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Influence of Fatigue on Jump and Land Movement Patterns

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ABSTRACT

BACKGROUND AND PURPOSE: Injuries to the anterior cruciate ligament (ACL) are common among both male and female athletes. Both the female gender and fatigue have been demonstrated to increase injury rates. This research aims to reconcile the movement pattern and fatigue protocol with what is seen in sport, while including both men and women to see differences between knee biomechanics. The research will compare the lower extremity biomechanics of a jump-land (double and single leg) between healthy men and women after a sports specific fatigue protocol. Ultimately, this research is intended to examine movement patterns which may predispose the subject to ACL injury.

METHODS: Twenty healthy subjects were studied (26.3 ± 3.5 years old), 10 of which were female. A 3D electromagnetic system measured knee kinematics and kinetics during 3 jumping tasks. The subjects completed 4 sets of 3 different, randomized jumps (bilateral to bilateral, bilateral to single-leg right, and bilateral to single-leg left) on force plates. The subjects completed a fatigue protocol consisting of jumping, sprinting, step-ups, and an agility ladder and were immediately re-assessed by completing the 3 different jumps. A paired t-test was used to analyze pre and post fatigue and a one-way ANOVA was used for analyzing gender comparisons for each variable.

RESULTS: Significant differences were found between pre-fatigue and post-fatigue internal rotation and adduction knee angles for all 3 landings; other angles and knee moments were significantly different dependent on type of landing. When comparing gender for each variable, internal adduction moments and ground reaction forces were significantly different for all landings. Knee angles were also significantly different dependent on type of landing and dependent variable. Finally, females demonstrated greater biomechanical changes in landing mechanics post-fatigue than males.

CONCLUSION: The results support previous literature that fatigue and gender have an impact on jump and land movement patterns at the knee. The differences in knee angles and moments from the current study, as seen by internal adduction moments and ground reaction forces, demonstrate that fatigue and the female gender are risk factors for ACL injury. This may support the current pattern of greater ACL injuries in female athletes, especially when doing a jump-land movement.

The undersigned certify that they have read, and recommended approval of the research project entitled:

INFLUENCE OF FATIGUE ON JUMP AND LAND MOVEMENT PATTERNS

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in partial fulfillment of the requirements for the Doctor of Physical Therapy Program

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CHAPTER I
INTRODUCTION

Injuries to the anterior cruciate ligament (ACL) are common; a 2007 meta-analysis estimated that annually over 200,000 injuries occur in the United States.[1] According to a study looking at the societal and economic impact of ACL tears, these new injuries and their long term effects annually cost society between eight billion and eighteen billion dollars.[2] Seventy percent of these incidences occur via a non-contact mechanism; often as a result of increased dynamic valgus and internal or external rotation of the knee. In regards to contact injuries, the mechanism often involves contact with a valgus stress while the foot remains planted.[3]

Females are at a higher risk for tearing the ACL. Depending upon the sport the individual participates in, females have approximately four to nine times the injury rate of males. Anatomical and structural factors, hormonal factors, and neuromuscular and biomechanical factors have been suggested as reasons for the increased incidence among females.[4] Chandrashekar et al.[5] found that having a smaller intercondylar notch angle is predictive of ACL tear for women. The same study also found that women have smaller ACLs in length, cross-sectional area, and volume. A number of studies have suggested that women tend to have greater ligamentous laxity because of less collagen, resulting in failure at lower loads.[5,6] Several studies have also demonstrated the influence of a woman's hormones on ligament properties. These studies have found that ligament laxity and collagen strength vary throughout a woman's menstrual cycle.[7,8] It has been proposed that the higher rate of ACL tears among women may be related to poor dynamic muscular control. It is common to see increased dynamic valgus and increased loading of the ACL, secondary to hip and core muscle weakness in women.[4] Women also tend to display more aberrant muscle recruitment or firing patterns, displaying increased or more rapid activation of the quadriceps muscles in comparison to the hamstrings.[4] Females also demonstrate reduced hip and knee flexion angles upon landing; this results in increased ground reaction forces at impact and increased stress on the ACL.[4,9]

In the United States 90% of patients who tear their ACLs go on to have surgical reconstruction.[10,11] The purpose of the reconstruction is two-fold: one, by recreating the ligament through autograft or allograft, restoration of the knees biomechanical control over anterior tibial translation is achieved, thus reducing the shear forces on the knee and hopefully slowing the progression of osteoarthritic changes over time and secondly, to restore function to the individual allowing them to return to an active lifestyle. Using the Multicenter Orthopaedic Outcome Network database, Wright et al.[12] found a 3% graft rate failure and a 3% rate of tears in the contralateral knee occur during the first two years after ACL repair. Five years after surgery, Salmon et al.[13] found graft rupture to occur within 6% of the population and contralateral ACL rupture to occur in another 6% of the population. Shelbourne et al.ref further looked at the differences between men and women as to which knee becomes injured after surgery based specifically on age, activity level, and time to return to sport. Similar to Salmon et al. [13], the study found that within five years 5.3% of participants had suffered an ACL injury to the contralateral knee and 4.3% had suffered an ACL injury to the repaired knee. In further analysis, Shelbourne et al.ref found that women suffered more injuries to the contralateral leg than men but not more injuries to the reconstructed knee. The report further details that younger patients were at a higher risk for a second ACL injury (17% for those younger than 18) as compared to older patients (4% for those older than 25).[14] One study by Paterno et al.[15] places the rate of reinjury as high as 27%.Patients post-ACLR see a marked decrease in proprioception as is evident in a study by Delahunt et al.[16], which showed a decreased postural stability of individuals with ACLR as compared to healthy controls on the Star Excursion Balance Test in the posterior medial and posterior lateral direction.

Likewise patient's post-ACLR note adverse effects on their jumping and cutting performance. This is linked, again, to problems with neuromuscular control, and muscular strength, which translates into poor movement patterns that put athletes at risk for tearing an ACL again with rates as high as 27%.[14,15] Multiple studies support the changes in vertical ground reaction forces (VGRFs) when compared to subjects' healthy knees.[11,17] These risk factors are furthered confirmed in studies that examine single leg tasks. Ortiz et al.[18] examined single leg drop and hop tasks, finding that in lower eccentrically demanding tasks, there was little difference in mechanics between involved and uninvolved limbs, however when the eccentric demand increased greater dynamic knee valgus and knee extension moments occurred.

All athletes reach a state of fatigue at some point. Extensive research has been conducted to determine if fatigue has a direct impact on injury risk through analysis of VGRFs, hip/knee/ankle kinematic changes, jump height, etc. Fatigue, in studies, is mimicked in different ways. Some studies choose endurance tasks such as running until failure is reached, others focus primarily on fatigue through muscular activity such as squats, while some try a combination by simulating gameplay as closely as possible by mixing endurance activities with plyometric movements to reach a full body fatigue before measuring lower extremity (LE) movement patterns. Post fatigue research, with healthy subjects, has observed decreased muscle activation of the quadriceps and hamstrings,[19] decreased jump height,[20,21] and changes in knee kinematics.[22,23] When measuring patients post ACLR, fatigue has shown to decrease hip and knee flexion, increase hip rotation, and postural sway in both healthy individuals and patients post ACLR increasing the risk of graft tear.[23,24]

Purpose

This research aims to reconcile the movement pattern and fatigue protocol with what is seen in sport, while also including both men and women, to note differences between knee and hip kinematics and kinetics. The research will compare the lower extremity kinematics and kinetics of a jump-land (double and single leg land) between healthy men and women after a fatigue protocol. Ultimately, this research may identify whether a dysfunctional or high risk movement pattern exists which may predispose the subject (male and/or female) to possible knee or ACL injury.

Hypothesis

Our hypothesis is that all subjects will display negative changes in their movement patterns. However, we expect that women, as compared to men, will exhibit more aberrant lower extremity kinematics and kinetics upon fatigue, further increasing their relative risk of ACL injury.

CHAPTER II

REVIEW OF RELATED LITERATURE

Gender Differences in Lower Extremity Biomechanics

The amount of literature available on biomechanical variances between men and women at the knee is vast. Through a review of the literature, a few main points were consistently seen in the results of many studies. A possible explanation for the biomechanical sex differences could be that, on average, men typically have more lean mass than females.[25] Montgomery et al.[25] conducted a study to explore sex differences in the absorption of energy in the lower extremity when amounts of lean mass were accounted for. They did this by calculating the landing height of a drop vertical jump test from the amount of lean mass available in each subject to dissipate energy. Each participant was assigned a specific drop height for their gender and amount of lean mass. This was calculated in an attempt to equalize the relative task demands. Participants received a fan-beam DXA scan to determine his or her body composition. Once the DXA scan was completed, each subject was paired with an opposite sex participant with the same BMI.

Subjects consisted of 35 male and female pairs.[25] Participants were recruited from NCAA Division I and club soccer and basketball athletes. Exclusion criteria consisted of: suspected pregnancy, a BMI greater than 30 kg/m[26], current lower extremity pain, and a history of lower extremity injury. Before the drop vertical jump test, each participant completed a dynamic warm-up. The testing consisted of five successful trials from the assigned drop height. Female participants were tested at their assigned height as well as the height of their male counterpart. This second testing height was used to compare biomechanics when females were subjected to an exaggerated task difficulty. The researchers did not see the biomechanical sex differences that they were expecting. The only statistically significant finding in this study was that males absorbed more total energy at the hip. Possible limitations include the use of highly trained female athletes and a drop height that was not high enough to produce biomechanical differences between the sexes.[25] The researchers concluded that it is likely that the biomechanical differences seen between the sexes arises from a factor other than relative task difficulty.

Olsen et al.[3] analyzed 20 videos, over 12 seasons 1988-2000, of ACL injuries. Three doctors and national team coaches were brought in to systematically analyze the videos for common ACL mechanisms, along with 32 ACL-injured players to recall and compare injury characteristics of the video analysis. Video was broken down into: player position, playing phase, activity, push-off knee, takeoff knee, landing leg, ball handling, contact with another player, disturbed by another player, balance, attention, speed, and anything unusual about the play. Coaches broke down and analyzed the game aspects and physicians analyzed knee kinematics (foot position at foot strike, knee position at foot strike, when during the phase did the injury occur, movement direction at the time, and weight distribution between the players legs).[3]

Statistical analysis was compared between the analyses of the different doctors and coaches.[3] Plant and cut was the main mechanism of injury followed by single leg landing; in all the cases the foot was placed outside of the knee in a position of slight flexion, valgus and external/internal rotation of the tibia. Sixty-six of 78 responses had agreement, while 7 of the 12 disagreements were over contact or disturbance by an opponent, and the last five were player position and activity at time of injury.[3] These findings are consistent with those supported in research, providing further bridging between the lab and real world settings.

A study completed by Lyle et al.[26] investigated the relationship between lower extremity dexterity and differing movement patterns in females versus males. Lower extremity dexterity is defined as: "the ability to dynamically control endpoint force magnitude and direction.[26]" To quantify dexterity, the researchers used the lower extremity dexterity test. This requires participants to compress an unstable spring with their foot while attempting to attain the highest vertical force possible. While performing the lower extremity dexterity test, participants were positioned such that all extremities and trunk were supported excluding the lower extremity that was being tested. Five practice tests were allowed before testing began. Each test consisted of a 16 second trial in which the subject was instructed to "slowly compress the spring with their foot with the goal to raise the force feedback line as high as possible and keep it there.[26]" Subjects each completed between 21 and 25 trials with a 30 second rest between each test and a two minute break between each fifth test.

