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Effects of Basketball Exercise Simulation Test (BEST) On Landing Mechanics in Active Females

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EFFECTS OF BASKETBALL EXERCISE SIMULATION TEST (BEST) ON LANDING MECHANICS IN ACTIVE FEMALES

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ABSTRACT

Intro: In the United States, female basketball players have the second highest rate of ACL injury, following female soccer players. This injury rate is four times higher than their male counterparts. These injuries occur more frequently at the end of the first and second halves and may be associated with exercise-induced fatigue causing knee instability. The purpose of this pilot study was to determine the effects of a basketball simulation test on landing mechanics. **Methods:** Six subjects completed both the intervention and the control trial consisting of four 10-min quarter game simulations of the validated Basketball Exercise Simulation Test (BEST) or on a separate day the equivalent rest. There was Pre- and Post-testing including vertical jumps, single leg triple jumps, drop jumps, and drop cuts as well as abbreviated intermediate testing. Dartfish 2D video analysis was used to assess changes valgus and joint angles for drop jump mechanics and first step time for the drop cut movement. The data were analyzed by a one-way repeated measures analysis of variance for time and a two-way analysis of variance with repeated measures on both intervention and time. **Results:** Power analysis indicated the study was underpowered, and that 30 subjects were necessary for adequate statistical power; therefore, results were interpreted with $p \leq 0.1$ as significant. There were significant interaction findings for measures of RPE ($p=0.004$, $\eta 2 = 0.0057$), percent max heart rate ($p=0.011$, $\eta 2 = 0.0113$), and for the change in relative trunk angle during the take-off phase of the drop jump ($p=0.0073$, $\eta 2 = 0.36$). There was an

effect of time on vertical jump height ($p=0.0007$, $\eta 2 = 0.019$) and the single leg triple hop on the left leg ($p=0.01$, $\eta 2 = 0.28$). There was no intervention effect on vertical jump height or the single leg triple jumps, measures of fatigue, as a result of the BEST. With the exception of the take-off relative trunk angle, there were no significant changes in the trunk, knee, or ankle kinematics during the six different phases of the drop jump and there was no significant effect on time to first step on the drop cut ($p>0.05$, $\eta 2 < 0.1$).

Conclusion: The data collected indicate that BEST was sufficient to stimulate physical exertion based on RPE and heart rate. However, this exertion did not result in significant changes in vertical jump or single leg triple jump performance, measures for fatigue. Thus, the BEST was not capable of eliciting fatigue in this study. In addition, there was limited significant change in landing kinematics which suggests the BEST did not affect landing mechanics. This result suggests that the lower extremity load of a singular basketball match may not be sufficient for the production of fatigue or joint angle adaptations. Further research is needed to determine whether repeated bouts of exertion will affect those measures or to determine what other factors in addition to the lower extremity load could result in increased injury risk.

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LIST OF ABBREVIATIONS

ACL ....Anterior Cruciate Ligament

ATP ....Adenosine Triphosphate

BEST...Basketball Exercise Simulation Test

CDC..Centers of Disease, Control, and Prevention

EMG..Electromyography

MRI ....Magnetic Resonance Imaging

PCL.. Posterior Cruciate Ligament

RPE ....Rate of Perceived Exertion

CHAPTER 1

REVIEW OF LITERATURE

Introduction

Anterior cruciate ligament (ACL) injuries are prevalent worldwide. There are two primary classifications of ACL injuries: noncontact and contact. Those noncontact injuries account for 70 to 84 percent of the total recorded ACL injuries.[52,2] The nature of a noncontact ACL injury being primarily mechanism-based suggests preventable or modifiable factors. Thus, research has focused in this area since it is believed that identifying those factors will prevent injuries as well as other physical, mental, or monetary strain on individuals.

In most settings, these noncontact ACL injuries occur during sport game play. Compared to other countries, the United States has been associated with higher rates of ACL injuries. In part, this might be due to the greater participation and the structure of sport in American culture as well as the prevalence of sports that utilize sport movements such as cutting and drop landings. In particular, soccer and basketball have been found to have the highest ACL injury rates in the United States, in that order.

Epidemiological studies have found that there are populations within those two sports that have further increased risk. The overall findings suggest that college-aged females during game play are at highest risk. Females can have up to a four to eight-time higher risk of injury compared to their male counterparts. While increased knee

valgus and anterior tibial translation has been associated with increased ACL injury in both male and female athletes, anatomical, hormonal, biomechanical, and neuromuscular differences have been investigated to explain or further understand reasons for this difference. Another potential factor that has been suggested to play a role in increased injury risk is fatigue.

Fatigue explains a temporary decline in performance or power that is induced by exercise.[19] It is often classified into two categories: central fatigue and peripheral (aka muscular fatigue). There are different physiological mechanisms that facilitate central or muscular fatigue, and as such, the fatigue inducing protocol as well as the tests to determine presence or lack of fatigue are imperative.

Across the literature, there is a large variety of fatigue protocols that have been used to investigate ACL injury risk. These often take the form of direct game play, lab-based simulations of game play, neuromuscular electrical stimulation, or resistance exercise protocols. While still predominantly researched in men, there is a large number of studies that have either compared genders or solely investigated females. However, the "sport specific" studies are almost exclusively soccer based. Soccer is one of the most universal sports and has the highest rate of ACL injury by far, so this is not too surprising. However, it is shocking that while the sport with the second most prevalence of ACL injuries, basketball, has very limited validated protocols. The Basketball Exercise Simulation Test (BEST) claims to be the first of its kind and was developed in 2011. It was only investigated in males. Thus, there is a need for further development of

accurate basketball simulation protocols as well as basketball protocols validated for women.

While there has been literature investigating the association between fatigue and ACL injury risk, there are evident gaps that remain to be understood. The purpose of this study is to determine the effect of basketball game related stress on lower extremity mechanics and risk of ACL injury for active females. We plan to determine whether the BEST elicits fatigue for active females. We hypothesize that there will be an increase in fatigue following the BEST. Additionally, we plan to determine whether the BEST effects lower extremity skeletal muscle kinematics in active females. We hypothesize that there will be significant change in this variable following the BEST when observing jump landings.

The Musculoskeletal System and Anatomy

Cartilage, Bones, Joints, and Ligaments

The structure of the body is formed and held together by cartilage, bones, joints, and ligaments. Skeletal cartilage is a primarily water-based tissue that is molded to fit its body location and function.[77] There are three main types of cartilage: hyaline, elastic, and fibrocartilage. Hyaline cartilage is the most abundant and provides support with flexibility. Hyaline cartilage includes articular cartilage, costal cartilage, respiratory cartilage, and nasal cartilage. The elastic cartilage is similar to the hyaline cartilage but has more elastic fibers in order to endure repeated bending. This cartilage is present at the external ear as well as the epiglottis. The final type of cartilage is the fibrocartilage which allows for the most compressibility as well as tensile strength. This type of

cartilage is present in the discs between vertebrae and the menisci at the knees. The meniscus will be further described later in this review.

Bones are more rigid structures within the body that function to support the body framework, protect organs, anchor muscles, store minerals and growth factors, store triglycerides, help with hormone production, and are the location of blood cell formation. There are 206 bones in the body which are classified in two groups based on their location (axial and appendicular). The axial bones are the bones of the head and spinal cord and then the appendicular bones are those branching off of this central, axial, structure. All bones are classified by their shape. The primary bones of focus for this paper will be the lower extremity bones: femur, patella, tibia, fibula, and some of the bones of the foot.

A joint is a site where two or more bones meet.[77] There are three structural classes of joints: fibrous, cartilaginous, and synovial. The fibrous joint bones by collagen fibers. The synovial joints are freely moving and are the primary focus of this paper. Synovial joints adjoin bones covered in articular cartilage, separated by a joint cavity, and enclosed within an articular capsule lined with a synovial membrane. Some types of synovial joints include hinge joints, ball-and-socket joints, and pivot joints. A ligament is a band of dense regular connective tissue that bind bones together at a joint. This paper will look most directly at the anterior cruciate ligament.

Skeletal Muscle

Understanding the structure of the musculoskeletal system will provide insight into how the system functions. The tendons that connect muscles or muscle to bone can

extend over joints and contribute to joint stability. Skeletal muscles are constantly adjusting in order to maintain the positions of the body and allow for locomotion.

As a unit, skeletal muscle is covered by an outer protective layer of dense irregular fibrous connective tissue.[77] This tissue is called the epimysium and it works to protect the muscle from friction due to interaction with other muscles or bones and also works to ensheath the muscular unit. The skeletal muscle consists of muscle fascicles. These muscle fascicles bind muscle fibers together and have their own protective layering: a perimysium on the outside of the muscle fascicle to prevent friction among fascicles, and an endomysium on the inside to prevent friction between muscle fibers. The muscle fibers also have a protective plasma membrane called the sarcolemma. The type of skeletal muscle fiber is linked with its function.

Skeletal muscle fibers are classified based on the speed of the contraction, as well as the major pathways that supply ATP for the contraction to occur. The three major classifications of skeletal muscle are: slow oxidative fibers, fast oxidative fibers, and fast glycolytic fibers. Slow oxidative fibers are usually first recruited. While they have a slower contraction speed, they are fatigue resistant and do not heavily impact glycogen stores. The fast oxidative fibers are generally the second fiber recruited, but result in fast contractions that are moderately fatigue resistant. Finally, the fast glycolytic fibers rely more heavily on anaerobic glycolysis and have the fastest contraction speed but are easily fatigable. The actions of the muscle are generally reflected by the composition of the fibers. For example, muscles that primarily work to facilitate posture or go against gravity need to be fatigue resistant and will primarily be

composed of slow oxidative fibers. Muscles rarely consist of only one fiber type and usually have different proportions of fiber types per muscle and per individual, which can be altered some through training.

The muscle fibers are comprised of bound together myofibrils, which are rod-like contractile elements. The myofibrils consist of sarcomeres, the smallest functional units of the muscle, that are connected in series. While skeletal muscle does contain multinucleated myofiber cells, the primary substance of skeletal muscle is proteins. In addition, because of the vast energy requirements needed to move skeletal muscle, there is a significant number of large mitochondria present. The major myofibrillar proteins are classified in three ways: contractile proteins (~65% of the proteins present), regulatory proteins (~10% of the proteins present), and structural proteins (~25% of the proteins present).[77] However, there are other myofibrillar proteins that play a significant role in muscle contraction that are present in lower quantities. The myofibrils are structured as a series of connected sarcomeres.

The sarcomeres are the contractile unit of skeletal muscle and have their own complicated structure. They are repeating units between two Z lines and are primarily composed of two types of contractile myofilaments: actin and myosin.[77] Areas with high concentrations of thick filaments (myosin molecules) are darker and called A bands, while areas with high concentrations of thin filaments (including actin tropomyosin and troponin complexes) are lighter and called I bands. These bands of light and dark appear as muscle striations. Muscular contraction (described in further detail later in this paper) occurs when the thin filaments slide past the thick filaments, resulting in a shortening of

the muscle. In addition, there are elastic filaments present which maintain the organization of the A band and provide elastic recoil when tension is released. Within the center of the A band, there is an H zone which is a lighter region and contains an M line. The M line is a vertical bisection line formed by myomesin which act to hold adjacent thick filaments together. T tubules of the sarcoplasmic reticulum are also fused to the H zones to help with communication and ionic calcium release, which is necessary for muscular contraction, explained later.

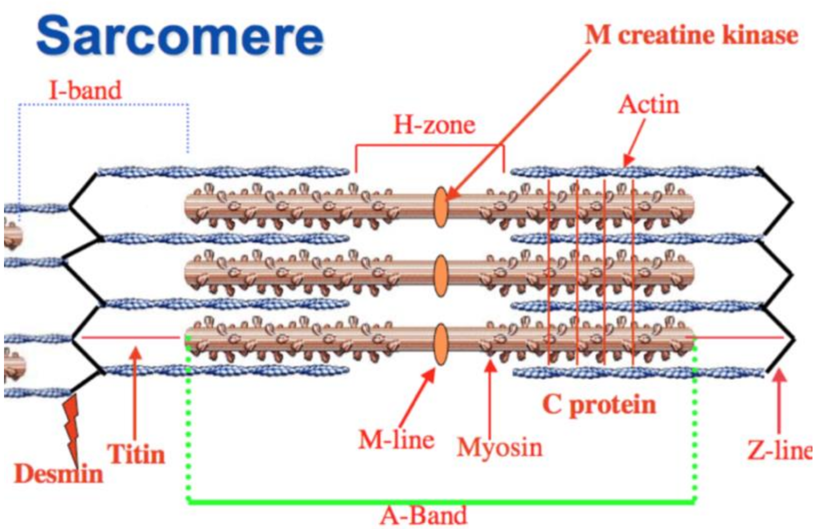


Figure 1.1 Sarcomere Diagram. This figure shows the structure of a sarcomere.[77]

The contraction of the sarcomeres in the myofibers result in the contraction of the muscle and regulate force production. This process will be explained in the "Muscular Contraction" section, later on. There are three main contraction types. A concentric contraction involves the shortening of a muscle which results in the generation of force. An eccentric contraction occurs when the muscle is elongated in

response to a larger opposing force. Finally, an isometric contraction generates force without changing the length of the muscle.

General Knee Anatomy

The knee consists of two primary joints: the patellofemoral joint and the tibiofemoral joint. The patellofemoral joint includes the patellar tendon and the patella. The patella is a sesamoid bone that is enclosed in a tendon and secures the anterior thigh, quadriceps, muscles to the tibia. It functionally protects the anterior surface of the knee joint and alters the angle of the thigh muscles acting on the knee to make for greater leverage. The patellofemoral articulation is commonly referred to as the extensor mechanism, given its role as a concentric action knee extensor. Functionally, the quadricep muscles also act as a knee extensor during eccentric running, gait, or jumping movements.[13] The tibiofemoral joint, as its name suggests, is formed by the meeting of the tibia bone and the femur bone. These bones support the majority of the body's weight and, as such, are the strongest bones in the human body.

The tibiofemoral joint is a modified hinge joint since it allows for flexion, extension, and some degree of rotation. It is stable due to a combination of static and dynamic elements. The static knee stabilizers include the passive structures of the knee joint capsule and various ligaments, such as the menisci, the coronary ligament, the menisco-patella ligament and the patella-femoral ligament. Although static and dynamic elements both play key roles in the stability of the tibiofemoral joint, dynamic stability plays a larger role in sports and, as such, is more heavily emphasized in this paper. The screw-home mechanism is the rotation between the tibia and femur which is essential

for knee stability while standing upright. Dynamic stability of the tibiofemoral joint is, in contrast, the result of the musculature surrounding the joint. The quadriceps act as a primary eccentric decelerator of the knee, the dynamic antagonist to an intact ACL and reduce posterior subluxation in a posterior cruciate ligament (PCL) injury event. The hamstrings function medially and laterally as antagonists to an intact PCL by directly reducing anterior subluxation, as well as tensing the ligaments of the knee through their insertion into the medial and lateral capsular ligaments in order to eliminate laxity and increase articular surface load.

