by the researchers and that a code number will be used to identify my data. I understand that my name will never be associated with any of the data or results in any public presentation or publication of this study. All data and records will be stored in room Sports Medicine Room in Scales Fieldhouse. Dr. Kulas and a research assistant will have access to the data. After the study is completed the data will be permanently stored in room 249 Ward Sports Medicine Building.

Video excerpts and/or still photographs from data recordings created during research are sometimes used in research presentations or publications, and/or for educational purposes such as a classroom lecture. I MAY INDICATE NOW MY PREFERENCE ON THE USE OF MY DATA FOR OTHER RESEARCH OR EDUCATION PURPOSES OTHER THAN DATA COLLECTION BY INITIALING TO THE RIGHT OF EIHTER "YES" OR "NO" HERE: YES______: NO______.

By failing to initial either yes or no above, none of my data will be used for research or education purposes other than the intended use for this study as outlined in this document. If you deny permission to use the recording for research or educational purposes, the tape will be used only for analysis, and will be destroyed at the end of the research study.

COMPENSATION

I will not be compensated for my participation in the study. The policy of East Carolina University does not provide for compensation or medical treatment for subjects because of physical or other injury resulting from this research activity. However, every effort will be made to make the facilities of Health Services available for treatment in the event of such physical injury.

VOLUNTARY PARTICIPATION

Participating in this study is voluntary. If you decide not to be in this study after it has already started, you may stop at any time without losing benefits that you should normally receive. You may stop at any time you choose without penalty.

PERSONS TO CONTACT WITH QUESTIONS

The investigators will be available to answer any questions concerning this research, now or in the future. You may contact the principal investigator, Anthony Kulas at phone numbers 252-737-2884 (days) or 252-414-9374 (nights and weekends). If you have questions about your rights as a research subject, you may call the Chair of the University and Medical Center Institutional Review Board at phone number 252-744-2914 (days), and/or the ECU Risk Management Office at 252-328-6858.

CONSENT TO PARTICIPATE

Title of research study: Effects of Sex and Landing Height on Trunk and Lower Extremity Biomechanical Control In Landing

I have read all of the above information, asked questions and have received satisfactory answers in areas I did not understand. (A copy of this signed and dated consent form will be given to the person signing this form as the participant or as the participant authorized representative.)

Participant's Name **(PRINT)** | **Signature** | **Date** | **Time**

If applicable:

Guardian's Name **(PRINT)** | **Signature** | **Date** | **Time**

PERSON ADMINISTERING CONSENT: I have conducted the consent process and orally reviewed the contents of the consent document. I believe the participant understands the research.

Person Obtaining consent **(PRINT)** | **Signature** | **Date**

Principal Investigator's **(PRINT)** | **Signature** | **Date**

Participant's initials