This study also examined biomechanical landing differences between men and women using a single-leg drop jump task.[26] Each participant completed four trials in which they were required to drop from a 30 cm platform, land on a single leg in the middle of a force plate, and then perform a maximum vertical jump using their dominant lower extremity. Three-dimensional kinematic data was obtained using an 11 camera system, and ground reaction forces were calculated using the data from the force plate. Subjects consisted of 14 male and female high school soccer athletes.[26] Participants were excluded from this study if they reported: history of ACL injury, history of knee surgery, or an injury that inhibited their full participation in soccer for more than three weeks in the last six months. The researchers discovered that females had decreased lower extremity dexterity when compared to males. Lyle et al.[26] further found that females had increased leg stiffness, higher rates of ankle and knee coactivation, and earlier peak vertical ground reaction forces in females than their male counterparts. One possible explanation for these findings is that females exhibit an anticipatory stiffness reaction before landing partially due to reduced lower extremity dexterity. The authors infer that this could contribute to higher ACL injury rates in females, as well as explain why exercise interventions have a positive impact on injury reduction. A limitation to this study includes a small sample size, as the researchers only had 28 participants.[26]

Ali et al.[27] authored a research article comparing total body mechanics of recreationally active males and females. Participants completed single-leg landings with their dominant lower extremity at various heights and distances. Leg dominance was established by asking each subject which leg they used to kick a ball. Each subject completed landings from a platform height of 20, 40, and 60 cm and were instructed to jump horizontal distances consisting of 30, 50, and 70 cm. These different heights and distances resulted in nine different landing conditions. Six male and six female subjects participated in this study. Each subject was a recreational athlete, which was defined as:

"[participating in a] jump landing sport for 30 minutes a day at least 3 times a week.[27]" Exclusion criteria consisted of any lower extremity injury at the time of data collection. Before collecting data, each subject was given time to warm-up and practice single-leg landings. Each participant was given enough time for them to feel comfortable with the task. Participants were instructed to: "stand on their dominant leg, jump forward, and land as naturally as possible with their dominant foot only centered on the force plate.[27]" Furthermore, the subjects were instructed to: "keep their hands on their iliac crests when landing to reduce any variability from swinging arms.[27]" Each participant completed two single-leg landings from each of the nine landing configurations. One trial out of the two completed was selected from each configuration from each participant for data analysis. A seven camera system was utilized to capture data from 42 retro-reflective markers placed on each subject. The force plate data was synchronized with the video information.[27]

Through analysis of their collected data, the researchers found that higher trunk flexion angles were associated with lower normalized peak vertical ground reaction forces, and that higher knee flexion correlated with lower peak knee abduction.[27] Furthermore, females had higher peak vertical ground reaction forces and a smaller ankle plantar flexion angle when compared to their male counterparts. These findings suggest that females generally have less ability to alter their ground reaction forces, increasing their risk for non-contact ACL injuries. Additionally, the female participants' limited ankle plantar flexion angles resulted in a rearfoot landing strategy. This may also contribute to their increased risk for ACL tears. Limitations include a small sample size leading to an inability to generalize these finding to the larger population.

Schmitz et al.[28] completed a study further examining sex differences in single-leg landings. Each subject completed a 45 minute data collection session. This session consisted of three to six practice jump-landings and five landings that were recorded for further data analysis. Subjects were instructed to jump down from a 0.3 m platform and land in the middle of a force place located 0.1 m away. These landings were completed barefoot with the subject's' dominant lower extremity. Researchers had their subjects keep their hands on their iliac crests to reduce variability based on arm swing. Data was obtained with the use of a force plate and six-degrees of freedom sensors attached to: "the anterior mid-shaft of the third metatarsal, the mid-shaft of the medial tibia, the lateral aspect of the mid-shaft of the femur, the sacrum, and over the C7 spinous process.[28]" Participants consisted of 14 recreationally active men and women, which resulted in 28 total subjects.[28] Recreationally active was defined as participation in any sport for at least 30 minutes three times per week or more. All participants were college students and did not have a history of orthopedic surgery or neurological conditions affecting the lower extremities.

Researchers found that females demonstrated a "stiff" landing style.[28] Females absorbed less total energy with their extensor muscles when normalized for body mass. Athletes with a stiffer landing style have larger ground reaction forces. Landings with a 20% increase in ground reaction forces have demonstrated increased rates of ACL tear.[28] The most limiting factor of this study, as reported by the authors, was that the landing height was set at 0.3 meters for both genders despite males being taller and more massive on average. This possibly increased the task difficulty for females compared to males.[28]

As sex differences in knee biomechanics continue to be observed in adult populations, the question remained whether these sex differences are demonstrated in children. Swartz et al.[29] investigated the relationships between developmental stage, sex, and biomechanics of jump landings. A maximum vertical jump height was determined with the best of three attempts for each subject. After the maximum vertical height was determined, a ball was suspended at 50% of their maximum vertical jump. Participants were required to jump to the ball and land in a double leg stance. However, data was only collected from the subject's dominant limb. When landing, subjects were instructed to land with both feet with their dominant foot in the middle of one force plate. A six-camera system was used to collect three dimensional data from markers placed on the dominant limb. Data collection occurred at two points during the landing: at initial contact and at peak vertical ground reaction force. Measurements at these points included flexion angles of the hip and knee, as well as knee valgus angles.

Participants consisted of 58 subjects without a history of back or lower extremity injuries.[29] Subjects were separated into groups based on their gender and developmental stage. Girls from seven to ten and boys from eight to eleven were included in the prepubescent developmental stage. These children were active in a youth sports program that included jump landing activities. Adults from 19-29 were included in the post-pubescent developmental stage. To be included in the study, adults were required to participate in 30 minutes of activity at least three times per week. Exclusion criteria included participation in an NCAA Division I jumping sport and the inability to complete a mature vertical jump. A mature jump pattern was defined as having a "preparatory crouch with 60° to 90° of knee flexion and a countermovement arm swing coordinated with complete extension at the hips, knees, and ankles at takeoff.[29]" Results of this study were inconclusive. The only statistical differences were seen in the biomechanics of children versus adults.[29] Children produced smaller hip and knee flexion angles while demonstrating higher knee valgus angles. Sex differences in landing biomechanics were not supported by this research. Limitations of this study include a task difficulty that was possibly too low to elicit biomechanical differences between genders.[29] More research with larger sample sizes would be useful to clarify these inconsistencies in the literature.

In summary, females demonstrate stiffer landing styles than their male counterparts. This single finding may have a profound impact on the ACL injury rates in healthy males versus females. Stiffer landings typically equate to increased ground reaction forces, which have been shown to increase rates of non-contact ACL tears.[28]

A study by Jacobs et al.[30] evaluated the hip abductor function and lower extremity landing kinematics of females and males. In this study there were 15 women and 15 men that participated. The subjects were volunteers that were free from orthopedic injury within the past 6 months. Ascension's Flock of Birds protocol was used to acquire three dimensional joint kinematics of the hip and knee, with sensors placed on the sacrum, distal lateral thigh, and proximal lateral shank.[30] The positioning of the subjects' hip joints were centered using the procedure by Leardini et al.[31] Prior to testing, a 5-minute warm-up was completed on the cycle ergometer.

The testing itself consisted of pre-fatigue functional landing trials, strength testing, endurance testing, submaximal 30-second isometric exercise, and post-fatigue landing trials. The functional landing tasks included a two-legged to one-legged forward jump to a target area which was a distance of 40% of their height. Height of the jump was also standardized as 15% of the subject's height. Joint angles were taken upon initial ground contact and peak joint displacement was calculated using joint data up to 500ms following initial contact. Following the pre-exercise landing trials, hip abductor strength and endurance tests were performed. Three 5-second maximal voluntary isometric contractions (MVICs) of the hip abductors were completed with thirty second rest periods in between. Strength and endurance were quantified using a dynamometer in sidelying. For endurance, the subjects completed a submaximal isometric contraction of the hip abductors until they were unable to hold the load for more than 3 consecutive seconds. EMG activity of the hip abductors was collected during the endurance testing to ensure fatigue of the hip abductors. The participants then had a 15-minute rest period followed by a 30 second bout of isometric hip abduction followed by the post-exercise landing trials. The results showed that overall women demonstrated higher valgus positions than men during landing.[30]

Another study by Russell et al.[32] examined frontal plane knee angles during initial contact (IC) following a drop landing. There were 16 men and 16 women participants who were free from lower extremity injury. The lower extremities were tracked with a 10-camera analysis system for both static and dynamic movements. A force plate was also used to analyze landing ground reaction forces (GRF). Surface electrodes were placed on the dominant limb to analyze EMG activity of the gluteus medius (GM). The drop landing procedure consisted of the subject standing on a 60 cm box and dropping onto the dominant limb in the force plate area on the floor below. In order for a trial to be considered successful, the subject was required to stick the landing for at least 2 seconds and to keep the opposite limb from touching the ground. Maximal knee flexion, initial contact, and gluteus medius EMG activity were collected and analyzed. The results show that females landed in valgus, while males landed in varus upon contact with the ground, when taking into account only frontal plane motions.[32] At maximal knee flexion angles, males also reached a greater varus position than females.[32]

A study by Brown et al.[33] used three-dimensional LE coordinates with anticipated and unanticipated cutting. The participants included 13 men and 13 women who did not have a history of serious knee injury, surgery, arthritis, or current knee pain. Leg dominance of the participants was determined by subjectively asking which leg they would use to kick a ball the farthest. The procedure of testing included single leg jump landing tasks with a randomized cutting maneuver. The type of cut (left or right) was determined by a colored light source. An anticipated cut was described by a light that was shown prior to the participant jumping off of a block. A light shown as the jumper crossed the boxes' anterior plane simulated an unanticipated jump. If the participant was shown a light to cut left, they would land on their right leg and cut immediately to the left. Similarly, if the light shown was to cut right, the participant would land on their left leg and cut to the right. The results showed that females presented with an increase in IC hip flexion, adduction, and internal rotation when compared with men during anticipated and unanticipated SL jump landings involving a cut.[33] When comparing dominant and non-dominant LEs, there was a significant increase in non-dominant hip and knee internal rotation angles during peak stance when compared to the opposite limb. This shows that there is potentially a difference between LE kinematics between the dominant and nondominant LEs.[33] These researchers suggest that there is a minimal increased risk of ACL injury with the slight difference in LE kinematics and explain that there would need to be further research to show increased disparities in leg dominance. It appears that there may not be consistency in the research with regards to the definition of leg dominance, which may be defined by one study as that leg which can kick a ball farthest or the leg you stabilize on by another study. When comparing anticipated and unanticipated jump cuts, there was a decrease in hip flexion during unanticipated cuts.[33] This could potentially place a person at a greater risk of injury by putting the hamstrings at a mechanical disadvantage to oppose anterior tibial shear loads.[33] The decreased ability of the muscles to stabilize the surrounding knee may increase the risk for injuring a person's ACL.[33]

Another study looking at both EMG activity of men and women was conducted by Dwyer et al.[34] This study consisted of 22 men and 22 women between the ages of 18 and 40. The participants were free from major lower extremity injury, LE surgery, and were able to complete the three functional tasks being studied. Three-dimensional kinematic data were collected at the knee and hip joints using the Ascension Flock of Birds protocol as mentioned previously. EMG data was collected using a 16-lead system. The skin on the muscle belly of the gluteus medius (GM) was shaved, cleansed, and abraded to decrease impedance. The GM muscle activity of the dominant limb was collected as described by Dwyer et al.[34] The procedure used for examining the participants included a single leg squat, lunge, and a step-up-and-over task which were taught to the participants by one of the researchers prior to data collection. A 5 minute warm-up was completed by each participant. The warm-up consisted of riding a bicycle, followed by a LE flexibility program concentrating on the hip flexors, hamstrings, quadriceps, and hip adductors. The electrodes were then placed on the skin and the participants completed three MVICs for the above 5 muscle groups. MVICs were performed 3 times each trial lasting 3 seconds, with a 30 minute break between each trial and 2 minutes between each muscle group. Following MVICs of the muscle groups, Flock of Bird's protocol was used to determine resting hip, knee, and ankle joint positions. The single-leg squat, lunge, and step-up-and-over exercises were performed randomly 3 times each with 30 second rest breaks between trials and 2 minute rests between exercises. The results of this study showed that women had smaller peak knee flexion angles during all three tasks when compared to men.[34] When looking at knee valgus, there were no significant differences between men and women during all tasks. Peak hip flexion angles in women were smaller than men during the single-leg squat activity, whereas peak hip extension was greater in all tasks for females when compared to males. There was no difference between hip adduction or hip external rotation during any activities and when comparing genders.