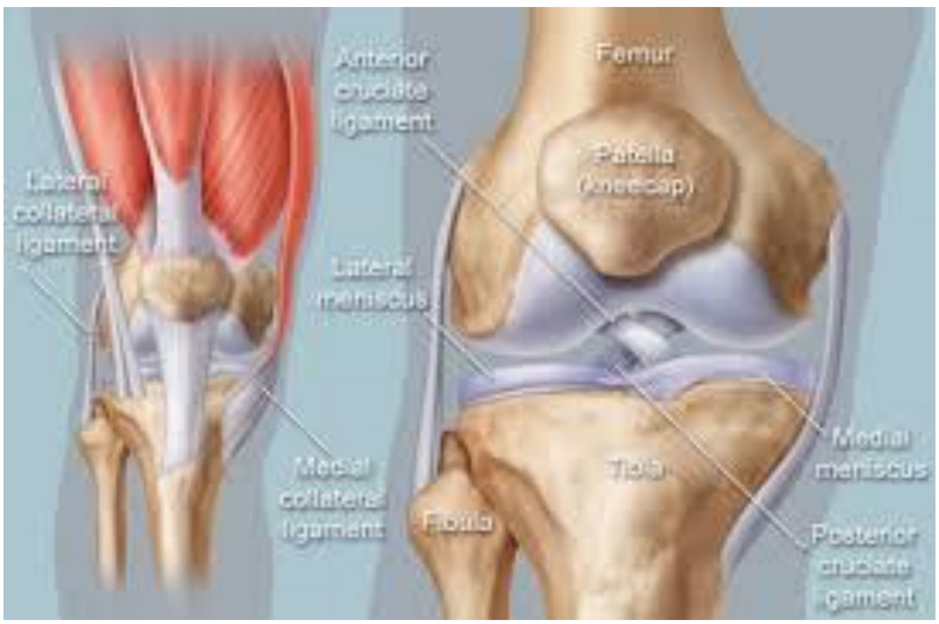


Figure 1.2 Knee Anatomy. This figure depicts the internal structure and anatomy of the knee.

The knee also contains cartilage-based menisci and five major ligaments. A meniscus is a flexible fibrocartilage cap resting proximal to the tibia but distal to the femoral distal epiphysis. Menisci play a vital role in supporting the load from the body's weight, and in dispersing forces in the body that result from movement or landing. The meniscus adds joint stability and acts as a buffer and spacer between the bones at the

tibiofemoral joint. The lateral meniscus anterocentral attachment merges with the distal attachment of the ACL addressed below. Because the lateral meniscus anterocentral attachment merges with the distal attachment of the ACL, patients with ACL injuries are more likely to tear their meniscus as well.

The five ligaments at the knee joint are: the anterior cruciate ligament, posterior cruciate ligament, transverse ligament, lateral collateral ligament, and the medial collateral ligament. These ligaments support various movements by muscles that surround the tibiofemoral joint.[13] It is not uncommon for subsequent injuries to occur when an individual ruptures their ACL, such as a torn meniscus or tear to the medial collateral ligament. However, the primary anatomical focus of this literature review is on the ACL because of its prevalence as an avoidable, non-contact sports injury, and because of the severity of the implications of tearing one's ACL.

Specific ACL Anatomy

The ACL has a proximal attachment within the posterolateral surface of the intercondylar notch of the lateral femoral condyle and courses distally to attach on the anterior medial aspect of the intercondylar eminence of the tibia.[34,78]Although some sources recognize a third bundle (the intermediate bundle), the scientific community generally recognizes two bundles that comprise the ACL: the anteromedial bundle and the posterolateral bundle.

These bundles are comprised of a highly organized collagen matrix where 90% are type I fibers and 10% are type III fibers (which accounts for three fourths of the dry weight of the bundles).[78]The collagen is organized in 20μm fiber bundles grouped in

groups 20-400µm in diameter. Water makes up 60% of the net weight of these complexes and the rest of the dry weight is from fibroblasts, elastin (<5%), and proteoglycans (1%).[34] The anteromedial bundle is located proximally and anteriorly in the femoral intercondylar notch and goes to the anterior aspect of the tibial ACL location. The posterolateral bundle starts in the distal and posterior aspect of the femoral intercondylar notch and goes to the posterior aspect of the tibial ACL location. These bundles allow for varying degrees of tension based on the degree of knee flexion.

With increased knee flexion, the bundles spiral lateral to medial, wrapping upon themselves as well as the posterior cruciate ligament resulting in increased tension as the tibia internally rotates. The anteromedial bundle is the primary restraint against tibial translation and as such is tighter in extension. The posterolateral bundle stabilizes the knee near full extension, primarily against rotatory loads and is tighter during flexion.[100] In full extension, the ACL lies against the intercondylar shelf assisting the ligament in preventing hyperextension. Other functions of the ACL include resisting anterior tibial translation (which biomechanically it is most effective at doing this at 20 degrees of knee flexion and accounts for 86% of total force resisting anterior draw), guiding the rotation of the tibia during the screw-home mechanism described above, proprioceptive feedback and signal to branches of the posterior tibial nerve, and assisting the quadricep hamstring interaction at stabilizing the knee joint. While the ACL plays a lesser role in resisting internal and external rotation, it has a max tensile strength of 1725±270 N.

The ACL is supplied blood from the anastomosis of the medial and lateral inferior geniculate arteries through the infrapatellar fat pad and the middle geniculate artery branching off the posterior capsule. The fat pad consists of adipose tissue and a small vascular system arising from the convergence of inferior medial and inferior lateral geniculate arteries and is suspended by a thin synovial fold, ligamentum mucosum. This blood supply system can be permanently disrupted with a substance tear to the ACL making the ability to heal after repair of a mop end tear to the ACL very poor. Because of this, reconstruction is the standard surgical practice, as opposed to repair.[34]

Anterior Cruciate Ligament Injury

Contact v Non-contact

ACL injuries are classified based off of the extent of the ACL tear. These classifications are known as Grade 1, Grade 2, and Grade 3 tears. A Grade 1 tear has only mild damage to the ligament, so the tibiofemoral joint is still stable. A Grade 2 tear is often referred to as a partial tear of the ligament. This grade suggests the ACL is stretched and becomes loose providing less stability to the tibiofemoral joint. A Grade 3 tear is a complete tear to the anterior cruciate ligament which results in that ligament no longer providing any stability to the tibiofemoral joint. These classifications are used to describe the severity of an ACL tear which occurs in one of the two main forms.

ACL injuries can take two primary forms: contact and non-contact. A non-contact injury is defined as one that occurs in the absence of body-to-body player contact.[144] A contact ACL injury is defined as an ACL injury that results from a direct blow to the knee.[93]An ACL injury where body to body contact occurs, but where the contact is not

directly with the knee, is referred to as a non-contact ACL injury with perturbation. Based on this broad definition, literature classifies between 70 and 84% of ACL injuries as non-contact.[52,2] It is believed that most noncontact injuries have the potential to be modified based on the mechanism of injury. As such, the prevalence of ACL injuries as well as the possibility of injury prevention has made this topic a popular research topic.

Prevalence of ACL injury in Sport

ACL injury occurs globally and has been evaluated in numerous studies. A study of New Zealand inhabitants, analyzing ACL injuries from 2000 to 2005, found that 58% of ACL injuries in the study were non-contact, and were more commonly nonsurgical and higher in incidence for males than females.[39] In contrast, incidence of ACL rupture rose from 86,687 (32.9 per 100,000 person-years) in 1994 to 129,836 (43.5 per 100,000 person-years) in 2006. During this period (1994 to 2006), the incidence of injury in females rose from 10.36 to 18.06 per 100,000 person-years, while the incidence in men rose from 22.58 to 25.42 per 100,000 person-years.[72] An epidemiologic study by the national football league in 2002 found that over 200,000 ACL injuries occur in the US annually.[47]It also found that female athletes are four to six times more likely to sustain an ACL injury than their male counterparts.[49,52,101]So, while there may be more cases of ACL ruptures in males overall, when focusing on sport situations, it is significantly more prevalent among females. ACL injuries are common in a large variety of sports, including football, soccer, basketball, volleyball, lacrosse, and handball, among others.

Sports are played around the world and in, conceivably, every minimally developed civilization. For this reason, the prevalence of ACL-related injury is of

worldwide impact and concern. A review by the English football league found that 808 players, of the 2,600 players followed, sustained a match injury that resulted in the athlete having to miss at least one game.[45]This English study found no significant difference in position, frequency of playing, or match half on the injury rate, but this finding was not specific to the incidence of ACL-related injury. When analyzing injuries in athletes playing soccer, a study including European athletes found that, although men and women had the same injury rate, women had a higher percentage of knee-specific injuries than the men.[96] There have also been numerous United States athlete studies focusing on soccer-based injuries with similar trend findings.

A study analyzing the incidence of ACL injury among different sports categories, within the United States, concluded that soccer players had the highest rate of ACL injury, with basketball players being a close second. Hawkins et al. found that professional soccer players have an injury risk that is 1,000 times higher than high risk occupational jobs (1999).[46] An epidemiology study focusing on knee injuries in athletics found that ACL injuries were the most prevalent (20.3% of injuries). Of the sports studied, soccer had the highest ACL injury rate (35% of ACL injuries by sport category). Another study found that 31.8% of the injuries faced by US women's soccer players were knee injuries.[40] Unlike other studies, this study also found that male US professional players had a higher overall injury rate per player hour when compared to female players (1.93 injuries per 1,000 player hours for females vs. 6.2 injuries per 1,000 player hours for men). Although contrary to the general incidence principles stated

above, this study was focused on all types of injury, as opposed to just specifically the ACL or knee.

One cohort study out of Olmsted county in Minnesota focused solely on the incidence of ACL rupture from January 1, 1990 to Dec 31, 2010.[109] The study found the annual incidence in this part of the US to be 68.6 new cases per 100,000 person-years. In this cohort, incidence was higher for males than females (81.7 cases versus 55.3 cases per 100,000 person-years). However, incidence for males decreased during the period of the study, while remaining relatively stable for females. As the study progressed, the incidence of ACL reconstruction increased significantly across all age groups. This increase was likely due, at least in part, to the increase in surgery as a medical trend, as well as new return to play procedures that were adopted in the field. These trends were, however, observed from a general cohort and demographic information as well as other subject characteristics were not stated. Thus, this cohort may not represent the findings for other cohorts or be representative of any specific athletic population.

In contrast, a comprehensive NCAA cohort study that looked at lacrosse, soccer, and basketball found that female ACL injury rates over a 15-year period were .28 (per 1,000 athlete exposures) for basketball, .32 for soccer, and .18 for lacrosse. Male injury rates were significantly lower, with .08 (per 1,000 athlete exposures) for basketball, .12 for soccer, and .17 for lacrosse.[83] A different NCAA review focusing on basketball and soccer found female soccer players had a .33 rate of ACL injury and female basketball players had a .29 rate of injury.[1] Male soccer players had an ACL injury rate of .11 while male basketball players had .08 injury rate. There was a similar contact versus non-

contact ratio between male and female basketball players, while male soccer players had a higher contact injury percentage compared to their female counterparts. During the thirteen-year period, the rate of ACL injury for male soccer players declined, but it remained constant for females.

Since noncontact ACL injuries are most common in soccer players, there has been significant amounts of research in this population. Several studies based in the United States have seen high amounts basketball related noncontact ACL injury, second to soccer. Given the broader scope of the world that participates in soccer, there have been fewer studies primarily focused on basketball ACL injuries. However, one United States professional basketball cohort study looked at 702 NBA players and 443 WNBA players. It found that WNBA players had a higher injury rate (24.9 per 1,000 people) when compared to NBA players (19.3 per 1,000 people). It found that about 65% of injuries were lower extremity injury (14.6 lower extremity injuries per 1,000 WNBA players and 11.6 lower extremity injuries per 1,000 NBA players). Specifically, the ACL injury rate was .8% for NBA players and .9% for WNBA players. This shows a slightly higher rate of ACL injury for female players. One could conclude that professional athletes have a lower ACL injury rate when compared with collegiate or high school athletes because of their greater experience, training, and access to resources.

Most at Risk Groups for ACL injury

There has also been ACL injury research that focused on identifying situations or populations that more frequently face ACL injury. Based on an epidemiological study of high school athletes, ACL injury was seven times more likely in game play than in

practice.[61] A cohort study found that ACL injury is more common among athletes in college than in high school.[11] In fact, one epidemiologic study focusing on knee injuries found that 43.1% of ACL injuries occur in the 20-29 year old age group.[73] When adjusted for sport and sex, there was a 2.38 times greater injury risk for college students than for high school students. It also found that the relative risk of a first time ACL injury for females was 2.1 times higher than males (when adjusted for level of play and sport). This suggests that female college students playing sport matches are at the highest risk of injury.

Other findings concluded that the ACL injury rate for American females in college sports was 2.49 times higher than for their male counterparts, and high school female athletes had a 2.3 times higher rate of ACL injury than high school males.[11] These trends are not consistent for the medial collateral ligament (which is higher for females in high school, but lower in college) or meniscus injuries (lower for females in high school, but no different between male and females in college). This information emphasizes the extent of ACL injuries as a national and global issue as well as introduces the increased risk females face for this injury.

Noncontact ACL injury is prevalent among athletes around the world. Within the United States, there is a clear trend of increased risk for collegiate females during game play, mostly from soccer and basketball. The two to eight time increase in injury risk for females when compared to males suggests structural or technical differences. It is important to understand the mechanism of this injury in order to understand the sex-based differences that could contribute to this risk. Determining those modifiable and

non-modifiable factors will be key for reduction of injury incidence and prevalence in this higher risk population.

Mechanism of Injury

There are a few mechanisms of injury that are consistent for ACL ruptures across all sports types. Some of the most common mechanisms include change of direction or cutting maneuvers combined with deceleration, landing from a jump in or near full extension, pivoting with the knee near full extension and foot planted, and knee hyperextension.[2,6,114,124] All of these actions result in knee valgus (also will be referred to as knee abduction), internal or external knee rotation, and anterior tibial translation force. When looking at the isolated forces alone, the most detrimental is anterior tibial translation with 20 to 30 degrees of knee flexion. However, most sport injuries happen in closed chain actions and are the result of a combination of forces acting upon the ACL.[2]

These findings have been consistent among several different sport-based studies. One study that focused on the ACL mechanism of injury for basketball concluded that the two primary mechanisms were: plant and cut movements that forced valgus and external or internal rotation to the knee close to full extension, and one-legged jump shot landings with forced valgus and external rotation with knee close to full extension.[93] Studies that focused on the soccer ACL mechanism of injury concluded that the most common non-contact mechanism was from a deceleration task with high knee internal extension torque, either with or without perturbation, combined with knee valgus, a planted foot, and body weight shifted over the injured leg.[2] Another

study that focused on female soccer players found that taller players with static knee valgus when standing upright were associated with higher degrees of knee valgus from vertical drop-jump landings.[91] The study found that only 11% of the variance was explained by the static valgus and suggested that other forces and mechanisms, like the ones above, work in addition to static knee valgus for the higher degree of valgus with drop jump landings.