Gender Differences in Electromyography (EMG) of the Lower Extremity

Women are at a much higher risk for tearing their anterior cruciate ligament (ACL) in a noncontact situation than males and as a result, the comparison of biomechanics and gender has been increasingly studied.[30,32,35,36] Muscle activation, arc of motion, kinematics, and torque are all variables that have been found to differ between the two sexes.[30,32,34-36] Electromyographic (EMG) activity of the lower extremity has been studied previously during many closed kinetic chain (CKC) activities.[32,34, 35] A study by Youdas et al.[35] was completed to determine EMG activity in the lower extremities comparing men and women on both stable and labial surfaces. Fifteen male and 15 female participants with no lower extremity pathology and a normal (grade 5) MMT score of the LE were included. The subjects were recruited from a graduate school physical therapy program and were all recreationally active. During testing, EMG electrodes were placed on the muscle belly of the quadriceps and hamstring muscles following an alcohol cleanse to reduce impedance. The ground electrode was placed on the anteromedial aspect of the tibia. To get a base measurement the researchers manually resisted the subjects' knee extensors and flexors separately for 5 seconds to get a MVIC. Order of labile or stable surface was determined randomly. The subjects completed a single leg squat barefooted with their arms crossed over their chest while flexing their trunk forward. The subjects squatted to a 45 degree angle using a 10 second cadence for practice. During the testing portion, the extension phase of the movement was analyzed for EMG activity. One single leg squat (SLS) on each limb was tested randomly with a one minute rest period in between.[35] EMG data was only taken from the 5 second extension phase of the SLS. During the SLS activity, it was found that females have a significantly higher activation of the quadriceps than males on both stable and labial surfaces.[34,35] In contrast, the results showed that males have a significantly higher EMG muscle activation of the hamstrings than females, which supports the hypothesis that females are quadriceps dominant and males are hamstring dominant.[35] Since females tend to be quadriceps dominant, this could create an increased shear of the anterior tibia on the femur and a decreased ability to stabilize the knee during activity, which in turn could place more stress on the ACL.[35]

A different study by Stern et al.[36] looked at EMG activity, motor evoked potentials (MEPs), and muscle strength of men and women pre and post-fatigue. Participants included 17 males and 17 females that were recreationally active. Subjects that had a concussion within the last 2 years, LE surgery in the past 6 months, LE joint or muscle sprain within the last 6 weeks, or were unable to perform aerobic activity were excluded from the study. EMG electrodes were placed over the muscle belly of the vastus lateralis and biceps femoris. They were marked with permanent marker to ensure their placement throughout testing. The subjects initially completed a 5 minute bike at a self-selected pace for a warm-up. Baseline measurements of MVICs of the knee extensors and flexors were then taken using EMG. This was followed by a functional exercise protocol to induce fatigue. The protocol consisted of treadmill walking and body-weight-resisted exercise. The treadmill endurance began at a self-selected pace at 0% incline. Incline was increased by 0.5% each minute for 5 minutes. Once completed, 10 alternating step-ups and 10 body squats were repeated until 1 minute was finished. The total 6 minutes of exercise was repeated 5 times for a total combined exercise of 30 minutes. Following each walking segment, rate of perceived exertion (RPE) was taken. Measurements were taken immediately following exercise in the order of MEP, EMG, and strength. This took no longer than 10 minutes. This study reported that males had increased torque of the knee extensors and higher MEP amplitudes of the hamstrings than females pre-exercise. Post-exercise, females exhibited decreased vastus lateralis activation and increased knee-extension torque when compared to males, which can indicate a difference in how each gender reacts to fatigue. The researchers suggest that this could potentially create an increased risk of knee injuries in females, similar to the previous study.[36]

Another study looking at EMG data was described previously.[34] The results of the study by Dwyer et al.[34] showed that women were found to have increased gluteus maximus activation during the three CKC exercises. This is hypothesized to be due to an overall decrease in strength in women when compared to men, because when overall strength is decreased the amount of activation needed for a task would be increased.[34] During SL drop jump landing, the gluteus medius was found to have no differences in activation between sexes.[34] Although there was not a significant difference in healthy individuals, the researchers suggest that in patients with lower extremity injury, there is a possibility for an increased difference in gluteus medius activation as shown in other studies. Therefore it is hypothesized that the gluteus medius could be responsible for knee valgus during activity when a lower extremity injury has occurred.[34]

A study by Zazulak et al.[9] looked at gender differences in EMG activity of hip musculature during a single-leg landing. They specifically studied hip-stabilizing muscles, including the gluteus maximus, gluteus medius and rectus femoris. Thirteen female and 9 male Division I soccer and track athletes were recruited to perform drop landings onto their dominant leg, as determined by participant report of which leg they would use to kick a ball. The subjects performed 5 trials of single-leg landings from two different box heights, 30.5 and 45.8 cm, and were told to "drop off the box, land on the platform, and hold the position for at least 1 second".[9] Peak and mean EMG data were recorded for two different time periods via surface electrodes. The 200 millisecond time window before initial contact and the 250 millisecond time window after initial contact were recorded. The EMG data were compared between gender using a 2 x 2 mixed-model analysis of variance. The study reported significantly lower peak and mean gluteus maximus activation in females compared to males after contact, and no difference before contact.[9] No differences were found between genders for gluteus medius muscle activation before or after contact. Before contact, females had significantly greater peak rectus femoris activation and no other differences were found in regards to the rectus femoris activation. The authors suggested the decrease in gluteus maximus activation may demonstrate that females may have a greater difficulty controlling the hip during dynamic movement. This may lead to altered energy absorption during landing and result in increased ground reaction forces believed to be associated with ACL injury.[9] Because imbalanced quadriceps contraction has been found to increase ACL strain, Zazulak et al.[9] also suggested that the increased activation of the rectus femoris may contribute to increased ACL strain and other factors associating with ACL injury. Therefore, the differences in EMG activity may play a role in the increased ACL injury rates in females.

Effects Of Fatigue on Healthy Subjects

Many of the previous studies on biomechanics and kinematics have been completed on subjects that have not had physical stress placed on them. Given the impact prolonged activity or fatigue may have on knee movement patterns, it is important to induce a state of physical stress on the body to better understand mechanics. When attempting to better understand the reasons behind why ACL injuries occur, it is important to get an understanding of what happens to the knee when fatigued as well, since the majority of ACL injuries occur when a person's body is physically tired. There appears to be a moderate amount of information/research studies investigating the effects of fatigue on jump-landing, muscle performance, and gender associated injury risk. However, this information is greatly varied in terms of fatigue protocol, types of jumps, and populations. Fatigue has resulted in many different changes in muscle activity, joint kinematics, ground reaction forces and jump performance.[20-22,38] Depending on the change, an increased risk for injury, particularly ACL injury, may be present.[37]

Quammen et al.[37] examined the effects of two different fatigue protocols on hip and knee kinetics and kinematics: a slow linear oxidative fatigue protocol (SLO-FP) and functional agility short-term fatigue protocol (FAST-FP). Power was calculated and determined to be 15 participants with an alpha at .05; 15 DI female soccer players with clearance from the team physician to practice were tested. Kinematics was examined via 8 high-speed infrared cameras and ground reaction forces were sampled through 2 force plates. Only the dominant leg was sampled, and was determined as the leg the players used to kick the ball the furthest. For the SLO-FP group VO2 levels were analyzed via a metabolic cart.[37] Running—stop—jump and side stepping were used to mimic soccer moves and were randomly projected onto a screen in front of the players, which they then had to execute at a minimum speed of 3.5 m/s to, again, mimic a real soccer game. Five successful trials were completed followed by a 1-minute rest before their fatigue protocol was initiated.

Prior to the FAST-FP, participants had their maximal vertical jump measured 3 times and then averaged and used for the protocol. The first activity was the step-up step-down onto a 30cm box for 20 seconds at a 220 beats per minute pace (metronome to pace); next participants completed an L drill via 3 cones spaced 4.5 yards apart from each other and required participants to run down and back between cones 1 and 2, and then back to cone 2 and to run around the left side of cone 3, back to cone 2 and to finish sprinting to cone 1; participants then immediately completed 5 consecutive vertical jumps of at least 80% of their maximal jump, which was marked on the wall; finally participants completed a ladder drill with a metronome pacing them at 220 beats per minute—the first time participants faced forward through the ladder drill, the second circuit they faced their right moving laterally through the ladder drill and the last circuit they faced their left. All these 4 drills completed once counted as 1 set. Four sets were consecutively completed, which took about 5 minutes. The SLO-FP required participants to run at

9km/h for 5 minutes followed by a 1-km/h increase every 2 minutes until exhaustion. Participants expressed exhaustion through grabbing the railings of the treadmill, at which point the treadmill was slowed down. Maximal fatigue was noted by 2 of the following: (1) heart rate reached 90% of calculated age-related norm, (2) respiratory quotient was more than 1.1, (3) unable to continue running. Participants were then given a 5-minute rest before completing a 30-minute treadmill run alternating between 2 speeds for 6 intervals: 4 minutes at 70% of the final VO2 max speed, then 1 minute at 90% of final VO2 max speed. SLO-FP took approximately 45 minutes; 15 minutes for the VO2 max and 30 for the intervals. Immediately following the fatigue protocol, participants completed 5 successful randomized running—stop—jump tasks. FAST-FP participants were required to complete 3 maximal vertical jumps to maintain fatigue between dynamic trials.

Results, via a paired t-test, showed the FAST-FP group had a greater adduction moment at initial contact (IC). Otherwise both fatigue protocols showed participants had less hip flexion' post-fatigue than pre-fatigue at IC. At peak vertical ground reaction force, participants had less hip flexion post-fatigue than pre-fatigue. At peak vertical ground reaction force, participants had less knee flexion post-fatigue than pre-fatigue. Despite the FAST-FP displaying more frontal plane displacement of the hip and knee when compared to the SLO-FP, the authors concluded that both fatigue protocols achieved an increased risk for non-contact ACL mechanisms being integrated into the athlete’s motor outputs overall, thus increasing the risk of ACL tears.[37]

A study by Oliver et al.[20] looked at the effects of a soccer-specific fatigue protocol on jump performance and muscle activity. They measured jump height, GRF data, and EMG activity. The subjects consisted of 10 youth male soccer players with an average age of 15.8 years old. Three different types of jumps were included in this study: a squat jump, a countermovement jump, and a drop jump. The squat jump was performed from a starting squat position with knee flexed to about 90 degrees. The countermovement jump started from an erect position and when cued a maximal vertical height jump was performed. The drop jump consisted of dropping from a 35 cm height followed by a maximal vertical height jump. The subjects performed 3 trials of each jump with both feet on a single force plate and were done before and after the fatigue protocol. The 42-minute soccer-specific fatigue protocol was completed on a non-motorized treadmill. It consisted of 3 bouts of 14-minute exercise periods with 3 minutes of recovery in between each bout. The 14-minute bout was made up of seven 2-minute periods that included 45 seconds of walking, 15 seconds of cruising, 15 seconds of being stationary, 40 seconds of jogging and 5 seconds of maximal sprinting. Jump height and GRF data were recorded via the force plate and EMG activity was recorded using surface electrodes. Paired *t* tests were conducted to analyze differences in jump performance before and after the fatigue protocol, and a repeated-measures analysis of variance was conducted to analyze differences in jump conditions. Oliver et al.[20] found jump performance, as determined by jump height, was significantly lower after the fatigue protocol for all 3 jumps. There was no significant difference in jump performance between each of the different jumps. The only significant difference in EMG activity due to fatigue was during the drop jump. Specifically, they found significantly lower total EMG activity and lower muscle activity in the vastus lateralis, biceps femoris, and tibialis anterior, but not the soleus. When measuring impact, peak, mean, braking, and propulsive GRFs, the only significant difference due to fatigue was found in the impact GRF during the drop jump. This was significantly greater after the fatigue protocol. Overall, there were changes seen in jump performance across all jumps with fatigue, a decrease in muscle activity and an increase in impact force during the drop jump after the fatigue protocol.[20]

Moran et al.[38] investigated the effects of endurance fatigue and increasing heights on drop jump performance and knee mechanics. They studied impact acceleration, knee joint kinematics, and jump height before and after a fatigue protocol. Fifteen female competitive soccer players who had at least 6 months of experience with drop jumps as part of their training within the past 2 years were recruited. They performed 3 trials of drop jumps from each of the 15, 30, and 45 cm heights before and after an endurance-specific fatigue protocol. The participants were instructed to jump vertically with maximum effort while trying to spend as little time on the floor as possible.[38] The fatigue protocol consisted of running at a speed of 6 miles per hour with a 3% grade incline for one minute before increasing the incline by 1.5% every minute afterward. The subjects' rating of perceived exertion (RPE) was used to determine fatigue and was taken every 2 minutes during the protocol. On a scale of 6-20, an RPE of 17 or "very hard" was used to determine when fatigue had been achieved. Impact accelerations were measured using an accelerometer, knee motion was measured using an electrogoniometer, and jump height was calculated using a foot switch apparatus. In order to compare the fatigued state to the differing drop heights, a 2 x 3 within-subjects analysis of variance was conducted. Moran et al.[38] found significantly lower jump heights from all drop heights with fatigue. Due to this finding of lower jump heights, the authors concluded that neuromuscular fatigue was achieved with ratings of 17 or "very hard" on the RPE scale. They observed tibial impact accelerations significantly increased with each increasing height and they all were significantly larger with fatigue from heights of 15 and 30 cm, but not 45 cm. No significant changes were seen in knee flexion angles at initial contact, peak knee flexion or the range of knee flexion. However, peak knee angular velocity was found to be significantly larger when fatigued from heights of 15 and 30 cm, but not 45 cm. Peak knee angular velocity was also found to significantly increase with each increasing height. Therefore, the authors concluded that caution should be taken when performing drop jumps when fatigued due to the increased impact accelerations and therefore, greater risk of injury.[38]

McLean et al.[22] studied the effects of fatigue on hip, knee and ankle mechanics during drop jumps. This was done via a fatigue protocol. They also compared results between males vs females, and dominant vs nondominant legs within the study. Ten male and 10 female Division I athletes, taken from basketball, soccer or volleyball, were recruited as subjects. In order to be included in the study each participant needed to be free of any past ACL injury or any current lower extremity injury that would prevent them from participating in the drop jump or fatigue protocol. The drop jumps were performed by stepping off of a 50 cm platform and, after landing, immediately jumping vertically with the intent to achieve the maximum height possible and quickly as possible.[22] There were two separate force plates for each limb during landings. Ten trials of the drop jumps were completed before and after the fatigue protocol. The fatigue protocol lasted exactly 4 minutes and consisted of “a series of continuous drills that loosely reflected tasks synonymous with actual game play”.[22] The subjects were required to perform 20 step-up and step-down movements as quickly as possible onto a 20 cm step. Then they switched to plyometric bounding movements for a distance of 6 meters before turning around and bounding 6 more meters to the starting point. The subjects then went back to the step-up task and repeated this sequence as many times as possible in 4 minutes. Kinetic and kinematic data were recorded using force plates and a three-dimensional motion analysis system. To determine the effects of fatigue and gender, a three-way mixed-design analysis of covariance was conducted. McLean et al.[22] found fatigue produced significant increases peak knee abduction and peak knee internal rotation in both genders and legs. Fatigue also produced significant increases in peak knee abduction and internal rotation moments in both genders. The authors cited previous research that stated internal tibial rotation movement contributes directly to ACL loading and, in turn, an increased risk of ACL injury. Therefore, their results imply fatigue from their fatigue protocol increases ACL loading and risk of ACL injury. They also found that females had significantly greater increases in peak knee abduction and internal rotation, and had greater increases in peak knee abduction with fatigue compared to males. The authors suggested that this difference may contribute to females’ increased risk of ACL injury.[22]

A study by James et al.[21] investigated the effects of two different fatigue protocols on drop landing performance. They looked at kinetics, kinematics, and EMG activity during the landings. Ten recreationally active males were recruited to be subjects and were required to have no current lower extremity or spine injuries, or no past surgeries. They completed 10 trials of the drop landings prior to either of the fatigue protocols, and 5 trials were completed after each of the fatigue protocols. The drop landings were done from a platform 61 cm high and each subject "initiated each landing by slowly stepping out with their right foot, shifting their weight forward, then by quickly bringing the left foot forwards and dropping straight down".[21] The subject's right foot landed on a force plate with the left foot landing on the adjacent floor. Before any fatigue protocol, subjects participated in an initial testing session in order to determine general fitness and peak work rate on an ergometer. The first fatigue protocol consisted of isometric squat contractions, while the other consisted of cycling. The isometric fatigue protocol included repeated bouts of 15 second maximal isometric squat contractions followed by 5 seconds of rest until the subject's force output of their initial maximal contraction dropped below 50% for more than half of a 15 second contraction. They were then instructed to complete one additional 15 second maximal contraction. The cycling protocol included cycling on an "ergometer at a self-selected pace between 60 and 80 revolutions per minute at an intensity equivalent to 60% of the peak work rate achieved during the initial session".[21] Fatigue was achieved once the subject dropped below 40 revolutions per minute despite encouragement. Each fatigue protocol was completed 10 days to 6 months apart. GRFs were recorded via a force plate, knee motions were measured using an electric goniometer, and EMG data were recorded using surface electrodes. A 2 x 2 repeated measures analysis of variance was used to determine the effects of fatigue and fatigue protocol. James et al.[21] found significantly greater EMG activity in the vastus lateralis and vastus medialis with fatigue. They concluded that this supports the leg stiffness mechanism of fatigue which may contribute to ACL injury. However, they found significantly lower GRFs at the second peak and total force measures with fatigue, suggesting decreased leg stiffness. There was a trend towards greater EMG activity with fatigue, but this was not significant. The authors also found significantly greater knee flexion at contact after the isometric fatigue protocol compared to the cycling protocol. The isometric protocol was meant to induce short duration isolated fatigue, whereas the cycling protocol was aimed to bring about whole-body fatigue that persisted into recovery. The knee flexion finding suggests that the fatigue protocol has an impact on changes in kinematics and highlights the importance that fatigue must be achieved in order to accurately study changes in kinematics.[21]

Although many ACL injuries occur during jumping and cutting, it is also well known that these injuries can often happen while in the deceleration or eccentric portion of a movement. Research was previously done by Zebis et al.[19] to examine the biomechanical and muscular effects of acute fatigue on the lower extremity (LE- hip, knee, and ankle) of healthy subjects. In order to do this, Zebis et al.[19] examined the impact of acute fatigue on neuromuscular activity in female handball players. Prior to fatigue participant's maximal isometric voluntary contractions of the quadriceps and hamstrings were taken, but the researchers noted this has little functional applicability to

ACL tears. Participants also had their quadriceps/hamstring EMG activity measured while performing a side-cutting maneuver over a force plate, which the researchers surmised would be more applicable to moments of ACL trauma. A fatigue protocol, lasting 50 minutes, mimicking a handball match with various segmented tasks was performed. Fatigue was based on heart rates collected from 4 handball athletes during a match, which aimed to match those heart rates with their fatigue protocol. The fatigue protocol consisted of low/high intensity running/sprinting with different levels of incline, countermovement jumps (CMJ), single and double leg jumps, side-cutting, and sidestepping. Post fatigue rate of perceived exertion (RPE) was measured at an average of 16±1, which rates “hard” to “very hard” on the 20 point Borg scale.[19]

Zebis et al.[19] found that maximal voluntary contraction (MVC) of the quadriceps and hamstrings decreased significantly, but did not have statistically different EMG results when compared against one another. No statistical differences were found in knee and hip joint angles or ground reaction forces (GRF); however, they did find significant differences in neuromuscular activity in the pre-landing and landing phases of side-cutting (with $p<.05$). Specifically they noted decreased activity in the biceps femoris, semitendinosus, and lateral gastrocnemius at 10 ms and 50 ms prior to landing. In the 10 ms after landing, biceps femoris and semitendinosus were again decreased, but 50 ms after landing no statistically significant decreases were noted in the observed muscles (vastus lateralis/medialis, rectus femoris, gluteus medius, biceps femoris, semitendinosus, gastrocnemius lateralis/medialis).[19]

The findings show that ACL-agonist muscles experience a marked inhibition after acute systemic fatigue. Zebis et al.[19] hypothesize that the unexpected finding of reduced sub-maximal neuromuscular activation of the hamstrings may be in part due to the increased need to recruit more motor units to resist the quadriceps counter force, thus being a totally different motor pattern and/or reduced synchronization of hamstring activation. Likewise it could also be a purposeful decrease in hamstring activity to help stabilize the knee joint angles and GRFs, which were not noticed to have altered in this study. Although this may be advantageous for stabilizing knee angles and GRFs, it is not an ACL protective strategy and has been shown to put the knee at greater risk for a valgus moment or hyperextension in non-contact explosive moments like side-cutting, jumping or landing. Limitations of this study include the fact that joint angles were not measured with 3D kinematics.[19]

Another study looked at both EMG and MEPs (motor evoked potentials) of males and females pre and post exercise.[35] The exercise consisted of treadmill walking at increasing incline, step-ups, and body squats. This study reported that males had increased torque of the knee extensors and higher MEP amplitudes of the hamstrings than females pre-exercise., Males exhibited greater vastus lateralis activation and a decreased knee-extension torque when compared to females post-exercise. This suggests that there is a potential loss of quadriceps torque after exercise, particularly in females, which can indicate a difference in how each gender reacts to fatigue. The researchers suggest that this could potentially create an increased risk of knee injuries in females, similar to the previous study.[35]

Another study that compared males and females, as well as dancers to team-sport athletes was completed by Liederbach et al.[39] This study was conducted with the purpose of comparing 40 dancers (20 males and 20 females) to 40-team sport athletes' (20 males and 20 females, D1-3) single-leg landing mechanics after going through a fatigue protocol. The protocol consisted of 50 step-ups onto a 30 cm box (leading with right leg) and 15 max effort vertical jumps. If vertical jump height decreased by 10% post fatigue, landings were assessed via 3D kinematics and kinetics. The importance of comparing dancers to team athletes was multifaceted: 1) dancers suffer less ACL injuries than team athletes even though single-leg landing is common throughout performances and 2) there is no sex disparity among ACL noncontact tears in dancers.[39]

Results showed that dancers took longer to reach a fatigued state, but no other group interactions appeared statistically significant.[39] A MANOVA showed female dancers landed with a significantly lower knee valgus angle, hip adduction moment, and trunk side flexion than female team sport athletes. However, there was no interaction with fatigue and sex or group. Post hoc testing showed increased trunk flexion and lateral trunk lean in a fatigued state of both the dancer group and athlete group. Quadriceps dominance, defined as "preferential use of the quadriceps muscles to stiffen and stabilize the knee joint during landings", showed with group x fatigue, and main effects of group and fatigue separately. Fatigue among both groups globally increased peak knee flexion angles and decreased knee flexion moments.[39]

As shown above, proper knee biomechanics and kinematics cannot be assumed to carry over post-fatigue even if they are exhibited in non-fatigued state. When an ACL injury is involved we know improper and unnatural mechanics have occurred to stress and strain the ligament to the point of rupture. Often times fatigue is a large contributor to these faulty mechanics.