However, there are some disparities between males and females which add to the likelihood of injury for females. A study comparing the kinematics and biomechanics of 15 males and 15 females during cutting movements found that, while there were no clear kinematic differences, females had smaller peak knee flexor moments and greater knee adductor moments during acceleration when compared to men. In addition, females had greater amplitude of quadricep electrical activity than men and increased frontal plane and decreased sagittal plane plant movements during early deceleration than men. The increased quadricep activity and smaller net flexor moments suggests less sagittal plane protection and more tibial translation at the knee joint for women than men.[117]

Another study found that, during landing maneuvers, females land with more knee and hip flexion. This results in 5.3 times greater relative risk of valgus collapse of the knee for females than males.[67]This risk is further increased in movement patterns perturbed by opponents. These results were further supported by other studies that found that females have more knee valgus and valgus velocity, significantly different dominant versus non-dominant strength, and different joint stability making them more

prone to injury.[37] These injury risks factors are often considered in two different classifications.

Factors that contribute to Risk of Non-contact ACL injury

The two main classifications of factors that contribute to the risk of an ACL injury are intrinsic and extrinsic. Intrinsic risk factors are present within the body, while extrinsic risk factors arise due to external forces that act on the body. Although this paper will briefly analyze some of the extrinsic factors that contribute to ACL injury, because extrinsic is more difficult to modify, the primary focus of this review is on intrinsic risk factors, such as anatomical, hormonal, biomechanical, and neuromuscular risk factors. Given that ACL injuries are more common in female athletes, this paper will also focus on those most influential in female ACL injuries.[6] Furthermore, it will analyze how sex can lead to greater risk in certain factors for female athletes.

Intrinsic: Anatomical differences. The main anatomical factors that put females at risk are anthropometric differences, static alignment, decreased notch width, increased joint laxity, increased muscle or hamstring flexibility, increased tibial translation, increased foot pronation and navicular drop, and changes during maturation.

There are clear anthropometric differences between men and women. Men have been found to, on average, be taller, heavier, and have a lesser percentage of body fat than females.[5] A longitudinal study that focused on female soccer players found that players with a body mass index one standard deviation above average had a 3.2 to 3.5 times increased risk of ACL injury. If the athlete was older than 11, that risk was

significantly higher as well.[130]It is noted that, around age 12, there is a link between the increase in BMI and pubertal development. However, there is no indication that BMI directly relates to ACL risk.[2] So, while there are clear differences between male and females anthropometrically, it is unclear any direct relationship between these measurements and ACL injury risk and is reasonable to assume other confounding variables might be at play.

Static alignment, including Q angle and pelvic width, are observably different between female and male structure. Q angle is formed by a line directed from the anterior-superior iliac spine to the central patella, and a second line directed from the central patella to the tibial tubercle.[2] Women naturally have a wider hip bone structure than men, which makes their Q angles larger.[52,124] Because of this, Q angle has been investigated as a factor related to injury risk for females.

While static Q angles do not seem predictive of knee valgus for injury risk, there is currently insufficient research on Q angle with dynamic landings to be able to determine whether there is any correlation.[52] However, one initial study suggests a relationship and supports the need for further research. In a basketball study, researchers observed that ACL-injured players had larger Q angles when compared with non-injured players.[2] While researchers were not able to conclusively determine that Q angle was associated with ACL injury risk or knee valgus, they did observe that a wider pelvic width to femoral length ratio was related to an increased risk of static and dynamic knee valgus. There was no direct link between static and dynamic valgus risk.

Thus, there is no clear evidence of Q angle and wider pelvic width as risk factors for ACL injury.

However, altered static alignment body positions, such as anterior pelvic tilt and torsion anatomic abnormalities, risks functional changes brought about by malalignment. Pelvic stability is a key factor for lower extremity kinematics and kinetics, which makes awareness of these body positions important.[2]

Anterior pelvic tilt places the hip in an internally rotated position where the gluteal muscles moment arm is changed, and the hamstrings are flexed into a weak, lengthened posture. The hamstrings normally function to prevent static and dynamic knee hyperextension and to prevent anterior tibial displacement, while the gluteal muscles assist in hip extension and act to prevent dynamic valgus collapse. While the exact degree of pelvic tilt that causes issues is unknown, anterior pelvic tilt positions increase knee valgus and subtalar pronation, which results in knee hyperextension, excessive navicular drop, and excessive subtalar pronation which is commonly found in ACL injured subjects.[2]

Torsional anatomic abnormalities also alter lower extremity biomechanics. Femoral torsion is the angle between the axis of the femoral neck and a transverse line through the posterior aspect of the femoral condyles. Femoral anteversion is an increased angle of femoral torsion which results in inefficiency of the gluteus medius by decreasing the moment arm. A weak glute medius impacts the amount of dynamic valgus collapse because the muscles become unable to keep the hip abducted, especially during weight bearing change of direction movements or cutting. A toe-in gait

indicates a femoral torsion position and is often associated with increased external tibial torsion related to functional valgus collapse at the knee joint.

In addition to the alignment-based risk factors, there are internal factors that can add strain to ACL, such as a decreased notch width. Decreased notch width size is an anatomical difference that some scholars believe increases ACL injury risk for women. Some studies have found that female athletes have comparatively smaller femoral notch widths, related to the size of their ACL, than do men. It is believed that a narrow notch leads to a smaller ACL, which results in higher ACL elongation under tension.[52] Uhorchak et al. found that women with narrow intercondylar notches had a 16.8 times greater risk ratio than those with larger notches.[130] These conclusions are disputed in the academic community because other studies have observed that, due to relative size differences, females tend to have narrower notches, but when those differences in size are normalized, the there is no longer a meaningful difference.

However, it is generally observed that, with increasing height, the notch width increases for males while remaining more consistent for females. The total condylar width, however, increases with height for both males and females.[52] But, one study found that a smaller intercondylar notch was positively correlated with injury risk.[2] While no clear relationship was established in this study as to the relationship of notch surface area to ACL size, there was a significant correlation between ACL cross-sectional area and notch surface area. Thus, the study found that smaller notch areas were related to less cross-sectional area of midsubstance ACL, which suggests that impingement of the ACL at the anterior or posterior roof of the notch may result in tibial

external rotation and abduction. Another study observed that most ACL injuries occur during partial knee flexion and that noncontact injuries tend to rupture close to the femoral attachment site. While likely for hyperextension injuries, impingement as a primary cause would suggest a midsubstance injury as opposed to the femoral attachment site.[16]

Another anatomical disparity between males and females, which is believed to influence ACL injury risk, is the increased general joint laxity observed in females.[2] Joint laxity is the result of increased stretch in ligaments. A study by Uhorchak et al took place at the US military academy and, while its findings are not specific to other sport-based populations, they still provide insight into the relationship between joint laxity and injury rates. The researchers found that women with generalized joint laxity had a 2.7 times greater risk for ACL injury when compared to those without laxity.[130] Proponents of the study believe laxity not only affects the sagittal plane with hyperextension but can also affect coronal knee motion with valgus. Females in the study were also found to have had higher anterior posterior knee laxity, and higher general joint laxity, than the men.[2]

One factor considered when looking at laxity is the composition of the anterior cruciate ligament itself. Some studies have found that female ACLs, unlike male ACLs, are smaller in length, cross-sectional area, volume, and mass.[2] In addition, females have lower fibril concentration and a lower percentage of area occupied by collagen fibers. Because of this, female ACL stiffness and modulus of elasticity is highly correlated with fibril concentration, while male ACL failure load and strength was more highly

correlated with percent area of collagen. A cadaveric knee study found that females had a lower tensile linear stiffness, with less elongation at failure, and had a lower energy and load absorption at failure. However, the reliability of results obtained from cadaver studies is very low because of structural limitations. For this reason, the translation of the cadaveric knee study's findings to real life mechanisms is unknown.

However, there have also been soccer and basketball-oriented, in vivo, studies on joint laxity. One study found that ACL-injured athletes had significantly higher general joint laxity than the healthy athletes in the control group. 78.7% of the ACL-injured athletes exhibited knee hyperextension, while only 37% of the uninjured controls displayed this trait. Based on this data, the study concluded that knee joint laxity increased varus-valgus and internal/external rotation knee laxity as well as functional valgus collapse (mechanisms for ACL injury) for young female soccer and basketball players.[2]

In summary, joint laxity is often associated with hyperextension and anterior-posterior tibiofemoral translation, which has been held to indicate a higher risk of ACL injury. Laxity alters the dynamic lower extremity motions and loads in a multiplanar fashion, which places the ligament in a position that has a higher risk of rupture. One of the biggest questions about joint laxity that still remains is what role general and specific laxity actually plays in the risk of ACL tear in conjunction with neuromuscular factors.

One specific muscle laxity seen in females is increased hamstring flexibility. Healthy and active men and women with more stiff hamstrings have a reduced risk of

ACL injury.[14] Hamstring laxity can result in delayed muscle action, which results in abnormal co-contraction of the quadriceps and hamstrings during foot strike.[5,52]The co-contraction of these muscles is a protective mechanism, but its functionality is reduced when the knee is fully extended. This likely contributes to the tibiofemoral joint compression force with minor posterior protective forces.[16]

One study, that supported this idea, applied direct stress to the ACL, which resulted in moderate inhibitory effect on the quadricep and excited the hamstring muscles. A similar response was observed for patients with ACL damage during a loaded knee extension with tibial subluxation. This suggested an alternative reflex arc unrelated to ACL receptors to maintain joint stability. In brief, the hamstrings were shown to assume the role of joint stabilizer in a deficient ACL.[121] Since females tend to have lax hamstrings, this secondary protective measure is inefficient and, while not the sole reason for ACL injury, ACL-injured athletes often have more lax hamstrings.[5,52]

In addition to the noted muscle laxity, there is observed ligamentous laxity as well. One of the most important risk factors for females is increased anterior tibial translation. The ACL is meant to limit tibial translation relative to the femur.[52] Given the natural laxity of females, it is hypothesized that this laxity allows for anterior shifting before muscles control movement. It is claimed to be a sport-specific adaptation resulting from the quadricep/hamstring activation ratio and ligamentous laxity. But, studies have found that females tend to have increased anterior-posterior laxity one standard deviation or more above average.

The final issue with female laxity is observable at the ankle, which can have a tendency towards increased foot pronation and navicular drop. A navicular drop plays a role in lower extremity alignment, as well as tibial translation, since it moves the tibia forward and increases ACL strain.[52] The impact of the foot on ACL injury risk is less understood, but it is believed that there is a link between ACL injury and subtalar joint overpronation. In addition, very little is understood about the navicular drop and its effect on knee motion and torque.

There has been more focus in the literature on increased foot pronation when compared to navicular drop. Beckett et al found a direct relationship between subtalar joint hyperpronation and ACL tears when comparing 50 patients with past ACL injury and 50 uninjured controls.[8] Those with ACL tear history had greater navicular drop test scores. A 1994 study that focused on gymnasts, American football, and basketball players with ACL injury history also found greater subtalar pronation in the ACL injured group.[140] However, conflicting information has been found where there was no significant difference in foot structure measurements for men and women. While another study concluded that subtalar joint neutral position measurements and navicular drop tests were not associated with ACL injury for male and female college soccer and basketball players.[2] In addition, it found that dynamic medial foot landing was not related to an increase in propensity to demonstrate high ACL injury risk in biomechanics. It suggested, instead, that subtalar pronation compensatory increases in internal tibial rotation during knee extension, which usually occurs during the contact phase of gait. This excessive internal tibial rotation transmitted abnormal forces up the

kinetic chains, preloading the ACL. It is believed that subtalar joint pronation and internal tibial rotation at the knee may produce increased internal femoral and valgus angulation at the knee, resulting in greater stress on the tibiofemoral joint and increasing the risk of ACL rupture.

The final anatomical factor commonly investigated is maturation. Maturation of the female body has been linked to biomechanical and neuromuscular change. While there is no gender-based difference in ACL injury rates in prepubescent athletes, the change after puberty in levers results in different amounts of force on the knee. It is believed that, neurologically, females do not adjust as quickly or efficiently to this change, causing an increase in injury risk.[52] An additional hypothesis for this maturation ACL injury gender gap was a lack of stiffness, or reduction in flexibility, for females when compared to males. Increased stiffness is viewed as being more protective against injury. A study that tested the mechanics of drop vertical jumps for males and females following maturation observed that, while both males and females increased active knee stiffness during a one year span, males had increased hip and ankle stiffness.[36] So, contrary to their hypothesis that post maturation females would not see increased stiffness, there was an increase in active knee stiffness. However, the male counterparts had increased knee, hip and ankle stiffness and at greater observed levels than the females.

In conclusion, there are several anatomical differences that need to be considered to determine the factors contributing to the increased risk of ACL injury in females. Of these anatomical differences, the ones that seem to contribute most are

increased joint laxity, anterior tibial translation, and hamstring and quadricep activation ratios.

Intrinsic: Biomechanical. Biomechanics focuses on the application of mechanical laws and underlying forces apparent during static structure or movements of individuals. Technology (such as force plates, 3D analysis, electromyography, etc.) is often used to quantify the forces or relevant lever lengths in biomechanical studies. When so used, there have been common trends observed that relate to the protection of the ACL and the increased risk of ACL rupture. Some of those features include knee abduction, axial/compressive forces, trunk and hip activation, tibial plateau geometry, ground reaction forces, tibiofemoral kinematics, and ankle and foot mechanisms.

Knee abduction is often a focus in biomechanical studies as a predictor for ACL injury risk. It is believed that knee abduction moments directly contribute to lower extremity dynamic valgus and joint load. Greater knee valgus, or knee abduction, can place a greater axial force on the lateral side of the knee, when compared to the medial aspect, and lessens the compressive load threshold of the ACL by increasing the lateral compressive force. In addition to the greater internal rotation component, this makes the lateral ligament relax and the medial ligament contract more forcefully, which may allow for the lateral tibial plateau to shift anteriorly with the internal rotations.[16,53] Unanticipated sport conditions, like side stepping, further results in greater varus-valgus and internal-external knee moments by increasing muscle activation (anywhere from 10 to 25 percent), making noncontact injury more likely.[52]

Knee abduction during stance or prior to completing actions such as landing, cutting, etc. is said to predict ACL injury risk with 73% sensitivity and 78% specificity.[52] One study, comparing people with ACL injury history with those without it, found that the ACL-injured group had more than eight degrees of knee abduction compared to the noninjured group. The study concluded that knee abduction angle correlated with peak vertical ground reaction force in the ACL-injured athletes, and that knee flexion alone could not predict ACL injury. While valgus moments do not inherently load the ACL, and valgus rotation is not associated with ACL injury, valgus is a stated risk factor for noncontact ACL injury, and training programs aimed at reducing abduction knee moment have been successful in decreasing ACL injury risk.[16]

While knee abduction findings have been fairly consistent, there is conflicting information on the effect of axial compressive forces on ACL injury. While some sources have found that excessive joint compressive load and internal force on a cadaver results in an ACL rupture with 2900 N to 7800 N of force and 30 to 120 degrees of knee extension, other studies suggest a compressive load of 1812N to 2659N will result in failure.[16] These other studies suggest that inadequate absorption of the ground reaction forces by the lower leg allows for the posterior displacement of the femoral condyles in the tibial plateau as a result of the compressive force on the ACL's posterior tibial slope. Thus, axial weight-bearing compressive forces, in conjunction with increased tibial posterior slopes, results in anterior tibial force on the ACL and leads to increased likelihood of a rupture. Historically, it was assumed that the axial compressive forces on

the tibiofemoral joint would not increase strain on the ACL, but recent studies show that this force results in anterior tibial translation and could, in turn, increase ACL injury risk.

Trunk and hip activation patterns can also make the individual more susceptible to ACL rupture. Hip torque was significantly different for females with ACL injury when compared with non-injured controls.[52] The injured athletes also had greater peak external hip flexion moments, or torque. A similar study found that individuals with ACL rupture had higher hip flexion angles at initial ground contact.[16] Mean lateral trunk angle, relative to the vertical plane, is also something more common in females, and it is believed to be associated with an increased risk of ACL injury. A lateral lean results in a change in the location of the center of mass, which adds significant axial force to the lateral knee compartment. It also adds to knee abduction and valgus moments during weight bearing. For females, side to side imbalances in neuromuscular coordination, as well as strength and flexibility, is a predictor of injury risk.[38] In addition, the torso is typically farther posterior compared to the knee in subjects with ACL rupture. This study found that hip flexion and knee extension torque is needed to stabilize the torso during landing but relying on the rectus femoris activation results in increased ACL strain by adding compressive and anterior force on the tibia.

Another study, focusing on thirteen female and nine male division I collegiate athletes, found that the females had a different muscular activation pattern when landing from drop landings than the males. The females had decreased gluteus maximus activation and increased rectus femoris activity, which further suggests that increased quadricep activation and decreased hip use could be a factor causing increased risk of

ACL injury in females.[142] In addition to this potential load on the tibia by an increased quad and decreased hip interaction, it is also important to focus on the tibial plateau geometry itself.

Tibial plateau geometry has also been considered as a potential predictor of injury risk. An MRI study of 23 female participants, which looked at their landings from double-leg drop jumps, assessed the tibial plateau geometry and landing response.[116] It found that lower coronal tibial slopes predicted initial and peak hip adduction and knee valgus angles, while greater coronal and lateral tibial slopes predicted greater hip internal rotation at initial contact. Also, greater coronal tibial slope and lower medial/lateral tibial slope ratios predicted greater knee internal rotation. However, there was no clear association between joint geometry and hip or knee peak joint moments. In addition, little information is understood about how the tibial plateau geometry can affect ground reaction forces.

If not absorbed efficiently, ground reaction forces can put the knee in a compromised position. It is estimated that the max peak vertical ground reaction force experienced by a one-leg landing after a jump can be 2 to 18 times body weight.[16] This impact can be reduced with a plantarflexed position, since it allows for more of the force to dissipate and displaces the center of mass posteriorly, which further protects the ACL. In contrast, having shorter time spans for foot landings shortens the calf muscle contraction and the absorption of forces, making the ground reaction forces impact the knee more directly. Another study estimated that the lower extremity absorbs 19% of the body's kinetic energy and the hip extensor eccentric contraction is responsible for

22% of the total kinetic energy absorption.[52] The limited use of hip musculature previously described could result in a failure to absorb these higher ground reaction forces in females.

Tibiofemoral kinematics have also been identified as predictors of ACL injury. Individuals with more vertical tibial slopes relative to the femur were found to have a significant risk for ACL injury.[16] As the tibial slope becomes more vertical, the point of contact of the lateral compartment is closer to the sulcus on the lateral femoral condyle. Also, the lateral femoral condyle will come into contact with the tibial plateau on the flatter anterior surface rather than the rounder posterior surface. As the knee extends and the hip flexes, the angle increases between the tibial plateau and the distal aspect of the femur. The femoral shift causes vertical displacement, which promotes anterior tibial shift. In addition, this differing angle is hypothesized to be associated with the increased risk from exceeding 20 degrees of flexion. However, more research is needed to determine the strength of the association of vertical tibial slopes and the knee flexion angles related to ACL injury.

Finally, ankle and foot mechanics have also been studied in biomechanical settings to determine their impact on ACL injury risk. It was found that subtalar pronation, measured by navicular drop, was reported to be greater in ACL-injured patients than in non-injured counterparts.[16] However, there is limited research in this area, as well as contrary findings in other studies. In addition, increased ankle eversion is also considered by some to be a factor that contributes to ACL ruptures. In terms of foot mechanics, subjects with ACL ruptures tend to land with flatfoot or hindfoot, while

non-ACL rupture athletes tend to land on forefoot. It is hypothesized that injured athletes have less ankle plantar flexion at the initial ground contact point and that people with ACL rupture reach flatfoot 50% sooner than controls. However, more research needs to be done to more fully understand ankle and foot compromises that relate to ACL injury.

In summary, knee abduction, axial/compressive forces, trunk and hip activation, tibial plateau geometry, ground reaction forces, tibiofemoral kinematics, and ankle and foot mechanisms have been observed via biomechanical settings to identify protective or compromised positions in relation to ACL injury. The factors most associated with ACL injury risk was increased knee abduction, compressive forces, reduction in hip activation, and excessive ground reaction forces.

Intrinsic: Hormonal. Although research is currently inconclusive, it is believed that female hormonal changes potentially impact neuromuscular control (related to ACL injuries), as well as central nervous system control.[52] The menstrual cycle has been broken up into three phases: follicular (days 0-9), ovulatory (days 10-14), and luteal (days 15-28).[2] During these different periods, there are fluctuations in sex hormones. While it is generally accepted that there is an estrogen surge during pre-ovulation, the exact stage of the menstrual cycle that results in hormonal imbalances which lead to ACL risk is greatly debated.[2,52] It is clear, however, that relaxin and estrogen impact ACL risk in females.[52] Also, the existence of estrogen and progesterone receptor sites on ACL cells suggests that changing hormonal levels could ACL injury risk. Thus, there have been

studies looking at the link between sex hormones and anterior knee laxity, tensile strength, and neuromuscular function.

It has been suggested that there is a positive association between sex hormones and anterior knee laxity. Although some studies had contrary findings, three studies found statistically significant associations between the menstrual cycle and anterior knee laxity (mainly during the ovulatory or post ovulatory phases).[2] Oral contraceptive use was also linked to a decrease in ligamentous laxity in female soccer players, which in turn lowered their traumatic rate. Females who used contraceptives saw a decrease in impact force, a reduction in medial and lateral torques at the knee, an increased hamstring to quadricep strength ratio, an increased stability on one leg, and a decreased knee laxity relative to non-contraceptive users. These findings suggest that hormonal stabilization increases dynamic stability, which reduces injury risk. In contrast, other studies observed a tendency for anterior tibial displacement to increase in those using oral contraceptives, when compared to those not on hormonal replacement therapy.[2] In addition to those studies that focused on hormones and knee laxity, there have been investigations on the impact on ACL tensile strength.

Estrogen and progesterone are believed to impact ACL tensile strength. Estrogen decreases fibroblast proliferation and type I pro-collagen synthesis,[2] suggesting high estrogen levels affect the structure of the ACL. Higher progesterone levels attenuate this estrogen inhibitory effect on collagen metabolism in females in both a time dependent matter and in dose response, though to exactly what extent is still uncertain. Further, these findings are primarily from animal models, making their application to humans

controversial. Other animal studies provide support based on findings that physiologic concentrations of estradiol decrease ligament strength, while relaxin decreases soft tissue tension. However, a sheep model found that, while the sex hormones altered the tensile properties of the ligament, there was no difference in max force, stiffness, energy failure, or failure site of ACL when compared to without estrogen implants.[2,52] These findings suggest that differing tensile properties might have less impact on individuals and their performance, though it is unclear whether the same results would be observed in a human study.

Other studies, assessing the effect of hormones on neuromuscular function, have reached contradictory results. One study found that, during the ovulatory phase, there is an increase in quadriceps strength, a decrease in muscle relaxation time, and an increase in muscle fatigability in young, healthy sedentary females. Female sex hormones were found to decrease motor coordination, aerobic capacity, and high-intensity endurance.[2] Another study comparing females (some using contraceptives and some not) to males found no significant difference in knee angles or phases during any phase between the two female groups, and no substantial difference between female and male measurements. It concluded that contraceptive use did not directly affect knee or hip joint landing mechanics for jumping and landing tasks.

Further research is necessary to determine the physiological effects that the different phases of the menstrual cycle, and their related hormonal fluctuations, have on the individual. Then, determining the extent to which these hormonal changes affect human athletic performance will help the community apply the knowledge gained.

Intrinsic: Neuromuscular. Dynamic stabilization of the knee is facilitated by the neuromuscular system. Coactivation of muscles, and a certain level of coordination, are required to protect the knee joint. The quadriceps and hamstrings play crucial roles in reducing knee motion and ACL load.[2,52] An in vivo study, conducted using a strain gauge, found that isolated quadricep contraction near extension resulted in more strain on the ACL than did co-contraction of the quadriceps and hamstrings. This suggests that the peak quadricep strain on the ACL was about 30 degrees of knee flexion, and that deficits in hamstring strength, in turn, would result in increased quadricep strain on ACLs.[2,52] It further requires quadricep and hamstring activation balance in order to reduce the likelihood of injury. The hamstrings functionally reduce the load of the quadricep, and it is believed that they also improve dynamic knee stability by resisting anterior and lateral tibial translation and transverse tibial rotation.[2] Low level hamstring muscle activity, coupled with relatively unopposed quadricep contraction, is believed to produce significant anterior displacement of the tibia during eccentric contraction at foot strike, and is also linked with ACL injury.

Those findings have been further supported in other studies. Studies have found that the females with ACL injury history had a significant decrease in hamstring strength compared to the groups without ACL injury history.[88] Also, the females without ACL injury history had hamstring strength similar to the males, but had weaker quadriceps.[88] In addition, co-contraction of the knee flexors is required to: balance active contraction of the quadriceps, compress the joint, assist in control of high knee abduction torques, and reduce anterior tibial translation.[52] Joint compression through muscular co-

contraction allowed more of the valgus load to be carried by articular contact forces, which protected the ligaments of the knee. Proper muscular contraction had three times less impact on valgus and varus laxity in the knee. Some scholars believe that the anterior force vector of the quadriceps is the primary contributor of ACL force injury because the quadriceps are the primary producer of anterior knee force at or near full extension.

In contrast, other studies have observed that the compressive vector of the quadricep is at least double the size of the anterior shear vector at zero degrees.[16] Models have shown an increase in ACL strain at or near full extension, with quadricep activation, which leads to a posterior force. This points to a larger quadricep influence on ACL injury risk, via increased compressive load placed on the tibiofemoral joint, as opposed to anterior force. There are however, other neuromuscular disparities between males and females.

The neuromuscular disparities between males and females that are most commonly attributed to increased ACL injury risk include: relative muscular strength and recruitment, and trunk proprioception. One study looked at neuromuscular performance characteristics in elite female athletes, including: anterior knee laxity, lower extremity strength, endurance, muscle reaction time, and muscle recruitment order (in response to anterior tibial translation). The females were found to have more laxity, less strength, and greater hamstring torques during isokinetic testing than the males. There was no difference in cortical or spinal reaction timing. However, muscle recruitment order in elite female athletes differed from their male counterparts, where

only the female group relied on the quadriceps during anterior tibial translation.[59] In the study, the main observable differences appeared to be in joint laxity and muscular recruitment.

There are conflicting beliefs about the magnitude and timing of muscle activation for females versus males. One study found that females had slower hamstring activation response following anterior stress on the ACL, while another study found females tended to have earlier hamstring activation before landing movements.[52] The latter study believed that the early activation altered muscle synchrony and increased valgus/varus and internal/external knee moments for unanticipated sport movements.

Similarly finding protective preactivation of the hamstrings, other research has also concluded that lower extremity musculature was significantly activated when the foot was on the ground. This muscular activation strategy of decreased medial quadricep and hamstring activation is believed to limit the effectiveness of active muscular control and the ability of active muscular control to synergistically work with the passive joint restraints to create dynamic knee stability. It also suggested that the preactivation of quadriceps is linked to increased valgus during cutting and landing movements.[52]

The unconscious activation of dynamic restraints surrounding the knee joint in response to sensory stimuli were found to have differing activation patterns between males and females. Females had increased adduction and internal rotation of the femur, reduced hip and knee flexion angles, increased dynamic valgus, increased quadricep

(and minimal hamstring) activity, and decreased muscle stiffness around the knee joint.[2]

Imbalanced muscle firing patterns is also considered a factor that contributes to female ACL injury risk. One study, comparing males and females, found that females had four times the firing of their lateral hamstrings during the deceleration of a jump landing.[52] A decreased ratio of medial quadriceps musculature recruitment to lateral quadricep recruitment was observed and, combined with the unbalanced medial hamstring recruitment, is thought to be related to decreased control of coronal plane forces at the knee.[2] In addition, decreased joint compression limits the ability to passively resist dynamic valgus and anterior tibial translation. Accordingly, knees in females are predisposed to exhibit medial femoral condyle lift-off from the tibial plateau, as well as increased loads on the ACL during decelerated landing or cutting maneuvers. This altered pattern of muscle firing is believed to increase ACL injury risk by compressing the lateral aspect of the tibiofemoral joint, opening the medial aspect of the tibiofemoral joint, and increasing the anterior shear force at the joint.

This imbalance of muscular firing has been observed in sport-based studies as well. Female soccer, basketball, and volleyball players, as compared with males, were all found to land with increased quadricep activation and decreased hamstring activation, which led to an increased load on their ACL during landing in jump-stop tasks.[2] They also had increased soleus and quadricep activation during hopping. In addition, the knee adduction and abduction moments significantly decreased after plyometric exercises and were suggested to be the sole significant predictor of peak landing force. A

decrease in adduction or abduction is thought to decrease the risk of femoral condyle liftoff from the tibial plateau. There were no significant findings in terms of peak landing flexion or extension moments as a predictor of peak landing force.

Finally, having decreased proprioception reduces the ACL's ability to sense torque and respond to elongation.[52] This makes the ACL more vulnerable to experiencing torques and translation. Generally, agonist musculature fires in response to perturbations that put torque on the joint. That firing pattern can be altered with neuromuscular training, which would enhance the stretch reflex response to the stretch of the ACL and the activation of the hamstring muscles.