Lower Extremity Biomechanics of Subjects with Anterior Cruciate Ligament Deficiency or Reconstruction

Injuries to the ACL are typically surgically repaired.[11] In much of the literature it appears that patients who have undergone ACL reconstruction (ACLR) continue to suffer deficits. These include deficits of knee function, postural instability, multi-planar hip and knee kinematic deficiencies, altered biomechanics with gait and dynamic landings, and kinetic asymmetries. Studies by Paterno et al.[11] and Miranda et al.[17] found reduced peak vertical ground reaction forces (VGRFs) during the landing phase of a drop vertical jump (DVJ) and a jump cut maneuver for those who had undergone ACLR when compared to the contralateral limb and to controls. In the Paterno et al.[11] study 56 subjects who had a unilateral ACLR, along with 42 healthy, activity-matched control participants, performed a DVJ maneuver from a 31-cm box. The subject was instructed to drop off the box with both feet leaving the box simultaneously and each foot landing on a separate force platform. Because athletes frequently injure their ACL when landing from a jump, the landing phase of the DVJ maneuver was used for analysis. Over three successful trials, the mean vertical ground reaction force (VGRF) was found for each subject. A three way, 2x2 analysis of variance was used to compare the differences in VGRFs between the involved and uninvolved leg of the group with ACLR, between the ACLR group and the control group, and between the sexes. While there was no significant effect of sex noted, the involved limb of the ACLR group showed significantly lower VGRF than their uninvolved limb and both the preferred and non-preferred limb of the control group.[11] The Miranda et al.[17] study included 10 healthy subjects and 10 subjects who had undergone ACLR at least five years prior. Each subject was asked to perform a jump-cut maneuver. Standing one meter from a force plate, subjects jumped forward onto the force plate after hearing the verbal command "Go". At the same time of the verbal command, a visual prompt for direction to the right or left was also given. In the direction of the visual prompt, the subject performed a sidestep off of the force plate at a 45 degree angle. Subjects had to perform 10 correctly executed trials. VGRFs were calculated based upon the average of these 10 trials. Using two-way analysis of variance, comparisons between gender and ACLR status were made. All subjects, regardless of sex, with ACLR landed with decreased VGRFs.[17] These findings are in support of other studies documenting compensatory patterns of load transference and force absorption after ACLR.[11,17]

Delahunt et al.[16] completed a study with 14 female athletes who had undergone ACLR and 17 age- and sex-matched healthy controls. Each subject was required to complete the International Knee Documentation Committee (IKDC) Subjective Knee Form and the IKDC Subjective Knee Form, Knee Injury and Osteoarthritis Outcome Score (KOOS). The authors reported a statistically significant difference, with large effect sizes on the IKDC Subjective Knee Form, KOOSpain, KOOSsymptoms, KOOSsport, and KOOSqol scales for those who had undergone ACLR. Each subject then completed three trials of the anterior, posterior-medial, and posterior-lateral directional components of the Star Excursion Balance Test (SEBT). The subject began the test by standing barefoot on two adjacent force plates. Each trial was initiated when the subject transitioned from double-leg to single-leg stance and ended when the subject returned to the double-leg stance position. A 1.5m measuring tape was used to measure the reach distances in each direction. Postural stability in the posterior medial and posterior lateral directions on the SEBT were significantly impaired for those who had undergone ACL reconstruction. These differences in postural instability did not hold true for the anterior reach direction. Delahunt et al.[16] further classified multiplanar hip and knee joint kinematic deficiencies in their ACLR group when compared to a control group. These deficiencies include greater hip adduction, less hip flexion, less knee flexion and increased hip internal rotation with tasks of the SEBT.[16] Similarly Webster and Feller[40] found reduced knee flexion and less maximal internal rotation of the knee during single limb hop and drop landings. In this study 35 ACLR patients and 13 healthy subjects performed two functional landing tasks: a one legged horizontal hop and a one legged vertical drop landing. The single leg hop consisted of a horizontal hop from a distance equal to the subject's leg length. The single-leg drop landing occurred from a 15 cm high wooden platform. Three practice trials were performed on each leg for both tasks. Significantly reduced knee flexion was documented for the ACLR patients and on the contralateral/non-involved limb. For both landing tasks the patients also displayed less maximal internal rotation .[40] It is hypothesized that these significant alterations in joint kinematics and loading patterns put individuals following ACL-reconstruction at a high risk for reinjury of the ipsilateral or contralateral ACL. Rates of re-injury have been reported as high as 27% in the literature.[14,15]

Another study by Paterno et al.[41] found 4 variables combined to predict re-injury. The study included 56 athletes who had undergone ACLR. Before return to sport each subject underwent a 3-dimensional motion analysis during a DVJ maneuver and postural stability assessment. These athletes were followed for an additional 12 months to record occurrence of a second ACL injury. With a force plate under each foot, subjects were asked to complete three successful trials of DVJ maneuver from a 31-cm box; VGRFs and kinematic data were calculated for the landing phase of the DVJ. Each subject further underwent single limb, dynamic postural stability assessed on both their involved and uninvolved limb. Positioned on a single leg in the middle of a dynamic, unstable platform, the subject was instructed to maintain a stable position for 20 seconds, 3 times on each limb. The Balance System recorded movement of the platform away from a level position. Three of the predictive biomechanical measures occurred during landing of the DVJ task; these included: decreased hip external rotation in the uninvolved limb, asymmetries in the sagittal plane knee moments of flexion, and increases in the frontal plane knee joint range of motion of abduction. Subjects who displayed deficits in single-leg postural stability of the involved limb also demonstrated higher rates of second injury.[41]

In contrast to most of the studies reviewed, Flanagan et al.[42] found no deficits in force production or reactive strength capabilities of those who had been well rehabilitated (as determined by their score on the International Knee Committed Subjective Knee

Evaluation Form and with functional performance test) postoperatively. This study utilized 10 athletes who had returned to sport following ACLR and 10 age- and activity-matched control subjects. Subjects completed three trials on each leg of two functional performance tests: single-leg hop for distance and the 6-m timed hop. The participants' best score for each leg in each test was used to calculate the leg symmetry index. Utilizing a force sledge apparatus, each participant performed 4 testing protocols for which the subject performed three jumps on each leg: the squat jump, countermovement jump, drop jump, and rebound jump. For the squat jump, the participants started at a position of 90 degrees of knee flexion; they were then instructed to "drive themselves into the air" with maximal effort. The countermovement jump had the participant starting in a position of full knee extension. The subjects were then instructed to jump as high as possible. During the drop jump subjects were dropped from a height of .3m above the force plate, along the sledge's inclined track. Subjects were instructed to land with legs toward extension and to jump rapidly off of the force plate. The final jump, the rebound jump, was similar to the drop jump. Subjects were instructed to perform four maximal jumps in quick succession after dropping from the same .3-m height. Statistical analysis revealed no differences in the IKDC scores of the ACLR and control groups, as well as no difference in leg symmetry index between the groups. The force sledge apparatus testing revealed comparable degrees of between leg difference in both the ACLR and control groups; however no differences between groups were noted.[42]

Another study by Ortiz et al.[18] looked at differences of LE mechanics between non-injured females and females with ACL reconstruction, as well as differences between involved and noninvolved legs of females who have undergone ACL reconstruction. They looked for any differences that occurred during a single-leg drop jump and a hop task. Thirteen physically active females with ACL reconstruction were recruited. On average, they were 7.2 years post-ACL reconstruction and ranged from 1 to 16 years post-reconstruction. Fifteen healthy, non-injured females were recruited for the control group. All subjects performed 5 trials of the single-leg drop jump from a height of 40 cm, and 2 trials of 10 single-leg up-down hops to and from a height of 20 cm. The drop jump was done to mimic a land-and-go maneuver which is commonly performed in sports.[18] Leg dominance was determined based off of individual preference for a single-leg hop for distance. Knee and hip angles, knee moments, GRFs, and EMG data were recorded via surface electrodes, motion analysis cameras and force plates.[18] Three different multivariate analyses were conducted to compare kinematics, kinetics and EMG data between reconstructed and non-injured legs. Paired *t* tests with Bonferroni correction were conducted on the same variables comparing involved and noninvolved legs of subjects with ACL reconstruction. For the drop jump, the authors found no differences in knee and hip angles between non-injured and reconstructed legs.[18] Significantly lower peak anterior-posterior shear forces were found in reconstructed legs and significantly greater knee extension and valgus moments were found in reconstructed legs. For up-down hops, no differences were found in knee and hip angles, joint kinetics or EMG data.[18] However, a significant difference was found between involved and noninvolved legs in the ACL reconstruction group. Peak knee extension moments were greater in noninvolved legs compared to involved legs. Ortiz et al.[18] have shown that women with ACL reconstruction demonstrate different lower limb landing techniques when compared to the lower limbs of healthy, non-injured women during a single-leg drop jump. However, in a task with lower eccentric and rotational loads, the single-leg up-down hop task, those with ACL reconstruction demonstrate similar landing techniques to those who have not been injured. They have also shown that women with ACL reconstruction demonstrate closely symmetrical mechanics between involved and noninvolved legs during a drop jump and up-down hop task.[18]

Lower Extremity EMG of Subjects with Anterior Cruciate Ligament Deficiency or Reconstruction

As previously mentioned, a study recorded EMG data while looking at LE biomechanical differences between non-injured females and females with ACL reconstruction, as well as differences between involved and noninvolved legs of females with ACL reconstruction.[18] This study consisted of observing single-leg drop jumps and an up-down hop task. The results showed significantly greater co-contraction ratios between the quadriceps and hamstrings, greater gluteus maximus activation, and greater rectus femoris activation in those who had ACL reconstruction compared to those who did not. This suggests that women who had ACL reconstruction surgery may be predisposed to future injury as their landing mechanics differ slightly from their non-injured counterparts. During the drop jump, there were no significant differences between involved and noninvolved legs in subjects with ACL reconstruction. For the up-down hop task, no differences were found relating to EMG data for either comparison.[18]

Effects of Fatigue on the Lower Extremity of Subjects with Anterior Cruciate Ligament Reconstruction

Fatigue plays a dramatic role in biomechanical effects of the knee not only in healthy subjects, but also in subjects following an ACL reconstructive surgery. Webster et al.[23] had 11 healthy men as control subjects and 15 men who had undergone primary ACL reconstruction (ACLR) 15-19 months previously complete a general fatigue protocol consisting of 10 body weight bilateral squats, 2 vertical jumps, 10 single-leg drop landings (5 on each leg). This cycle was repeated 5 times or until fatigue—which was stopped when jump height was reduced by 20%, or subjects could not complete the protocol.[23] Pre and post fatigue, subjects performed single-leg landings from a 30 cm platform which were analyzed with 3D kinematics using 10 infrared cameras to see if fatigue would affect landing mechanics, both among groups and also between surgical leg and contralateral leg.[23]

The findings showed no significant differences between subjects with an ACLR or the control subjects.[23] Although, the ACLR group had a decrease in peak hip flexion and ankle dorsiflexion as well as an increase in hip abduction and knee abduction and internal rotation (IR) on the operative side when compared to the control group. However, there were no significant interactions between fatigue level and group/limb for any kinematic variable. Fatigue did impact the kinetic component of both groups; decreases in knee flexion and adduction moments were noted, showing that individuals are at a greater risk of tearing both ACLs regardless or ACLR history. Smaller knee moments were noted in the ACLR limb compared to the contralateral limb as well as an increased hip flexion moment on the operated limb compared to a decrease in the control group. Webster et al. hypothesized that this was evident of compensatory strategies occurring to help preserve lower limb stability.[23]

Frank et al.[24] studied the effects of neuromuscular fatigue on landing mechanics in active females with ACLR. The mechanics of fourteen physically active subjects between the ages of 18 and 30 were assessed by 3D analysis when jumping off a 30 cm box onto a force plate and then jumping as high as possible and landing on both legs. Physical activity was defined as being active 3x/week for at least 30 minutes and rated on the Marx scale (a self-report measure of frequency of cutting, running, jumping, etc. activity). Mechanics were assessed by 3D analysis when jumping off a 30 cm box onto a force plate and then jumping as high as possible and landing on both legs. Single-leg balance was tested pre and post with eyes closed on a force plate. Fatigue protocol was completed with squats from 0-60˚ at the rate of 25/minute with metronome pacing and the barbell loaded at 30% of participant's body weight. Participants went through 2 sequential squat cycles and rated their exhaustion on the Borg scale.[24] Results showed significantly decreased hip flexion, but no other statistical significances of kinematics at the hip or knee were found.[24] No kinetics were statistically different. Center of pressure, as a measure of single-leg balance through sway speed (cm/s), significantly increased post fatigue, which showed an increase in dynamic single leg stance. This study showed that ACLR individuals are at an increased risk of injury when exposed to fatiguing activity as was evident by less hip flexion at IC, which points to a reduction in muscular resistance to fatigue and the neuromuscular system's ability to sustain quality landing patterns after fatigue.

As seen in many of the previous studies, there is a high amount of variability between fatigue protocols, types of jumps, and recorded measures. It is often difficult to see similarities in the presented research due to this inconsistency.

Landing Error Scoring System (LESS) and LESS-Real Time (LESS-RT)

Padua et al.[43] created the Landing Error Scoring System (LESS) in order to provide an "inexpensive clinical assessment tool … to provide a standardized instrument … for identifying potentially high-risk movement patterns during a jump-landing maneuver."[43] This assessment tool consists of 17 scored items, evaluating the positions of the lower extremities and trunk at different times of the landing. The higher the score, the worse the landing technique. The jump-landings are recorded using two cameras viewing the sagittal plane and the frontal plane, and are later viewed for scoring. The jump-landing maneuver consists of a subject jumping forward off of a 30 cm box to a distance of 50% of their height away from the box, and immediately rebounding for a maximal vertical jump upon landing.[43]

Padua et al.[43] recruited 2691 males and females from 3 different military academies to be included in their study. At the time of testing, participants needed to be healthy and free of orthopedic injury. They tested the concurrent validity of the LESS by comparing it to the "gold standard" of a 2-camera three-dimensional motion analysis system.[43] They also tested the interrater and intrarater reliability via intraclass correlation coefficient (ICC). The authors found significant differences in biomechanics and ground-reaction forces between those who scored high on the LESS and those who scored low on the LESS. Those who scored high demonstrated biomechanics that have been shown to be related to ACL loading and injury mechanisms. Therefore, the LESS was considered to be valid with regards to assessing poor landing biomechanics during a jump-landing. Interrater reliability was found to be “good”, ICC = 0.84. Intrarater reliability was found to be “excellent”, ICC = 0.91. The authors also concluded the LESS was sensitive due to standard error of measure (SEM) values below one. Low SEM values represent a low estimate of standard deviation or variation in sample scores, meaning little variation in LESS scores. The authors noted that although the LESS may assess poor landing mechanics, it is uncertain if it predicts risk for ACL injury because this was not investigated in this study. [43]

Onate et al.[44] tested the interrater reliability between expert and novice raters and criterion validity of the LESS. They recruited 19 females who played soccer for a Division I college to perform the LESS and be observed by a three-dimensional motion analysis system. At the time of testing, all participants were free of low back or lower extremity injury within the past 6 months or surgeries within the past 2 years. The two raters of the LESS were both certified athletic trainers (ATCs). The expert rater of the LESS had 15 years of experience as an ATC and 5 years of experience using the LESS. The novice rater had less than 1 year of experience being an ATC and no experience using the LESS, but received a one-hour training session. Overall, there was “excellent” interrater reliability between expert and novice scores, ICC = 0.835. Within nine different items of the LESS, Kappa values ranged from 0.459 – 1.0. Therefore, there is excellent expert vs novice interrater reliability of the LESS. To assess validity of the LESS, it was compared to a three-dimensional motion analysis system. If found to be valid, the LESS would be able to accurately assess motion patterns similar to that of the 3-D system. Onate et al.[44] concluded the LESS has a "moderate to excellent level of validity," but it is item dependent. Percent agreement was used to determine validity. Ankle flexion at initial contact, knee flexion range of motion, trunk flexion at maximal knee flexion, foot position at initial contact, and when stance width was greater than shoulder width had "excellent" percent agreement. Trunk flexion at initial contact, when stance width was less than shoulder width, knee valgus at initial contact, and knee valgus range of motion had "moderate" percent agreement. Knee flexion at initial contact, initial foot contact, and lateral trunk flexion at initial contact had "poor" percent agreement. Onate et al.[44] concluded that the LESS was "an ideally suited component of any baseline examination for developing prevention programs associated with reducing lower extremity injury" and that their "findings support moderate to excellent validity to accurately assess three-dimensional kinematic motion patterns".

Padua et al.[45] created a different version of the LESS, the Landing Error Scoring System – Real Time or LESS-RT. This was done in order for assessment of landings to be done in real time instead of using cameras, making it more time efficient for clinicians. With the jump-landing performed in the same way, the LESS-RT consists of 10 scored items over four trials of jump-landing, but it is observed and scored in real time. The authors tested the interrater reliability of the LESS-RT. They recruited 43 healthy males and females free from any injury or illness to participate in their study. Three raters evaluated the participants using the LESS-RT. All of the raters were ATCs with over 5 years of experience and previous training and experience with the LESS. The authors found ICC ranging from 0.72 – 0.81 and concluded “good” interrater reliability. Padua et al.[45] noted that they did not investigate the validity of the LESS-RT and therefore it is unknown if the LESS-RT shares the same validity as the LESS.

CHAPTER III

MATERIALS AND METHODS

Subjects

A convenience sample of 20 healthy subjects volunteered for the study (26.3 ±3.7 years; 159.6 ± 27.3 pounds), 10 female (25 ± 2.7 years; 138.8 ± 17.9 pounds) and 10 male (27.6 ± 3.9 years; 180.4 ± 15.3 pounds). When the participants arrived, a consent form was administered and signed by each individual prior to the assessment. A subject intake form was then completed which included the participant's age, activities, types of competitive sports participated in, history of illness, and leg dominance. Leg dominance was determined by asking which leg the participant would kick a ball with. The subject's height, weight, and waist circumference were also taken by the examiner. BMI was calculated using the patient's height and weight. The following exclusions prevented a subject from participation: a recent history of orthopedic injury, a positive response to the PAR-Q & You form, or if the participant did not consider themselves healthy enough to participate given the demands of the tasks. This study had been approved by the St. Catherine University Institutional Review Board prior to data collection.

Instrumentation

Three-dimensional joint kinematics were measured using Ascension's Flock of Birds electromagnetic motion capture system (Ascension Technology Corporation, Burlington, VT) and Motion Monitor Software (Innovative Training Sports Inc., Chicago

IL). Two Bertec force plates (Bertec Corporation, Columbus, OH) placed side to side were linked to the Motion Monitor system through an A/D interface panel (Measurement Computing's PCIM 1602 – 16 bit PCI board) for measurement of ground reaction forces (GRF). Sensors measured 19.8 mm x 7.9 mm, which allowed precise placement over the boney segments to be analyzed. The electromagnetic sensors were placed on the participant's sacrum, lateral thighs, and proximal lateral shanks. To control for interrater error, only one researcher identified the placement of the sensors and marked the sensor position with a pen so they could easily be placed on the lower extremity (LE) following the fatigue protocol. Each sensor has an orthogonal axis system embedded within it and is capable of an independent sampling rate of 100 Hz. The Flock of Birds system has a reported static positional accuracy of 0.3 inch root mean-square (RMS) within a five foot range from the transmitter and 0.6 in RMS within a 10 foot range. Static angular accuracy is 0.5 RMS within a five foot range and 1.0 RMS within 10 feet (Ascension Technology Corporation, Burlington, VT). The reliability and validity of electromagnetic motion capture systems in gathering 3-D movements has been previously documented.[28,46,47]

Procedures

Kinematic Assessment

For kinematic assessment using the Motion Monitor integrated system (Innovative Sports Training, Inc., Chicago, IL) with Ascension's Position Capture Technology, electromagnetic sensors were affixed to the skin with double-sided Velcro and adhesive tape to sacral level two. Four other sensors were attached to each distal lateral thigh near the iliotibial band (ITB) to avoid excessive movement, and to the mid shank of each tibia. These sensors were secured with athletic pre-wrap and Velcro straps.

Anatomic bony landmarks on the pelvis, thigh, and shank were digitized for data capture of lower extremity movement using International Society of Biomechanics (ISB) recommendations for the hip and ankle and Grood and Suntay recommendations for the knee.[48,49,50] The Leardini method was used to determine the location of hip joint centers.[31] A minimum of 5 different static positions per leg were used to estimate hip joint center with a maximum error rate of 0.01. If an error rate above 0.01 occurred, the sensors were rechecked and digitization was repeated until it dropped below. The global reference system was defined using the right hand rule for all body segments with the positive x-axis defined as the posterior to anterior axis, the positive y-axis defined as the inferior to superior longitudinal axis, and the positive z-axis as medial to lateral. The Euler angle sequence was ZY'X".

With the subjects standing with their arms relaxed at their sides, a minimum of three landmarks per segment were palpated on the lower extremities and digitized with the stylus. Bony landmarks on the femur, tibia and fibula were palpated and digitized for transformation of sensor data to the local anatomic coordinate system.[49,50] Sensor landmarks are shown in Figures 1 and 2. Digitization of the bony landmarks allowed the receiver data to be transformed from a global coordinate system to a local anatomic coordinate system embedded in each of the bony segments. Local anatomic axes systems are advantageous to compare segments to one another, allowing a clinically meaningful comparison as opposed to the use of a global reference system. The local coordinate system of the thigh was set up using the medial/lateral femoral epicondyles.[49] The local coordinate system for the shank was determined using medial and lateral malleoli and medial/lateral joint lines.[49] Data was captured at 100 Hz and low pass filtered at 30 Hz using a Butterworth 4th order zero phase shift filter (optimal filtering for smoothing of data without losing too many points, as noted with trials of different filter levels). Force plate data was sampled at 1000 Hz, which has a default set up with an analog anti-aliasing filter of 500 Hz. The force plate was calibrated for each subject prior to testing. No other software filtering occurred, as this was sufficient.

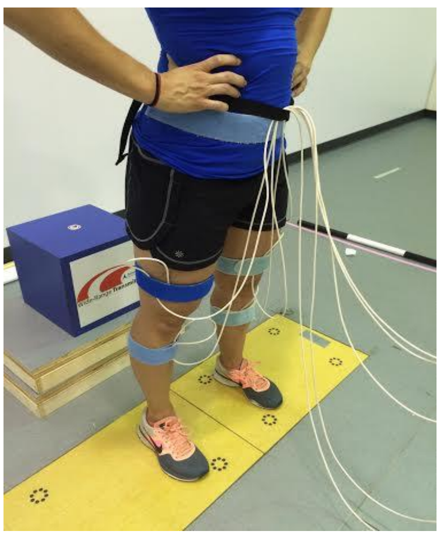

Figure 1. Sensor placement on the distal, lateral thigh and mid shank.

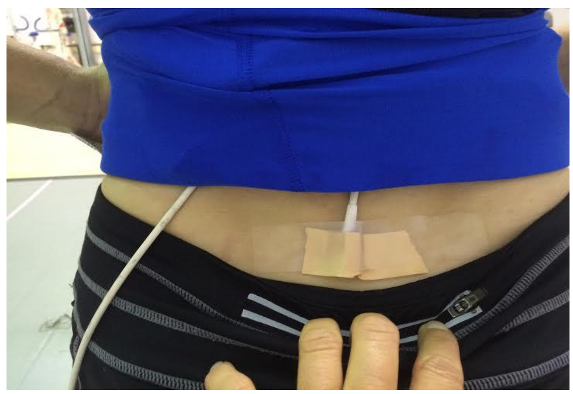

Figure 2. Sensor placement on the sacrum.

Prior to testing, demographic information was collected from the subjects. Maximal jump height was found by having the participant complete a bilateral (BIL) standing vertical jump next to a wall. Jump height was measured against a wall with the middle finger on the participant's right hand being used to mark reach without jump and peak jump height. Three jumps were completed and the average was taken. The participants then completed a maximum of five bilateral drop jump vertical jumps which was scored by the researchers using the Landing Error Scoring System- Real Time (LESS-RT) protocol. The participants completed the jumps by bilaterally jumping forward from a 30 cm box and landing on a marker which was 50% of the participant's height away from the box. Jumps were redone if the participant failed to land on the marker or if the participant did not leave the box from both feet.

Task

Following the LESS-RT, the participants were marked for each sensor placement, digitization occurred, and the subject was allowed 3-5 practice jumps for each condition (BIL to BIL, BIL to single-leg right, and BIL to single-leg left). The subjects then completed the jumping protocol. This protocol consisted of the subjects completing three different jump maneuvers: BIL to single-leg (SL) on the right, BIL to SL on the left, and BIL to BIL. Jump order was chosen at random prior to the testing session. Participants positioned themselves with one foot on each force plate prior to each jump. Jumps were considered a success if the participant was able to land with one or both feet (depending on jump type) completely on the respective force plate (i.e. SL jump right would require the right foot of the subject to land within the right force plate area). The participants were instructed to jump as high as they can without looking down while trying to reach a dowel positioned above their head. An attempt would be considered a failure if the participant was unable to land on the respective force plate with their entire foot or if the participant lost their balance within the first two seconds of landing. Four successful jumps were completed in each direction. Following the initial data collection the fatigue protocol was completed. After the participant completed the entire fatigue protocol, RPE was taken to measure and ensure fatigue was reached, sensors were attached and digitized for some subjects on the participant's body, and the previous four jumps were completed in each direction. The re-digitization occurred in cases where excessive sensor slippage occurred and the computer animation displayed non-analyzable data.