In conclusion, female athletes tend to have different muscular coordination than males, which makes them more susceptible to injury. The increase in quadricep activation and subsequent decrease in hamstring co-activation, combined with an increased general joint laxity, makes females generally at greater risk for injury than males. Players that are out of balance or face additional perturbations see further impacts on their coordination and intended movement.[2]

Extrinsic. External factors believed to have the most impact on the likelihood of an ACL injury include: environmental conditions, footwear, and whether the athlete uses a brace.

Environmentally, it is believed that hot conditions are more likely to contribute to risk of injury than cold conditions.[94] Heat is more likely to add to athlete exhaustion and lead to mechanical changes for tasks. In addition, in colder conditions, individuals tend to move more slowly resulting in less extreme demands on the body. In terms of

field conditions, studies have found that 95% of ACL tears occur during dry field conditions and are more often on natural grass than artificial grass,[52]but less likely on outside grass than indoor turf.[94] In dry conditions with natural grass, there is increased friction and torsional resistance from the shoe-surface interface when compared to wet conditions resulting in this disparity. For turf, hot conditions affect the turf temperature which will also change the shoe interference and friction.[2]

However, there are conflicting views about the shoe surface interaction. This is because many different factors are influential, including: ground hardness, ground coefficient of friction, ground dryness, grass cover and root density, length of cleats, and the relative speed of the game. In terms of the footwear itself, the number, length, and placement of cleats could play a role as well, but there is no conclusive evidence. One "edge" cleat design has been specifically linked with higher torsional resistance and injury. It has longer irregular cleats on the peripheral margin of the lateral sole of the shoe and some small pointed cleats on the medial sole of the shoe. However, there are still too many components to determine if shoe design, or other factors, such as foot mechanics, foot-shoe interaction, or surface interaction, are the primary cause.

Finally, braces can alter the likelihood of injury. Without stabilization of the knee by the gastrocnemius, hamstrings, and quadriceps, a brace was found to reduce anterior tibial translation by 29% to 39%. With muscle activation, this reduction in tibial translation rose to between 70 and 80%.[137] There were, however, observed decreases in hamstring reaction time with use of a brace. So, while the brace was useful for reducing tibial translation, it also reduced the natural ability of the hamstrings to control tibial

translation. Wu et al found no functional performance improvement in those using a brace after ACL reconstruction when compared with those not using a brace, and found that bracing slows down the rate of turns and running.[141]In addition, another study found that postoperative bracing of the ACL had no significant impact on knee stability, functional testing, subjective knee scores, and range of motion or strength testing.[79]

In conclusion, intrinsic factors are more easily modifiable and have been further researched than extrinsic factors when looking at ACL injury. While not always controllable, being aware of the environmental conditions, shoe surface interaction, and impact of wearing a brace as well as how they could impact injury rate for players is important. Coaches could consider these factors prior to the match and attempt to alter game plan or pace to protect the athletes. These extrinsic factors should at least be considered, if possible, when designing the surface for the players or selecting shoes or braces in order to prevent ACL injury.

ACL Reconstructive Surgery

The impact of an ACL injury often goes beyond the injury itself. According to a 2006 national survey within the United States on ambulatory surgery, done by the Centers of Disease, Control, and Prevention (CDC), about 127,446 ACL reconstructions took place in 2006 alone (confidence interval 95,124 to 159,768).[64] A study found that the cost, on average, for an ACL surgery in New Zealand was $8,574,[39]while in America the average cost for ACL surgery and rehabilitation ranged from $17,000 to $25,000.[52] For both studies, these average costs represent some combination of both partial and complete rupture reconstructive surgeries. While differing health care systems could be

related to the cost differences, the emphasis of this point is the increased burden on Americans athletes or individuals who need ACL reconstructive surgery. However, this cost is greatly related to the extent of injury.

Usually, the first step in determining the extent of an ACL injury is analyzing a magnetic resonance image (MRI). The best visual of an ACL on an MRI is from two or three sagittal sections, since a sagittal view has a higher specificity than a coronal view and is better imaged on a T2 sequence.[78] A non-ruptured ACL will have fairly low signal but may appear linear towards the distal ACL aspect. A rupture in the fibers or soft tissue mass in the notch will have high signal characteristics from edema and hemorrhages indicating the tear. A partial tear may appear by an increased signal with thickening or redundancy in the ligament image, but partial tears are still very challenging to accurately diagnose.

The blood supply matrix for the ACL makes it increasingly difficult to naturally repair, and thus, the standard treatment for this injury is reconstructive surgery. An epidemiological study focusing on high school athletes found that 76.7% of ACL injuries resulted in surgery.[61] The primary aim of reconstructive surgery is to restore the function and range of motion to that of an intact ACL.[78]

Surgery typically requires graft insertion. There are three main graft options: autografts from the hamstring or patellar tendon, synthetic grafts, or cadaver grafts. The choice of graft is generally made by the treating physician's preferences based on their training, but it is believed that autografts are the ideal option since the body is more likely to recognize and accept the graft. In special cases, other considerations for

the surgery must be contemplated. For example, differences in structure between females and males, addressed above, makes females a higher risk population for ACL tears. Anatomically, females might have weaker or smaller hamstrings or patellar tendons which can affect their ideal graft option.

Following graft insertion, it has been suggested that 8-10% of ACL reconstructions result in recurrent instability or graft failure.[78]There are different factors that can contribute to graft failure, including: graft selection, tunnel placement, initial graft tension, graft fixation, graft tunnel motion, and the rate of graft healing. It is believed that femoral tunnel placement that is too far anterior on the femur can result in vertical orientation of the graft. This improper placement often leads to excessive tension and graft flection. Ideally, the graft should be placed near the center of the ACL attachment site on the femur.

Timing of the reconstruction can also play a major role in the success or failure of the surgery. A retrospective cohort study, analyzing the relationship between second intra-articular injury and time from injury reconstruction, found that longer periods prior to surgery were associated with increased incidence of injuries for trochlea, lateral femoral condyle, medial tibial plateau, and medial meniscus.[105]There were also trends that less active athletes, prior to the ACL reconstruction, had a greater risk of medial meniscus and trochlear injury, while those more active prior to reconstruction had a greater risk for medial tibial plateau injury.[105]This suggests that it is important not to wait too long after injury before receiving surgery, since increased time can result in additional problems.

However, an NFL surgeon questionnaire found that there was no firm timetable established for reconstruction.[21]Typically, it is recommended to wait at least a week after the time at which the inflammatory response subsides before performing surgery. In addition, the severity of the tear and the role of the athlete sometimes influences the decision for reconstruction. The NFL survey found that, for a quarterback, surgeons were more likely to wait until the season was over to perform surgery, while for the team's place kicker, they would try to seek surgery much sooner.[21] This suggests that in a sport setting, the role of the player on the team could also be a deciding factor for timing of reconstructive surgery.

ACL reconstruction can be a time-consuming process laced with expensive medical bills. There is a heavy importance on the skills of the doctor and their graft selection as well as placement for the success of the surgery. Following surgery, rehabilitation is often required to help progress the individuals back into functional movement patterns and range of motion.

Rehabilitation

The rehabilitation process is a gradual one. Initially, the individual must rest to allow for a certain level of healing and fusing of the graft to take place with their fibers. After ample rest, physical therapy sessions are set up to phase the individual back into returning to their position.

The initial phase is to work on bracing, controlling edema and increasing the range of motion to the joint through passive movements. Some quadricep motions might be included at this point.[2]Studies have found that the inclusion of neuromuscular

electrical stimulation (NMES) may be a viable option post-surgery to ease into rehab.[68]NMES has been used for strengthening, has been found to reduce atrophy, and is a treatment for edema.

The second phase encourages patients to leave crutches, discontinue the use of a brace, and to start progressive resistance exercises.[78]This phase mainly includes biking with resistance, leg extensions, and increasing flection.

The third phase consists of increasing quadricep and hamstring strength via both closed chain and open chain exercises. This phase also allows for the introduction to use of a stair climber and plyometrics. Squats have been found to be useful at this stage, since hamstring muscle activity in vivo during squat exercises functions synergistically with the ACL to provide anterior knee stability.[87]

Fourth, full weightlifting programs, including balance exercises and progressive running, is allowed. The fifth and final phase allows for up and down running with more sport-specific drills.

To be released, the individual should have restored quadricep and hamstring strength to at least 90% of the opposite side. Muscular endurance should also be considered before returning to play, since findings suggest that the hamstring muscular endurance is reduced for at least one year, even with recovered strength, if not a focus of rehab. This could lead to a re-injury if the muscle is believed to be strong enough to accomplish a task but cannot handle the load as expected. [31]At each level, the needs of the individual should be considered and modifications should be made with any pain or

swelling, including potentially reassessment of their anticipated surgical outcome. Sport bracing is optional following rehabilitation.

There have been studies that investigated the return to play conditions following rehabilitation for athletes. One study followed 19 males and 6 females five years after ACL injury rehabilitation to analyze changes in their strength, stability, function, and sport activities. While no relationship between instability and functional activity score was found, a positive correlation between functional activity and functional ability score with the ability to participate in sports was found. Those who engaged in sports with more cutting and twisting motions were less successful in returning to re-injury participation levels and reported more pain, swelling, and instability.[115] This suggests that it is harder for those with cutting or twisting sport specific movements to return to their pre-injury functional activity scores, but those who reach higher functional scores after rehabilitation are more likely to participate in their sport.

Rehabilitation is the primary tool for returning to preinjury levels. It can be a very slow process and must be altered for the individual in order to be successful. The goal of ACL rehabilitation is to return the individual's range of motion, strength, and stability to preinjury levels and work to transition back into more sport specific tasks. If too aggressive, rehabilitation can stunt the healing of an injury or can reinjure the individual. However, it is important to challenge the patient for them to grow and succeed as well as gain confidence in their abilities. Additionally, it is important to understand the other repercussions of an ACL rupture to truly understand why it is important to prevent this injury and the true toll of the injury on the individual.

Repercussion of ACL rupture

The repercussions of an ACL tear can be very severe. While there are different rates of injury for athletes depending on the type of sport, their gender, or the prevention steps they took during training, ACL injuries largely affect athletes in common ways. Along with the obvious injury and monetary costs of an ACL tear, there are also long-lasting mental disparities that an ACL injury can have on an athlete.

For example, athletes face social losses due to sudden loss of sports involvement, potential loss of scholarship or related funding, they can face lowered academic performance, long term disability, mental strain from the recovery period associated with regaining full mobility, range of motion, and associated strength, as well as an increased risk of osteoarthritis.[19,52] MRI's taken following an acute ACL tear detected bone bruises and found the cartilage oligomeric matrix protein (COMP) was broken down. These findings suggest a major impact to cartilage and a strong potential for posttraumatic osteoarthritic lesions.[29]

In addition to the bone changes following an acute ACL injury, both men and women who face ACL injury are more likely to rupture the reconstructed knee following surgery. In addition, it has been found that a history of ACL reconstruction is a risk factor for a subsequent ACL injury on the contralateral knee.[16, 124] Also, long term studies report that there is a high incidence of joint degeneration (as much as 52-56%) within 12 to 13 years after surgery.[78] This suggests the toll of injury will compromise the individual for years beyond their initial injury and would increase the risk of subsequent injuries for that individual.

Finally, there can also be long term effects once returned to play. A case control study from the NBA found that 40% of their ACL ruptures happened in the fourth quarter of a match and, while 98% of those players returned to play the next year, there was a decline in performance that resulted in significant differences in games played per season when compared to control NBA players.[43]This suggests that, while athletes can be released for play following an ACL rupture, it is difficult for the athlete to return to previous levels of play, or overcome mental blocks from fear of rupturing their ACL again, which frequently causes a reduction in their performance.

In summary, the initial impact of an ACL rupture can have long-lasting effects on the individual. Reconstruction itself can have a heavy toll on the body as well as a large monetary burden. However, the repercussions do not stop there. The mental implications can be just as severe, if not more so, than the long-lasting physical toll. ACL injury results in a reduction in social interaction with teams, fear of losing one’s spot or position on the field, as well as a fear of repeat injury. It is important to monitor injured athletes and help them overcome these obstacles as well as take measures to prevent injury incidence.

ACL Injury Prevention Plans

An effective ACL injury prevention plan must be multifaceted and address the modifiable risk factors that can increase the likelihood of injury. Each individual requires a tailored approach and, as such, the professional implementing the injury prevention plan must adapt based on the starting physical abilities of the individual, the demands of the sport, and the relative timeline or periodization phase, among other factors. ACL

injury prevention programs need to address certain fitness domains in order to be effective: plyometrics and high intensity, dynamic balance, muscle strengthening, agility, neuromuscular training, and body awareness.

Plyometrics and high intensity exercises are often included in ACL injury prevention plans. One plyometrics based study found that by the end of a 6-week protocol, vertical jump had increased by 10% for participants and there was a significant effect on knee stabilization and the prevention of serious knee injuries. At the start, the female participants were weaker and exhibited less knee adduction/abduction moments and external knee extension and had imbalances between their sides.[54] Another study found that untrained females had a 3.6 times higher incidence of knee injury compared to trained females. While there was not a significant difference between trained females and untrained males, noncontact injury and incidence of ACL tears were significantly lower following plyometric training.[51]

High intensity and plyometric programs have been tested as screening tools for ACL injury, as well. A high intensity training program tested single-leg hops for 15 active females after training sessions. The participants experienced a reduction in their eccentric hamstring peak torque at 0, 15, and 30-minute intervals compared to the initial measurements. They also experienced a reduction in functional hamstring to quadricep ratio at 15 and 30 minutes after the session, greater extended knee angles at 30 minutes post, and increased knee internal rotation at 0 minutes after and 15 minutes after, which were significant. These findings suggest that this specific type of high intensity training program, and similar alternatives, would be a useful injury screening

test, since the program was able to classify and alter landing mechanics which are often associated with increased ACL injury risk.[18]

Another commonly included element for an injury prevention program is dynamic balance training. A three-year soccer-specific balance training intervention study, conducted on 24 elite female soccer players, found that protective balancing has been linked to a reduction in noncontact hamstring injuries, as well as ACL, patellar, and achilles tendon injuries. The study identified a dose response relationship between balance training and injury incidence but found no effect on contact injuries.[66]

A third training domain often included is muscle strengthening. Given a common mechanism of injury is believed to be quadricep hamstring ratio disparities favoring the quadriceps,[127] many studies have sought to find hamstring training programs that effectively increase hamstring strength. It is believed that a conventional quadricep to hamstring activation ratio less than .6, or a functional ratio less than 1.0, is predictive of increased injury risk. An NCAA study of 12 female soccer players found that 6 weeks of strength training with an emphasis on hamstring exercise significantly increased the functional ratio.[110]

The type of hamstring exercise performed impacts the primary muscle that is strengthened, so it is important to include a variety of hamstring muscle group exercises. For example, hip extensions emphasized the long head of the bicep femoris, while Nordic exercises, a partner based eccentric phase exercise, preferentially recruited the semitendinosus.[20] Another study, comparing a traditional hamstring curl program to a Nordic hamstring curl based program for trained soccer players, found that the Nordic

training resulted in a significant increase in the hamstring:quadricep ratio, which suggests that maximal eccentric hamstring strength would be a good training focus to reduce ACL injury risk.[85]

In addition to the type of hamstring exercise performed, the timing of hamstring protocols has also been investigated. A 12-week intervention included Nordic hamstring exercises before compared to after a soccer training session.[70] It found no significant difference in increased strength or surface EMG activity relative to timing of the program but found a difference in the architectural adaptation to support the strength gains that were observed. The before and after session Nordic hamstring intervention found an increase in bicep femoris muscle thickness and small pennation angle, compared to the controls. However, the study also found that the increases in bicep femoris length were greater in the Nordic exercise group that did hamstring exercises before soccer training. This increase in length of the hamstrings they described as an increase in sarcomeres which would allow for greater force production. While hamstring length alone has been investigated as a risk factor, the flexibility and coordination of the hamstring muscles activity in relation to the quadriceps has been found to be most directly related to ACL injury risk. Therefore, the additional Nordic hamstring work was beneficial for the soccer players, and the adaptations to the timing (either before or after the soccer training session) mainly differed by the architectural changes.