Fatigue Protocol

The Functional Agility Short-Term Fatigue Protocol (FAST-FP) was utilized for this study.[37] The FAST-PF consisted of multiple agility exercises including: step-up onto a 30 cm-height box, 'L-drill', vertical jumps, and agility ladder drills.[37] One round consisted of all four agility drills and four total rounds were completed. Participants started by completing step-up movements onto the 30 cm-height box for 20 seconds to a metronome beat set at 200 bpm. Immediately afterwards, the participants completed one repetition of the 'L-drill.' The 'L-drill' involved three cones that were placed in the shape of an L (as seen in Figure 3). Each cone was 4.5 yards apart from one another. To complete the drill, the

L-Drill

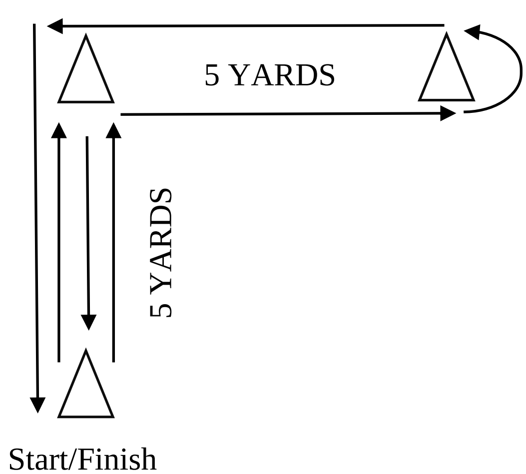

Figure 3. Diagram of the fatigue protocol L-Drill.

participant sprinted forward from the starting cone to the second cone. They immediately turned around and sprinted back to the starting cone. Then, the subject turned again and sprinted back to the second cone where they ran around it to cut to the left towards the third cone. The participants ran in a circle around the third cone from the inside to the outside, ran back to the second cone around the outside, and completed the sprint back to the first cone.[37] Immediately after the sprint, the participants completed five vertical jumps at 80% of their maximal jump height. Height was controlled by having the participants jump next to a wall and touch a marker on the wall at 80% of their maximal jump height. After jumping, the subject immediately progressed to completing the ladder agility drill. During the first and third rounds, the participants ran through the ladder (down and back) forwards with both feet touching the inside spaces of the ladder. During the second and fourth rounds the participant ran through the ladder sideways (facing opposite directions each round), placing both feet in each ladder space. During all four rounds a metronome was set to 200 bpm for the step ups and ladder drills. All four rounds of the fatigue protocol were completed with no rest breaks in between them.[37] RPE was used to determine the level of fatigue each participant was feeling at the end of the protocol (Mean = 8.4±0.75 using Modified Borg Scale). Immediately following the protocol, RPE was taken and the second set of jumps were completed using the Motion Monitor as described above.

Data Reduction

Raw kinematic data collected by the Motion Monitor system was selected for export and further processing. Each trial was viewed and data was cut to capture only the task requirements by a single investigator to ensure consistency. Original variables included in the export file were pre and post fatigue protocol Euler angles (*x, y, z*), normalized knee moments (*x, y, z*), and ground reaction force (*y*) for bilateral land, right leg land and left leg land. Pre and post fatigue conditions were also coded into a separate column for ANOVA analysis with all pertinent kinematic and kinetic data included per condition by variable to use pre and post fatigue conditions as a factor.

Statistics

A paired t-test was used to analyze significant differences between all subjects for pre and post fatigue conditions, as well as for each gender, in knee mechanics before and after the fatigue protocol for bilateral, and right and left landing trials. The independent variables were pre and post fatigue protocol, and landing style. The dependent variables included knee angles and moments in all 3 planes, and ground reaction forces. An independent t-test was also used to analyze significant differences between genders in knee mechanics following the fatigue protocol for each landing style. The independent variables were gender, while the dependent variables were knee angles, moments, and ground reaction force differences between pre and post fatigue protocol. Significance was set at a p-value of 0.05.

CHAPTER IV

RESULTS

Demographic data for participants is presented in Table 1.

Table 1. DEMOGRAPHICS OF PARTICIPANTS COMPARED BY GENDER

Variable	Male	Female
Age (years)	**27.6 ±4.1**	**26.1 ±8.5**
Height (inches)	**69.9 ±4.0**	**64.5 ±21.3**
Weight (pounds)	**180.4 ±16.2**	**165.2 ±54.7**
BMI (kg/m^2)	**26.2 ±3.4**	**23.1 ±7.3**
Waist circumference (inches)	**34.0 ±3.0**	**31.2 ±10.3**
Vertical Jump (inches)	**21.2 ±4.0**	**19.0 ±6.5**
RPE for Fatigue Protocol (0-10)	**8.4 ±1.0**	**7.7 ±2.5**
LESS-RT Score (0-15)	**7.6 ±3.2**	**7.8 ±3.2**

Data represented as: means±SD

All Subjects Pre-Post Fatigue

A number of significant differences were found in knee kinematic and kinetic values before and after the fatigue protocol for all subjects during the three types of landings (Table 2). All significant results had a p-value equal to or less than 0.05; a number had a p-value equal to or less than 0.01. Table 1 has the corresponding statistics. Knee flexion and internal rotation angles significantly increased after the fatigue protocol for left single leg landings, right single leg landings, and bilateral landings only in the right leg. Knee external rotation and adduction angles significantly increased for left single leg landings, right single leg landings, and bilateral landings in both legs. Knee abduction angles significantly increased for right single leg landings, and in both legs for bilateral landings. Flexion moments significantly increased after the fatigue protocol only for left single leg landings. External rotation moments significantly increased for both left and right single leg landings. Abduction moments significantly increased for right single leg landings, and bilateral landings in the right leg. Extension moments significantly decreased after the fatigue protocol only for right single leg landings. No significant differences were found for internal rotation moments, adduction moments, or ground reaction forces when comparing pre- and post-fatigue for all subjects.

Table 2. KNEE KINEMATICS AND KINETICS FOR ALL SUBJECTS PRE-POST FATIGUE

	Bilateral Landing				Left Landing		Right Landing	
	Left		Right					
Variable	**Pre**	**Post**	**Pre**	**Post**	**Pre**	**Post**	**Pre**	**Post**
Flexion Angle	-	-	**77.9±17.9**	**84.9±22****	**60.8±11.3**	**64.7±17***	**61±13.6**	**66.8±16.6****
IR Angle	-	-	**8.9±4**	**12.5±12.2***	**11±6.5**	**16±15.8***	**7.8±3.4**	**12.3±10.7****
ER Angle	**12.6±5.9**	**20.7±17****	**13.3±4.9**	**20.5±17.8****	**10±4.9**	**21.3±15.5****	**12±4.8**	**20.8±18****
ABD Angle	**11.8±10.5**	**16.9±14.3****	**16.3±9.4**	**21±17.6****	-	-	**15.1±8.3**	**17.8±11.1***
ADD Angle	**4.8±3.5**	**10.8±10.5****	**8.9±9**	**19.3±20.5****	**4.7±3**	**12±10.3****	**8.7±8.9**	**19.3±20.6****
Flexion Moment	-	-	-	-	**0.69±0.26**	**0.79±0.36***	-	-
ER Moment	-	-	-	-	**0.09±0.13**	**0.12±0.16***	**11.4±15.5**	**15.8±22.9***
ABD Moment	-	-	**0.78±0.45**	**0.96±0.73****	-	-	**0.63±0.61**	**0.85±0.72***
Extension Moment	-	-	-	-	-	-	**7.4±1.8**	**6.9±2.1***

Data represented as: means±SD, * = p<0.05, ** = p<0.01
IR = Internal Rotation, ER = External Rotation, ABD = Abduction, ADD = Adduction
Angles are measured in degrees, Moments are measured in Nm/kg

Males Pre-Post Fatigue

Significant differences were also found in kinematic and kinetic values pre- and post-fatigue for male subjects during all three landings (Table 3). Once again, all significant results had a p-value equal to or less than 0.05; a number had a p-value equal to or less than 0.01. Table 3 displays the corresponding statistic for each value. External rotation angles significantly increased post-fatigue for left single leg landings, right single leg landings, and bilateral landings in the left leg only. Abduction angles significantly increased in the left leg for bilateral landings. Adduction angles significantly increased for left single leg landings, and bilateral landings in the left leg. Abduction moments significantly increased for right single leg landings, and bilateral landings in the right leg only. No significant differences were found for flexion and internal rotation angles, flexion, internal rotation, external rotation, adduction and extension moments, or ground reaction forces.

Table 3. KNEE KINEMATICS AND KINETICS FOR MALES PRE-POST FATIGUE

	Bilateral Landing				Left Landing		Right Landing	
	Left		Right					
Variable	Pre	Post	Pre	Post	Pre	Post	Pre	Post
ER Angle	10.7±7	28.1±23**	-	-	9±4.8	29.6±18.7**	10.5±5.1	23.9±21.3**
ABD Angle	18.3±11.2	25.1±17.3*	-	-	-	-	-	-
ADD Angle	3.4±2.2	14.1±13.8**	-	-	3.5±1.8	15.2±13.6**	-	-
ABD Moment	-	-	0.62±0.25	0.9±0.8*	-	-	0.52±0.23	0.95±0.83**

Data represented as: means±SD, *=p<0.05, **=p<0.01
ER = External Rotation, ABD = Abduction, ADD = Adduction
Angles are measured in degrees, Moments are measured in Nm/kg

Females Pre-Post Fatigue

Numerous significant differences were found in kinematic and kinetic values before and after the fatigue protocol for female subjects during all three landings (Table 4). All significant results had a p-value equal to or less than 0.05; a number had a p-value equal to or less than 0.01. The corresponding statistics are displayed in Table 4. Flexion, external rotation and adduction angles significantly increased after the fatigue protocol for all landings and legs. Internal rotation angles significantly increased for right single leg landings, and bilateral landings in the right leg. Flexion moments significantly increased for left single leg landings, and bilateral landings in the right leg. External rotation moments significantly increased for left and right single leg landings. Internal rotation moments significantly decreased post-fatigue for left single leg landings, and bilateral landings in the left leg. Ground reaction forces within the knee were found to significantly decrease for right single leg landings. No significantly differences were found for abduction angles, or abduction, adduction and extension moments.

Table 4. KNEE KINEMATICS AND KINETICS FOR FEMALES PRE-POST FATIGUE

Variable	Bilateral Landing				Left Landing		Right Landing	
	Left		Right					
	Pre	Post	Pre	Post	Pre	Post	Pre	Post
Flexion Angle	80.8±15.9	86±17.6**	81.2±16.8	89.2±16.8**	60.8±9.9	64.1±13*	58.9±10.4	66.8±16.5**
IR Angle	-	-	8±3.6	11.5±10.5*	-	-	6.8±3.3	13.6±11.3**
ER Angle	6.6±6	10.3±6**	13.5±8	17.5±7.9**	10.7±5	15.9±10.2**	12.9±4.4	18.8±15.5*
ADD Angle	5.7±3.9	8.8±7.2*	7.3±3.8	17±17**	5.4±3.4	9.8±6.8**	6.7±4.3	15.9±17.1**
Flexion Moment	-	-	0.68±0.16	0.77±0.23*	0.53±0.18	0.68±0.35*	-	-
IR Moment	0.36±0.26	0.26±0.21**	-	-	0.5±0.45	0.4±0.32*	-	-
ER Moment	-	-	-	-	0.08±0.12	0.13±0.17*	8.6±13.1	12.8±19.3*
GRF	-	-	-	-	-	-	3175.9±698	2989±541**

Data represented as: means±SD, *=p<0.05, **=p<0.01
IR = Internal Rotation, ER = External Rotation, ADD = Adduction, GRF = Ground Reaction Forces
Angles are measured in degrees, Moments are measured in Nm/kg, GRFs are measured in N

Differences Between Gender Pre-Post Fatigue

Finally, significant knee kinematic and kinetic differences were found between genders following the fatigue protocol (Table 5). Significant values were found for all three landings. All significant results had a p-value equal to or less than 0.05; a number had a p-value equal to or less than 0.01. Table 4 displays the corresponding statistics. Males demonstrated significantly greater differences between pre- and post-fatigue with flexion, external rotation and adduction angles when compared to females. These greater differences occurred for bilateral landings in the left leg for flexion angles, for left single leg landings for external rotation angles, and for both left single leg landings and bilateral landings in the left leg for adduction angles. Males also showed significantly greater differences with internal rotation and adduction moments for bilateral landings in the left leg, and with abduction moments for right single leg landings. For ground reaction forces, males had significantly greater values compared to females for right single leg landings. No significant differences were found for the right leg during bilateral landings. Also, there were no significantly differences for internal rotation and abduction angles, or flexion, external rotation and extension moments.