Finally, studies have also investigated the differences of hamstring muscular endurance compared to muscular strength training. A study randomly split 22 male soccer players up between two different types of hamstring conditioning interventions

for a period of 4 weeks. One hamstring conditioning intervention was focused on muscular strength and the other was focused on muscular endurance. It found that there was a clear decline in eccentric hamstring torque at longer muscle lengths during a 45-minute simulated soccer match. The short-term maximum strength-based intervention and the endurance hamstring intervention equally reduced the fatigue induced loss of strength during the 45-minute soccer simulation.[76]

While there are clear benefits to strength training interventions, there have also been successful injury prevention programs without the inclusion of strength training components, though insufficient research has been done to determine the true impact this factor, by itself, has on injury prevention.[49]

A fourth component that is often considered for ACL injury prevention plans is agility training. One 6-week agility training program that used 32 volunteers (16 male and 16 female) with 3 session days per week split its participants into 4 different training groups.[136] It found that the agility group had a significant improvement in spinal reflex for the lateral and medial quadricep muscles in response to anterior tibial translation, as well as a significant cortical response for the gastrocnemius, medial hamstring, and lateral quadricep response improvement. An isokinetic group from this study found a slowed medial hamstring and quadricep cortical response. The results suggest that, for muscle reaction to anterior tibial translation, only agility training was effective.

Another aspect of agility is conditioning. One study, which employed preseason conditioning for 42 players from a group of 300 female soccer players, ages 14 to 18,

found that all injuries during the subsequent season took place in the lower extremities, with 61.2% of them occurring at the knee or ankle. The trained group had significantly less injuries than the untrained group and, while not statistically significant, there was an overall decrease in ACL injuries for the trained group compared to the untrained group.[47]

A fifth component to consider including in an ACL prevention plan is neuromuscular screening and training. Based on a meta-analysis, all ACL injury reduction interventions contained at least some element of neuromuscular training. Therefore, neuromuscular training appears to be crucial in altering knee joint stabilization and aiding in the reduction of ACL injury rates in female athletes. However, all successful studies have also included plyometrics.[49] This suggests that neuromuscular training is likely insufficient, in isolation, to reduce ACL injury risks. Thus, it is important to have a multifaceted design.

Another study that focused on 18 high school females found that higher risk populations which underwent neuromuscular training had a significant decrease in ACL tear injury. This decrease, however, did not lower their risk levels all the way down to the levels present in naturally lower risk populations.[89] They also noted signs of higher injury risk in knee abduction during a dynamic landing, which would require increased training volumes and specific techniques to rectify.

Several studies have researched neuromuscular training programs, but with differing methodology and results. One 6-week neuromuscular training program, conducted 3 times per week for 60-90 minute sessions before a competitive volleyball

season, found that noncontact ACL injury risk was decreased by 72% compared to an untrained group.[49] Another neuromuscular training study, that primarily included agility, footwork, and minimal balance training for a random control trial, found no significant changes in injury risk. One neuromuscular screening test, assessing ACL risk among female handball and soccer players, found that noncontact ACL injury was often associated with reduced EMG pre-activity of the semitendinosus, and increased EMG pre-activity of the vastus lateralis, during side cutting.[143] This suggests that screening for these EMG activities prior to training will help identify players that are at an increased risk, allowing for more specific attention and focus in order to reduce this risk.

In contrast, very little research has been conducted in relation to body awareness and core stabilization as a means of reducing injury. These factors, though frequently included in injury prevention interventions, have not been the exclusive factor tested. Improved body awareness would theoretically allow athletes to move with greater efficiency and be more prepared for unplanned movements. Core stability would help improve posture and body muscular control. Having a weak core could result in unfavorable muscular activation patterns in response to perturbations, resulting in an increased risk of injury and a decrease in knee stabilization. Further research remains to fully understand what impact these factors have on ACL injury risk.

Finally, preventative biomechanics is a more recent trend that works to employ basic training techniques in manners clinically proven to both reduce the incidence of injuries as well as to enhance performance.[79] A study exploring this idea took place at a controlled lab and 624 females were classified among three groups of biomechanically

established risk profiles based on peak knee abduction moment. Those classified as moderate or high-risk groups had a greater likelihood of ACL tear than those in the low group.[50]

Another study attempted to use load-velocity tests to predict 1 repetition max values, which were used for load prescription in well trained men. They found that this technique was not accurate enough to be used for daily load prescriptions, but that it would be successful for general performance or for warm up exercises.[58] While biomechanics screening can aid in the design of an injury prevention plan, biomechanics testing as a whole has been shown to improve movement, but has not been established as a factor that can solely reduce injury risk.[49]

A prevention program will work to decrease the risk of injury but should not be expected to eliminate the potential of all potential injuries. Other factors, such as consistent compliance with the program, recovery, and lifestyle choices made by the individual that affect recovery are what ultimately determine the effectiveness of the program. In terms of recovery, it has been suggested that a 72-hour period is not long enough to restore homeostasis to the body for muscle damage and well-being status. Thus, training sessions must be structured to manage the load during that recovery period.[118] Further, relatively longer intrasession rest results in greater improvement in hamstring muscle strength during short term high intensity training.[100] Similarly, a cohort pro soccer player study showed that increasing the number of days between matches could lessen the risk of injury.[102] Given that knee extension strength increases while hip abduction and the hamstring quadricep ratio strength decreases from

prepubertal to pubertal stages, pre-adolescence is thought to be the optimal time to institute strength training programs aimed towards injury prevention.[102]

In conclusion, a prevention program has the greatest impact when it consists of multiple different types of training. The program benefits most from being at least 6 weeks long and with more than one session per week.[49] The sessions should include a combination of plyometrics, balance, and strength training to achieve optimal results. In addition, factors such as stretching, body awareness, decision-making, and core/trunk control are also beneficial for reducing risk factors, i.e. decreasing landing force, decreasing varus/valgus, and increasing effective muscle activation.[3] Including game play like distractions will also improve sport-specific adaptations.[67] While any training option will be more beneficial than not training at all, including isokinetic exercises results in higher motor performance and can have a specific athletic performance impact.[120] However, there needs to be a balance among cost, performance, and injury risk reduction. While all of these factors are important for risk reduction, performance enhancement must still be emphasized. The most successful interventions occurred during either pre-season or in-season.[49] The timing of the program helps shape the intensity. If in-season, there needs to be a balance between the program being challenging enough to maximize results, but not so intense that performance is negatively impacted.

Muscular Activation

Role of the Nervous System

The nervous system consists of two divisions: the central nervous system and the peripheral nervous system. The central nervous system includes the brain and spinal cord and primarily functions to process and interpret the data collected from the periphery. The peripheral nervous system controls sensory input (the gathering of information) and motor output (the response to the information gathered).

The most basic units of the nervous system are neurons, which can be multipolar, bipolar, or unipolar.[77] The general structure of a nerve cell has dendrites gathering information surrounding the cell body, containing the nucleus, mitochondria, the Golgi apparatus, etc., which then lead to an axon hillock. The axon hillock attaches the cell body to the axon. The axon transports the action potentials gathered from the dendrites. The axon has myelin sheath coverings which vary in number in order to either amplify or diminish the signal it received. Finally, the signal reaches the terminal branches and axon terminal, where it can be sent to other cells.

The neural control of muscles is somatic (meaning voluntary). There are motor units that have axon terminals at neuromuscular junctions, which allow for movement to take place. A motor unit is a motor neuron and all the muscle fibers it innervates. There are three main types of alpha motor neurons: S, FR, and FF. The S type is small but highly excitable. The FR type is big, averagely excitable, and more fatigue resistant. Finally, the FF type is the largest and is considered to have “low” excitability. These motor neurons are highly fatigable. Motor units are recruited from small to large and

receive common neural input according to the Henneman's Size Principle. While there are exceptions to this rule, it is the standard recruitment pattern for motor units.

Muscular contraction

Muscular contraction is the result of an interplay of several complex interactions. The basis of muscular contraction is rooted in the generation of an action potential, excitation contraction coupling, the sliding filament theory, and a length tension relationship.

Action Potential. The ionic equilibrium of the muscle cell sarcolemma creates a resting membrane potential around -90mV when not contracted. This homeostasis is maintained through an ion pump, which results in higher negative intracellular charge when compared to the external surface.[65] Activation of the alpha motor neuron anterior horn cell by either the central nervous system or by a reflex will result in excitation of the motor nerve. The generation of an action potential begins with a local depolarization called an end plate potential.[77] During this stage, acetylcholine (ACh) binds to ACh receptors at the neuromuscular junction. This binding opens a ligand gated ion channel where Na^+ rushes in.

The second stage is depolarization by generation and propagation of an action potential. This is spurred by the end plate potential transcending an action potential to the adjacent membrane areas. Voltage-gated ion channels in the membrane open, allowing for Na^+ to enter. Once a certain amount of the sodium ions follows the electrochemical gradient into the membrane, a voltage threshold can be reached, generating the action potential. The action potential is an all-or-nothing mechanism, so

it will not occur without the threshold breach, but once the threshold is met, the action potential will propagate in all directions from the neuromuscular junction. This further spreads the depolarization of the sarcolemma and opens more voltage gated ion channels. The propagation of the action potential along the sarcolemma and down the T tubules leads to excitation contraction coupling and the release of calcium ions.

In the final stage, the sarcolemma tries to restore homeostasis and its initial resting membrane potential by undergoing repolarization. During repolarization, the Na^+ channels close and K^+ channels open. As the potassium ions flow outside of the sarcolemma, the resting negative condition of the muscle fiber is restored. This usually results in a hyperpolarization of the membrane, which is then restored to the resting membrane potential.

Excitation Contraction Coupling. Excitation contraction coupling is a process where neural activation can result in the excitation of muscle protein interaction, which causes contraction. During this process, Acetylcholine is released from the axon terminal of a neuron and travels across the synaptic cleft to bind to receptors on the sarcolemma (muscle cell plasma membrane).[77] An action potential is then generated and travels down the T tubule. In response to the change in voltage, Ca^{2+} is released from the sarcoplasmic reticulum. The released Ca^{2+} binds to troponin-C and forms a cross bridge between actin and myosin. Acetylcholinesterase moves the acetylcholine from the synaptic cleft and then Ca^{2+} is transported back into the sarcoplasmic reticulum. Finally, the tropomyosin binds to the active sites on actin, causing the cross-bridge to detach.

Sliding Filament Theory. The main muscular component involved in contraction is thought to be represented by the sliding filament theory. Calcium release triggers muscle action. At rest, tropomyosin blocks the myosin binding sites on actin.[77] When the calcium binds to troponin C, there is a conformational change to troponin I that allows for the movement of tropomyosin. The myosin binding sites are exposed on the actin. After myosin heads reorient in response to the breakdown of ATP into ADP and Pi, the myosin heads can bind to the actin, forming a cross bridge. The myosin heads rotate towards the center of the sarcomere, causing a power stroke. Once ATP is present, the myosin heads bind to the ATP and detach from the actin crossbridge. This contraction cycle will continue as long as Ca^{2+} concentration is high in the sarcoplasm and ATP is present.

Length Tension Relationship. The tension the muscle generates varies according to how far the muscle is stretched and relies on adequate electrical stimulation from action potentials. The length tension relationship refers to an optimal overlap of the thick and thin filaments in the muscle fiber sarcomeres as a result of a slight stretch of the muscle.[77] If the muscle is overly stretched, then the filaments do not overlap and myosin heads no longer have anything to attach to in order to generate tension. If the sarcomeres are overly compressed, the thin filaments can begin to interfere with each other, and very little shortening can occur.

Electromyography

Electromyography (EMG) is a commonly used measurement system for recording the myoelectrical signals formed by physiological variations in the state of the

muscle fiber membrane.[65] In brief, EMG systems measure the action potential of the muscle cell membrane, described above. It reflects the muscle response to nerve stimulation. EMG is heavily incorporated and used in physiology, biomechanics, kinesiology, and other fields of study. Some of these usages include medical research, ergonomics, rehabilitation, and sports science. There are three primary types of EMG: subcutaneous, intramuscular, and surface.[108] Among them, intramuscular and surface EMG are the most common. The primary focus of this paper is on surface electromyography.

Intramuscular EMG. The most common form of intramuscular electromyography is fine wire. In this choice of electrode, paired fine nylon-coated wires are placed in situ by means of a hypodermic needle.[108] The wires are barbed (or have hooks at the end) to ensure the electrode is fixed in the muscle. This increases the likelihood that the electrode will follow along with the movement of the muscle fiber so that the recording area is the same. This type of electromyography is ideal for thin or deep muscles and for recording singular muscular activity. Other benefits of this type of EMG are the high level of sensitivity and the low concern for cross talk between muscles. However, this is a more invasive process where accurate placement is more difficult, and repositioning is nearly impossible. Additionally, minor discomfort for the participant can result, especially if movement is involved, as well as the potential for the wire to fracture under the dermis. This method also has a small detection area so, depending on the purpose of the study, this could be less ideal.

Surface EMG. Surface EMGs (sEMG) are more prevalent in studies, due to their noninvasive nature and simplicity in application. This type of electrode can be a wet gel or an adhesive gel, and generally contains silver or silver chloride solutions. Typically, surface EMGs appear as patches that can be stuck on the surface of the skin and have to be connected to an amplifier in order to send a signal. They offer real-time fatigue monitoring of muscles during performance of defined tasks, and they also provide insight into the correlation between biomechanical and physiological changes with fatiguing muscles.[24] However, given the origins of sEMG use with static positioning, transitioning to dynamic movements could provide inaccurate findings if sensors are not secured properly. Further, with sEMG, there is a greater chance of cross talk between muscles. Cross talk occurs when neighboring muscles produce large enough voltage changes that the resultant recording from the sEMG local site reflects some of the neighboring activity.[65] Since sEMG senses more general electrical activity, as opposed to the specific location where it is placed, this is often a downfall of the system. Crosstalk is usually counteracted by algorithms during analysis, allowing for control.

Application of sEMG is really important in ensuring findings are accurate and reliable. Prior to the placement of electrodes, the participants' skin must be prepped.[65] This involves shaving any hair that could be over the testing area, using sandpaper to gently remove excess dirt or dead skin, and then using sanitation wipes to clean the area. This preparation removes physical barriers that could potentially impede electrode signals or allow stronger electrode adherence to the skin. It is important to note the interelectrode diameter, as this plays a role in the placement. There are standard

electrode placement guides for muscles, and SENIAM is frequently used. It is standard to normalize findings, and one way to do so is by comparing the tested electrical activity to the maximal voluntary isometric contraction (MVIC) of the tested muscle. This normalization typically occurs after the data collection as a part of the processing and interpreting of the information. However, based on the purpose of the study and the protocol used, this could vary.

Many factors can contribute to the choice and effectiveness of the EMG type. Some, such as tissue type, tissue thickness, physiological changes, and temperature changes, can alter the EMG outputs and therefore must be considered or controlled for when applicable.[65] In addition, external noise in electrically condensed environments must be controlled, since the noise could alter the signal and validity of the information collected. The selection of electrode amplifiers also impacts the resultant electrical recordings. While the direct relationship between the electrical signals recorded and the muscular work is unknown, it is often assumed that greater amplitudes equate to larger muscular exertion. While contrary theories exist related to the impact fatigue has on muscular activity, it is important to note the change in the muscular electrical coordination pattern in order to get an idea of how the body is compensating for the fatigue.

The use of Surface Electromyography in Research

EMGs are commonly used because of the reliability and insight they offer. One study found EMGs to be a reliable method of assessing the reproducibility of both quadricep and hamstring muscle activation during ballistic and isometric exercises.[30]

There are conflicting views about the impact that exercise has on EMG activity. Some theories suggest prolonged exercise results in muscular fatigue, leading to a reduction in muscular activation. One factor that may contribute to this idea is perceived fatigue and reduction in effort by the individual as time increases. One study, which provides support for this theory, used a treadmill simulation of game play for amateur soccer players. It found that the EMG activity of the rectus femoris, bicep femoris, and tibialis anterior were lower than at the beginning, except for the gastrocnemius. This suggests that, even with prolonged intermittent exercises where work-rate is sustained, there is still an effect on muscle activity.[103] However, one other common (yet conflicting) theory, based on the size principle, is that increased periods of muscular activation will result in larger amounts of electrical activity as large motor units are recruited over greater periods of time.

The use of EMG can be very insightful if properly utilized, but there are a lot of limitations that must be clarified and understood to properly apply the information collected. As such, the findings from EMG studies are mainly useful to identify changes in the muscular activation in response to different protocols and researchers should be weary of using EMG findings as suggestive of fatigue in absence of other measurements.

Fatigue

What is Fatigue?

Fatigue is a concept commonly tested and incorporated into research. There are various interpretations of how fatigue manifests and, as such, many different definitions of fatigue and ways of measuring it. One source defines fatigue as a transient exercise-induced reduction in the ability to produce force or power.[19] However, one key aspect of fatigue which separates it from an injury is the ability to recover from the muscle weakness or damage with rest.[33] Thus, fatigue can be considered an observable reduction in performance of a task over time which can be reversed following a bout of rest. While the break time interval between actions or sets of an exercise will affect the resultant fatigue, the period of time associated with muscular recovery for a fatigued muscle group is believed to be 48 to 72 hours, depending on the size of the muscle group and the actions that elicited the fatigue. Fatigue can be produced by different internal mechanisms depending on the fatigue protocol or principle.[33]

Central and Peripheral Fatigue

Fatigue is often classified in two categories: central or peripheral. Central fatigue refers to an exercise-induced reduction in the ability to voluntarily activate muscle that arises from spinal and supraspinal factors.[19] Some of these factors include a suboptimal neural drive from the motor cortex, a reduced motor neuron firing rate, and an inhibition of spinal excitability due to muscle afferent input. Accordingly, central fatigue is established at the level of muscle activation.[28] Peripheral fatigue arises from changes that take place at or distally from the neuromuscular junction and is often attributed to

a reduction in the release of calcium ions from the sarcoplasmic reticulum or slowed Ca^{2+} reuptake.[19] Peripheral fatigue is more directly related to changes that influence contractile function.[28]

However, this model of defining fatigue as central or peripheral has been held by many to have two major limitations.[28] The first is the implicit assumption that adjustments in neuromuscular activity are necessary in order to counteract exercise-induced decreases in force capacity. To sustain task performance, other independent sensations or emotional states can affect force capacity and still be unrelated to neuromuscular deficits. The proponents for this limitation state that, while it is possible to isolate decline in contractile function in controlled recordings of electrically evoked forces, in voluntary settings, adjustments in activation signals discharged by motor neurons begin before the reduction in muscular force. In addition, sensory feedback can influence the integration of synaptic input of spinal motor neurons and also contribute to pain perception. This suggests that it is not possible to clearly identify etiology of fatigue by attempting to disassociate decline in muscular force from sensations about fatigue, especially during long-lasting contractions or contraction periods.

The second stated limitation is that most physiological processes that involve the performance of voluntary action can be challenged under the appropriate experimental conditions and, as a result, contribute to the development of fatigue.[28] Thus, there is a task dependency for fatigue that can result from the contribution that arises from the decline in performance at any phase of the muscle action complex. For example, a

decline in MVC from low intensity could be linked to the nervous system, but during a high intensity task it could also be the result of a decline in contractile function.

Fatigue with Performance

Very little is understood about the impact of fatigue on human performance. Some believe that there are two main reasons for this. First, there is an inability of current technology to accommodate the scope of conditions ascribed to fatigue.[28] This means that the scope of current technology cannot determine the full effect fatigue has on the body, so most information is based on assumptions or only presents one part of the full picture. A second issue is the paucity of validated experimental models.[28] While there is countless research available on fatigue, most studies fail to utilize models validated to elicit fatigue.

It is important to distinguish performance-based fatigability from perceived fatigability. Performance fatigability is defined as the decline in an objective measure of performance over a discrete period of time.[28] This will ultimately depend on the contractile capabilities of the involved muscles, as well as the capability of the nervous system to provide adequate activation signals from descending commands and ability to give afferent feedback for the prescribed task. A concept that affects performance fatigability is perceived fatigability. Perceived fatigability is the change in sensation that regulates the integrity of the performer and is impacted by the individuals' psychological state and attempt to maintain homeostasis. General fatigue in this setting would be defined as a disabling symptom in which physical and cognitive function is limited by

performance fatigability and perceived fatigability. The scope of fatigue is determined by the interactions between each fatigability attribute.

Physiology of Fatigue

There are many individual factors that contribute to the impact of fatigue, including: the fitness level of the individual, dietary habits, fiber type composition, and the intensity or duration of the exercise.[33] There are some current theories as to the physiological changes that facilitate fatigue, but more research needs to be done to fully understand the underlying processes.

One clear physiological change that has been observed in both animal models and in vitro human studies is the fatigue-related depression of skeletal muscle Na^+, -K^+, and -ATPase. Mice models allowed for isolated muscle preparations to clarify the role of Na^+, -K^+, and -ATPase. It found that the prevention of the rundown of transmembrane Na^+ and K^+ preserved membrane excitability.[35] However, attenuation of inhibition of this rundown by ouabain accelerated muscle fatigability and stunted subsequent recovery. Muscles paralyzed in a high K^+ solution saw a delay in muscle fatigability and accelerated recovery. This was observed in mice models, where reduction in gradients also decreased the soleus M wave and tetanic force. However, subsequent electrical stimulation or salbutamol-induced stimulation elicited marked recovery. While not a direct measure from human contracting muscle, there is speculation that the decline of femoral venous plasma K concentrations and rapid K^+ clearance from blood following exercise would similarly impact fatigue.

In the human study, eight resistance trained, eight endurance trained, and eight untrained individuals participated in a maximal in vitro 3-O-MFPase test.[35] Following a quadriceps fatigue test of 50 maximal isokinetic contractions, muscle biopsies were done on the vastus lateralis to measure changes in Na^+, -K^+, and -ATPase activity. Regardless of training type, it was clear that there was a decline in concentration of Na^+, -K^+, and -ATPase activity, suggesting that these factors are a determinant for fatigue and that there is an obligatory response to fatiguing muscular contractions. This study was not able to determine the underlying mechanisms for this change. However, other sources have considered changes in excitation-contraction coupling that cause the observable effects of fatigue.

Given that muscular contractions are the result of excitation-contraction coupling, all of the stages between action potential and binding of Ca^{2+} to troponin, to the reuptake of Ca^{2+} by the sarcoplasmic reticulum following the development of tension, are going to be affected by fatigue, as well.[101] While the exact mechanism of this impact in not clearly defined, it is believed to result due to: the failure of Ca^{2+} to release, the reduced Ca^{2+} sensitivity of contractile proteins, the failure to conduct an action potential in T tubules, or a reduced Ca^{2+} pumping ability of the sarcoplasmic reticulum.

One study followed the changes in Ca^{2+} in an isolated single fiber of mouse skeletal muscle during a low-frequency fatigue.[133] During low levels of fatigue, there were substantially reduced tetanic Ca^{2+} levels at all stimulus frequencies, but there was no change in Ca^{2+} sensitivity or max Ca^{2+} activation with tension. They contributed to

reduction in tetanic Ca^{2+} to a reduction in Ca^{2+} release from the sarcoplasmic reticulum. This suggests that there was not a failure in the conduction of the action potential in the T tubules, since there was no gradient change, making Ca^{2+} uniform across the fiber. Researchers also measured sarcoplasmic reticulum pumping function to see if this was a potential factor contributing to the change in Ca^{2+}. However, they only found minor slowing of Ca^{2+} re-uptake following the low frequency fatigue. So, it was concluded that sarcoplasmic reticulum pumping function is not the main cause for the reduction of tetanic Ca^{2+}.

These findings, however, must be limited in generalization, because it was an animal model and may not directly correlate with humans, higher frequency fatigue could potentially elicit a different physiological response, and this was done in an isolated muscle fiber, so the muscle fiber type could have affected the response, and it cannot be assumed that voluntary active muscles respond in the same physiological way. But, this does provide meaningful insight as to potential mechanisms of change with fatigue.

Some studies have focused on the crossbridge cycle changes that follow fatigue. Generally, strongly bound cross bridge states are thought to peak during maximal isometric contractions, since the rate of transition from low to high force state is thought to limit the peak rate of force development.[33] With prolonged endurance and high intensity exercise, research suggests a decline in twitch and tetanic force, max shortening velocity, and peak power will result due to fatigue. The post fatigue twitch has been seen in studies to have had a reduction in twitch force, prolonged contraction

periods, and half-relaxation times. This leads to reductions in peak rate of tension development, suggesting a reduction in amplitude and an increase in the duration of intracellular Ca^{2+} transient. However, these findings relate Ca^{2+} as a more direct effect with fatigue than cross bridge modifications.

It has been argued that training mechanisms can enhance resistance to fatigue. Specific activity training regimens can modify the muscle sarcoplasmic reticulum and, in turn, affect Ca^{2+} regulation and Ca^{2+}, -ATPase inactivation with exercise training.[35] The increased endogenous antioxidant enzymes as a result of exercise are believed to have a protective effect on Na^{+}, -K^{+}, and -ATPase activity.

In conclusion, while not completely understood, there are underlying physiological changes that temporarily occur during the presence of fatigue. While greatly impacted by the frequency of how the fatigue manifested, it is clear that changes lie in Na^{+}, -K^{+}, and -ATPase activity, cross bridge cycling, and Ca^{2+} interaction. There are limitations for measured levels of fatigue where only part of the manifested changes are presented or the perception/motivation of the subject affects the response.

Research on Fatigue and ACL Injury

Given the frequency of ACL injury in sports, research has increasingly focused on the effects fatigue can have on the individual, as well as the impact it can have on ACL injury. However, in both males and females, fatigue response is highly dependent on the muscle fatigued.[63] One study suggests that there is a linear relationship between fatigability and the percentage of fast twitch fibers. They claim that the more fast twitch fibers present, the more likely the quadriceps are to tire out faster, and slow twitch

activation was more likely to be sustained in men.[128] As a result, the protocol to induce fatigue as well as the testing measures used to quantify fatigue-induced changes will play a large role in the findings of the study. Some of the most common focuses of research have been the implications of fatigue on landing, side-step cutting, proprioception, neuromuscular control, and torque.

Implication of fatigue on landing

One of the primary ways that noncontact ACL injury occurs in sports is during the landing. To address this, numerous studies have focused on the implications of fatigue on the ability for the individual to land with “normal” mechanics. Fatigue has regularly been seen to result in abnormal landing, which can increase ACL injury by reducing and delaying muscular activation, especially in response to destabilizing perturbations.[127]

In biomechanics settings, a Landing Error Scoring System (LESS) is commonly used to evaluate landing technique. One study, comparing male and female landing following a functional exercise program, found that both sexes showed poor landing technique after the exercise protocol, women to a greater extent than men, suggesting fatigue adds to injury risk regardless of sex.[132] However, one study’s review of the LESS as an objective scoring system found it reliable, but only moderately valid when used with motion capture, which suggests that it needs more improvement before it can be used to predict ACL or other noncontact lower extremity injury.[42] While this study generally addressed how landing changes and is affected by fatigue, other studies have identified more specific changes in both double and single-leg landings following fatigue protocols.

Double-leg landings. Double-leg landings allow for a greater base of support and more surface area to absorb the forces from landing. It is a commonly tested landing style due to its increased stability and safety for the individual.

A 2006 study tested 15 active males on drop jumps from 30 cm high and 50 cm high, with and without fatigue.[86] There was a significant difference in the landing mechanics from the 30 cm drop following fatigue, but not from 50 cm. They concluded that fatigue can result in a reduced capacity to attenuate the impact of accelerations on the tibia, which could result in injury. They attributed the nonsignificance at the higher drop to the increased amount of acceleration, which made the landing forces less impacted following fatigue.

While that last study focused on the differences in landing based on height, most studies have focused on more specific changes in lower extremity upon landing. It has been found that, especially in females, landing following a closed chain exercise protocol had increased flexion velocity, knee joint abduction angles, and delayed lateral hamstring activation.[38] In addition, they noted that fatigue led to a reduction in preactivation of the medial and lateral hamstring, as well as gastrocnemius in both males and females. This suggests a different knee joint control exists for females, as opposed to males, due to different neuromuscular strategies, knee flexion velocity, abduction angle, and muscle activation. Other controlled lab studies have concluded similarly.[23, 80] Finding that, females showed greater initial ankle plantar flexion and peak stance ankle supination, knee abduction, knee internal rotation, and smaller ankle

dorsiflexion than male counterparts. Fatigue resulted in increased initial and peak knee abduction and internal rotation in both sexes, but it was more pronounced in females.