Table 5. KNEE KINEMATIC AND KINETIC DIFFERENCES BETWEEN GENDER POST FATIGUE

	Bilateral Landing				Left Landing		Right Landing	
	Left		Right					
Variable	Male	Female	Male	Female	Male	Female	Male	Female
Flexion Angle	6.0	-5.2**	-	-	-	-	-	-
ER Angle	-	-	-	-	-13.8	-5.2*	-	-
ADD Angle	-10.6	-2**	-	-	-10	-3**	-	-
IR Moment	-0.05	0.1**	-	-	-	-	-	-
ABD Moment	-	-	-	-	-	-	-0.43	-0.03*
ADD Moment	3.9	2.2**	-	-	-	-	-	-
GRF	-	-	-	-	-	-	-87.8	186.2*

Data represented as: means±SD, *=p<0.05, **=p<0.01
ER = External Rotation, IR = Internal Rotation, ABD = Abduction, ADD = Adduction, GRF = Ground Reaction Forces
Angles are measured in degrees, Moments are measured in Nm/kg, GRFs are measured in N

CHAPTER V

DISCUSSION

Summary of Previous Research Findings

Extensive research has been conducted to determine if gender differences exist for landing mechanics, if fatigue has a direct impact on injury risk through analysis of VGRFs, hip/knee/ankle kinematic changes, jump height, etc., and a combination of both. Biomechanical gender differences do exist. Females demonstrate stiffer landing styles than their male counterparts. This single finding may have a profound impact on the ACL injury rates in healthy males versus females. Stiffer landings typically equate to increased ground reaction forces, which have been shown to increase rates of non-contact ACL tears.[28] Females also exhibit increased abductor moments and knee valgus, as well as decreased hip and knee flexion upon landing.[27,28, 30, 32,33,34]

Fatigue, in studies, is mimicked in different ways. Some studies choose endurance tasks such as running until failure is reached; others focus primarily on fatigue through muscular activity such as squats, while some try a combination by simulating gameplay as closely as possible by mixing endurance activities with plyometric movements to reach a full body fatigue before measuring lower extremity (LE) movement patterns.[19-22,30,36-39] Post-fatigue research with healthy subjects has observed decreased muscle activation of the quadriceps and hamstrings, decreased jump height, and changes in knee kinematics.[19-22] Some studies demonstrate that females have higher risk movement patterns with fatigue than males; however other studies show that both genders are equally impacted by fatigue.[39] One study compared dancers with other team sport athletes landing mechanics. They showed that dancers took longer to reach a fatigued state, but no other group interactions appeared statistically significant.[39] A MANOVA showed female dancers landed with a significantly lower knee valgus angle, hip adduction moment, and trunk side flexion than female team sport athletes. However, there was no interaction with fatigue and sex or group. Post hoc testing showed increased trunk flexion and lateral trunk lean in a fatigued state of both the dancer group and athlete group. Fatigue, among both groups, globally increased peak knee flexion angles and decreased knee flexion moments.[39] Another study reported that males had increased torque of the knee extensors and higher MEP amplitudes of the hamstrings than females pre-exercise. Post-exercise, males exhibited greater vastus lateralis activation and a decreased knee-extension torque when compared to females. Authors suggested that there is a potential loss of quadriceps torque after exercise, particularly in females, which suggests a difference in how each gender reacts to fatigue. The researchers suggest that this could potentially create an increased risk of knee injuries in females, similar to the previous study.[36]

James et al.[21] found significantly greater knee flexion at contact after the isometric fatigue protocol compared to their cycling protocol. The isometric protocol was meant to induce short duration isolated fatigue, whereas the cycling protocol was aimed to bring about whole-body fatigue that persisted into recovery. The knee flexion finding suggests that the fatigue protocol has an impact on changes in kinematics and highlights the importance that fatigue must be achieved in order to accurately study changes in kinematics.[21] McLean et al.[22] found fatigue produced significant increases peak knee abduction and peak knee internal rotation in both genders and legs. Fatigue also produced significant increases in peak knee abduction and internal rotation moments in both genders. The authors cited previous research that stated internal tibial rotation movement contributes directly to ACL loading and, in turn, an increased risk of ACL injury. Therefore, their results imply fatigue from their fatigue protocol increases ACL loading and risk of ACL injury. They also found that females had significantly greater increases in peak knee abduction and internal rotation, and had greater increases in peak knee abduction with fatigue compared to males. The authors suggested that this difference may contribute to females' increased risk of ACL injury.[22]

When measuring patients post ACLR, fatigue has shown to alter knee mechanics.[23,24] Webster et al.[23] found no significant differences between subjects with an ACLR or the control subjects. Although, the ACLR group saw a decrease in peak hip flexion and ankle dorsiflexion as well as an increase in hip abduction and knee abduction and internal rotation (IR) on the operative side when compared to the control group, there were no significant interactions between fatigue level and group/limb for any kinematic variable. Fatigue did impact the kinetic component of both groups; decreases in knee flexion and adduction moments were noted, showing that individuals are at a greater risk of tearing both ACLs regardless or ACLR history. Smaller knee moments were noted in the ACLR limb compared to the contralateral limb as well as an increased hip flexion moment on the operated limb compared to a decrease in the control group. Webster et al.[23] hypothesized that this was evident of compensatory strategies occurring to help preserve lower limb stability. Frank et al.[24] demonstrated that individuals with ACLR are at an increased risk of injury when exposed to fatiguing activity as evident by less hip flexion at IC, which points to a reduction in muscular resistance to fatigue and the neuromuscular system's ability to sustain quality landing patterns after fatigue.

This research aimed to reconcile the movement pattern and fatigue protocol with what is seen in sport to note differences between knee and hip kinematics and kinetics for healthy men and women. The research compared the lower extremity kinematics and kinetics of a jump-land task (double and single leg land) between healthy men and women after a fatigue protocol. Ultimately, this research aimed to identify whether a dysfunctional or high risk movement pattern existed which may predispose the subject (male and/or female) to possible knee or ACL injury. Our hypothesis was that all subjects will display changes in their movement patterns that would predispose them to knee injury. However, we expected that women, as compared to men, would exhibit more aberrant lower extremity kinematics and kinetics upon fatigue, further increasing their relative risk of ACL injury. It is well supported that females demonstrate stiffer landings characterized by increased ground reaction forces, decreased knee flexion angles, and increased motions that are consistent with knee valgus.[27,28, 30, 32,33,34] It was hypothesized that these differences would be more pronounced in the female population.

Impact of Fatigue on Subjects

All subjects demonstrated increased knee flexion, internal rotation, external rotation, adduction, and abduction angles post-fatigue protocol. Flexion, external rotation and abduction moments also significantly increased post-fatigue. Extension moments decreased after the fatigue protocol. There were no statistically significant changes in ground reaction forces post-fatigue in this study. These findings are not necessarily inconsistent with previous literature. Many studies support statistically significant changes in ground reaction forces when fatigued, but the literature is inconsistent.[20,21]

Some studies support increased ground reaction forces when fatigued, while others demonstrated decreased ground reaction forces post-fatigue.[20,21] Other literature supported no significant changes in ground reaction forces after a fatigue protocol, as seen in the current study.[19] This examination into the ground reaction force data between fatigue literature indicates the inconsistent findings across this body of research.

Males demonstrated increased external rotation, abduction, and adduction angles post-fatigue protocol. Additionally, abduction moments were increased. A study completed by McLean et al.[22] also found that males demonstrate increased abduction angles and moments when fatigued. Females demonstrated increased flexion, external rotation, adduction, and internal rotation angles. Flexion and extension moments increased while internal rotation moments and ground reaction forces decreased following the fatigue protocol. In opposition to this, McLean et al.[22] found increased internal rotation angles for female subjects with fatigue.

When comparing genders, the data supports that males demonstrate higher adduction, external rotation, and flexion angles. Internal rotation, adduction, abduction moments were also increased in the male population. Finally, males demonstrated greater ground reaction forces when fatigued. These findings contradict the previous literature on landing mechanics between males and females when the subjects are not fatigued.[27,28, 30,] [32,33,34] However, the post-fatigue landing literature is too varied at this point to accurately portray the typical landing changes that occur when a subject is fatigued.[22,36,39]

These results do not closely align with the hypothesis. An explanation of these findings is that men and women display decreased dynamic control for bilateral and single leg landings when fatigued, resulting in greater angles and/or moments overall. A relatively new body of research is examining how much variability is required to safely land on different surfaces without increasing injury risk. Movement variability was previously thought of as ‘noise’ that should decrease as an athlete becomes more proficient at a given task.[51-54] However, recent literature has supported that high level athletes have a certain amount of compensatory variability, while beginners have a more consistent pattern that is typically more rigid.[54] Given the results, it may suggest that males demonstrate increased variability that serves them well in sport, consistent with compensatory variability. Females may have a more consistent, yet rigid, pattern that may increase knee injury risk. Previous literature has supported this hypothesis, as females have demonstrated less hip and knee flexion with dynamic valgus at the knee during landing tasks when compared to males.[27,28,30,32-34]

Gender differences noted in our specific research population may have impacted the results. The first noted observation was a different level of athleticism and body awareness between the male and female subjects. The second observation arose from the data collection process. Participants were instructed to jump as high as they could and land in the center of the force plate. It was noted that males typically jumped higher and exerted greater effort than females, while the female participants typically required less trials to land in the center of the force plate.

Limitations

This study revealed several limitations. One of which includes a small sample size with a narrow age range. These two factors decrease the power of the results and reduce the ability to generalize our findings to other populations. Another limitation is the possible interference of skin motion artifact as bone pins were not utilized to more accurately measure joint angles. Given human involvement, there is the potential for error in regards to placement of sensors and validity of trials. Additionally, only some subjects were re-digitized after the fatigue protocol. Due to the need for re-digitization and multiple trials post-fatigue, there was increased recovery time, which may decrease the validity of the post-fatigue data. The nature of the study required the subject to land on force-plates, which increased the complexity of the task. This ultimately increased the number of trials and a provided a potential for change in jump and landing mechanics in order to land in the center of the force plate. Finally, hip and ankle mechanics were not analyzed in this study. Those joints may provide additional insight to the landing mechanisms that differ between men and women when fatigued. Further research is recommended, specifically with an increased sample size to increase the power of our results and to further evaluate the concepts of movement variability suggested in this study.

CHAPTER VI

CONCLUSION

This study supports that both gender and fatigue impact landing mechanics at the knee, which are consistent findings in previous literature. Furthermore, this study suggests that movement variability may influence landing patterns in men more than women. Further research is warranted to explore the relationship between fatigue and movement variability between genders when landing.

Our hypothesis was that all subjects would display negative changes in their movement patterns. However, we expected that women, as compared to men, would exhibit more aberrant lower extremity kinematics and kinetics upon fatigue, further increasing their relative risk of ACL injury.

The collected data did not necessarily support our hypothesis. Men exhibited more varied landing mechanics, while women's landing patterns tended to be more consistent and rigid. These findings pertain to a larger discussion surrounding movement variability. It has yet to be determined if and how much movement pattern variability post fatigue play in the reduction of risk for lower extremity injuries.

Future research should build on this study with a larger subject base. The relationship between landing mechanics and fatigue has yet to be determined. It is important to continue this body of research to understand the mechanics leading to increased injury ACL injury rates among female athletes.

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