A study of 20 recreationally active students also tested double leg landing biomechanics following fatigue. It concluded that, while there were no differences in frontal vertical ground reaction force, isolated bilateral hip-abductor fatigue altered frontal plane lower extremity orientation, leading to an increased risk of valgus.[22]

Thus, following fatigue, there was clear changes in double-leg landings for both sexes, but more pronounced in females. While height of the landing does affect the magnitude of change, knee abduction and internal rotation as well as preactivation of the hamstrings and gastrocnemius have been noted changes in landing following fatigue.

Single-leg landings*.* While double leg landings are safer for the individual, single leg landings occur in play settings all the time. In fact, most ACL injuries occur as a result of single leg landings. Thus, the impact of fatigue on this style of landing is very important to researchers as well.

One study tested 25 NCAA female athletes' unanticipated and anticipated single-leg landings following a squat and jump sequence-based protocol.[17] Fatigue resulted in significant increases in initial contact hip extension and internal rotations, and in peak amounts of knee abduction, internal rotation, and ankle supination angles. These findings were more pronounced in the unanticipated landings, suggesting that, when combined, fatigue and decision making inhibit performance and could result in an overload in central control mechanisms, leading to injury. In contrast, a fatigue protocol

utilizing a bike ergometer no statistically significant difference for peak knee flexion angle, peak knee flexion acceleration, peak knee adduction/abduction, knee velocity, or acceleration.[122] This study concluded that fatigue only decreased the ability to attenuate shock by increasing angular velocity in the direction of the knee flexion for the single leg drop landing. This result can be attributed to the cycling mechanics which involves more quadriceps activation an resultant fatigue than hamstring.

A 2011 study tested the effects of hip adductor fatigue on single leg landings for 20 active women.[98] After repetitive side lying hip adduction, there was a decrease in peak external knee addiction by 27%, and a decrease in peak hip adduction of 24%. However, there were no clear indications of hip abduction fatigue being unfavorable to ACL. Thus, this suggests that further research is necessary to determine the role the hip abductor plays in protecting the knee during landing. The only noticeable change from this study was a delay in glute medius activation in single leg landing following the protocol, and there were no significant findings in peak or integrated signals with landing.

Finally, one study tested the single leg hopping ability of 13 healthy men following 2 sets of 50 step ups.[95] It suggested that potential ankle compensations will occur in single leg landings to compensate for knee weakness following fatigue.

In summary, single-leg landings occur all the time in sport. Due to their complexity and small base of support, the likelihood of injury following improper landing mechanics is greatly increased. Studies have look at muscular and activation responses when completing single leg landings following various fatigue protocols.

There are potential changes in ankle response, knee protection responses, and hip responses following fatigue which increase the likelihood of ACL injury.

Implications of fatigue on Side-step cutting

Cutting is another common movement often associated with ACL injury. Many studies have sought to determine whether fatigue makes it more likely to have an injury while performing this movement. A 2009 study examined the knee mechanics during cutting, for 15 recreational athletes ages 22-36, directly after, 20 minutes after, and 40 minutes after a fatigue protocol.[129] It concluded that even 40 minutes was insufficient to return knee mechanics to pre-fatigue measures, and that the internal rotation angles and internal knee adductor moments were significantly higher post fatigue.

A cutting program assessed by female athlete EMG and force plate data following a simulated handball game found maximal voluntary contractions for the quadriceps and hamstrings decreased. During the side cutting task, there was a selective decrease in hamstring neuromuscular activity.[144] This seemed to suggest that an impairment in ACL agonist muscle (the hamstrings) during cutting occurred, in response to the fatigue, as a result of the handball game. This further supports the change in strength ratio for the hamstrings and quadriceps which is often observed in females and linked to ACL injury risk.

While very little research has been done on this topic, the current findings suggest following fatigue there is an impairment in the ACL agonist muscle, the hamstrings, during cutting which can be impaired for over 40 minutes following completion of the fatiguing task.

Implications of fatigue on proprioception

There are conflicting views as to the effect of fatigue on proprioception. One study that focused on knee joint proprioception following a cycle ergometer fatigue protocol saw only minimal, non-significant, changes in squat.[69] However, a 1999 study of collegiate male and female soccer and basketball players found a decrease in the ability to detect joint motion leading into extension following a Biodex, isokinetic induced, fatigue procedure.[107] They also observed an increase in the onset time of contraction for the medial hamstring and lateral gastrocnemius, and an increase in the first contraction area of the vastus medialis and vastus lateralis (mainly in men). They concluded that both males and females have a decrease in proprioceptive ability and face alterations in muscular activity following muscular fatigue. More information is needed to fully understand the implications of fatigue on proprioception and how that could impact ACL injury risk.

Implications of fatigue on neuromuscular coordination

Neuromuscular coordination is a risk factor for ACL risk, so identifying changes that occur due to fatigue gives insight into preventative training that could be beneficial.

One test of neuromuscular coordination is the ability to respond to planned and unplanned perturbations. Twenty female NCAA athletes were evaluated in their ability to respond to light, in planned and unplanned movement, following a unilateral fatigue protocol in a 2009 study. They found cross over fatigue to the contralateral limb during single leg landings, suggesting a shift in neuromuscular control due to central fatigue.[81] Another lab experiment of 16 controls and 17 post ACL subjects found that both groups

had impaired strength, central activation, and biomechanics following a fatigue protocol.[42] This suggests that neuromuscular fatigue can increase ACL risk and is not limited in impact to only specific populations.

Even in isokinetic testing, there were clear changes in EMG muscle coordination for 14 subjects. The coactivation of the vastus medius was greater than the vastus lateralis, and the bicep femoris was greater than the medial hamstring, but the only significant velocity of coactivation was present in the bicep femoris. This suggests that fatigue can alter the motor patterns of these muscles.[84]

Quadriceps and Hamstring ratio. One commonly noted muscular coordination difference between males and females is the ratio of quadricep to hamstring activation. Females have a tendency to over-activate their quadriceps (compared to their hamstrings), putting their knee in a compromised position. In fact, due to this sex-based difference, some studies that exclusively looked at males did not note any muscular coordination differences following fatigue.[104]

Females and male athletes underwent a repeated squat to fatigue protocol, and that compared the change in hopping ability for the participants before and after fatigue.[97] They found that both males and females had a reduction in hamstring and anterior tibialis activation after fatigue, as well as an increase in quadricep hamstring ratio and gastrocnemius anterior tibial coactivation ratio. The females had a greater quadricep to hamstring activation ratio than the males had at any point and had increased knee flexion at the initial contact of hopping following fatigue. This study noted that the quadriceps hamstring ratio is a risk factor for females and that ankle

strategies become more dominant when the knee becomes compromised due to fatigue. However, in 2010, Thomas et al. proposed that there was no clear sex by fatigue effects, and supported the idea that fatigue alters quadricep hamstring mechanics, producing a significant increase in hip internal rotation, knee extension, and knee external rotation angles.[127]

It has been suggested that submaximal fatigue of the hamstring could lead to anterior tibial translation and decreased knee stability.[82] Thus, an increased hamstring fatigue rate, overactivation of quadriceps, or a delay in coactivation is likely to add to injury risk. Hamstring failure was attributed to reduced motor activity, rather than a latency in the hamstring response. Another study found that the recruitment order of lower extremity response to anterior tibial translation did not differ with fatigue. However, there was an increase in tibial translation of 32 percent, as well as an increase in displacement with fatigue, that it attributed to the cortical level activity. Thus, they suggest that fatigue with pathomechanics for men and women results in a less stable knee and an altered neuromuscular response.[138]

In contrast, a study that focused on eccentric quadriceps fatigue found delayed vastus lateralis, vastus medialis, and rectus femoris activity, when compared to controls, and no difference in medial hamstring or biceps femoris activation.[92] In addition, they noted early gastrocnemius activation with quadriceps failure, suggesting that the gastrocnemius also plays a protective role in the knee.

As previously mentioned, the quadriceps hamstring activation ratio has been identified as a risk factor in females. These studies suggest that the presence of fatigue

increases this sex-based activation difference that can result in anterior tibial translation and knee instability.

Sprint Mechanics. Sprinting is common in most sports and often results in fatigue. Repeated anaerobic sprint tests (RAST) were used to determine alterations in stride mechanics and leg-sprint behavior in 8 pro soccer players in a 2015 study.[15] It concluded that there was a relationship between repeated sprint ability and changes in neuromuscular activation, as well as muscle de- and re-oxygenation rates. There was a clear decrease in performance over time, as sprint times and total stride durations increased. This study found a large reduction in the root mean squares EMG activity in the rectus femoris and biceps femoris, but not in the vastus lateralis. However, there was a substantial and rapid decrease in the vastus lateralis muscle oxygenation levels during the initial sprint repetition as de-oxygenation exceeded re-oxygenation. Muscle de-oxygenation rates progressively declined during successive sprint repetitions, while muscle re-oxygenation displayed an opposite pattern.

Rampinini et al. evaluated sprint performance following a 90-minute soccer match.[106] It observed a clear reduction in maximal voluntary contraction of 8%, EMG amplitude reduction of 12%, and sprint performance, but noted that short-passing ability was preserved. It took 48 hours for the individuals to return to baseline numbers following the game. This study concluded that, in general, there was central fatigue that hindered the MVC and sprint performance, but peripheral fatigue contributed with muscle soreness for mechanical response indicators at lower frequency levels.

However, a 12 female soccer player study in 2008 found no sagittal or frontal plane kinetic or kinematic changes in cutting following a fatigue protocol.[110] Their protocol consisted of a 60 minute shuttle run, with fatigue being determined by changes in countermovement jumping power. While there was fatigue and knee internal rotation at stance and hip, ankle, and knee external rotation at foot touchdown, they suggested that a threshold of fatigue before function is affected. This threshold limit has not been determined or conclusively researched.

Implications of fatigue on torque

A few studies have addressed the potential changes in torque that result from fatigue. One study of 8 amateur male soccer players was completed in 2014.[74] It tested the central motor output and hamstring fatigue following a simulated soccer match, and found that there was a max torque reduction by minute 45 of 7.6 $\pm$ 9.4%, and that the rate of torque development reduced after 15 minutes by 29.6 to 46.2%. While there was no clear hamstring fatigue, there was a minor reduction in the max bicep femoris EMG. It concluded that central mediated reduction of max torque and rate of torque development are factors for hamstring injury mainly in the first half of a match, and the declined activity of the bicep femoris in the second half could result in ACL injury.

Spendiff et al. studied the torque-velocity relation in muscle. They concluded fatigue-inducing exercise at low velocity led to reductions in peak and mean torque in subsequent exercise at higher velocities. The greater decline in torque during subsequent exercise at high velocities could be attributable to the greater exhaustion of type 2 fibers, and low velocity is less affected due to its greater use of type I fibers.[121]

While limited in scope of research, studies suggest a change in torque following fatigue. It is believed the decline in toque is related to the muscle fibers effected.

Basketball fatigue protocols for females

In the United States, basketball has the second highest rate of ACL injury, following soccer. Since soccer is a more global sport and has the highest ACL rate of injury sport, the vast majority of fatigue protocols have been established to simulate soccer load. Very few studies have focused specifically on basketball. More specifically, there have not been many validated basketball simulating protocols used to test fatigue. In most settings, the researchers are relying on actual game play, lab designs using neuromuscular electrical stimulation or isokinetic dynamometers, or basketball-based resistance exercise fitness testing.

The Basketball Exercise Simulation Test (BEST) was designed in 2011 based off of the competition requirements of elite and sub-elite Australian male basketball players.[111] It claims to be the first of its kind and simulates the load of one quarter of a game. A 2012 study found the BEST reliable and valid for males.[113] However, this has not been validated in females. In addition, this protocol was not established directly to be a fatigue inducing protocol and is generally used to determine fitness level. Given the increased risk associated with female athletes, there is a limitation in research where there is a lack of protocols available to apply to this population. For female basketball players, they are both limited by the prevalence of valid basketball simulations and the applicability of the simulations to female basketball athletes.

Conclusion

Noncontact ACL injury risk is believed to be greatly modifiable in nature. Female basketball athletes make up one of the most affected populations by this injury in the United States. However, very few studies have focused on this cohort and when studied have used generic protocols that have not been validated for this population. We plan to determine whether the BEST elicits fatigue for active females and to determine the changes in lower extremity skeletal muscle kinematics and activation following the BEST.

CHAPTER 2

METHODS

Study Design

This study will implement a repeated measures design where each participant will complete two randomly ordered sessions, a control and an intervention session. The control session will consist of sedentary tasks in order to determine the true effects of the protocol and the potential presence of fatigue.

Participants

Six active females will be recruited to participate in this research study. Due to the increased risk of ACL tears in females, as analyzed in the literature review, only females will be recruited for this study. In order to qualify for this study, participants must be between the ages of 18-30 years, have no current limitations that would inhibit them from performing the basketball exercise simulation protocol, and must participate in moderate to vigorous cardiovascular exercise for 4 or more hours each week. In addition, participants with prior history of ACL tears or lower extremity injuries will be excluded.

Setting

All testing and data collection will occur in the Solomon Blatt Physical Education Center at the University of South Carolina (1300 Wheat Street, Columbia SC 29208).

Specifically, the basketball courts on the third floor of the Blatt Center (Blatt 305 and 308) will be utilized, based on availability, for all equipment and data collection. The order of the sessions will be randomly selected, with at least three days but no more than two weeks between the sessions. The Basketball Exercise Simulation Test (BEST), a validated basketball simulation protocol by Scanlan et al., will use cones and signs on the courts as well as court lines (2014).[112] A drop jump exercise has been chosen for measurement in order to standardize the test and maintain safety for all participants. In addition, a drop cut has been chosen for measurement to assess differences in cutting movements following the protocol, since ACL injury more commonly occurs with cutting or landing movements.

Design and Procedures

The experimental sessions will include a control session and an intervention session. The intervention or simulation session will employ the Basketball Exercise Simulation Test (BEST) protocol. The Best procedures, shown in the diagram below, consists of circuits of basketball specific movement patterns (jogging, sprinting, jumping, shuffling, and walking) in a specified order that mimic a basketball game. The BEST will simulate a collegiate Women's basketball with 4 10-minute quarters. A circuit will require approximately 30 seconds. Accordingly, a participant will complete a maximum of 20 circuits per quarter. The completion of a quarter will be defined as either 10 minutes of continuous exercise or the completion of the 20 circuits, whichever comes first. Participants will be given 3 minutes of rest after the first and third quarters, and

10 minutes after the second quarter. Between quarters participants complete tests discussed below.

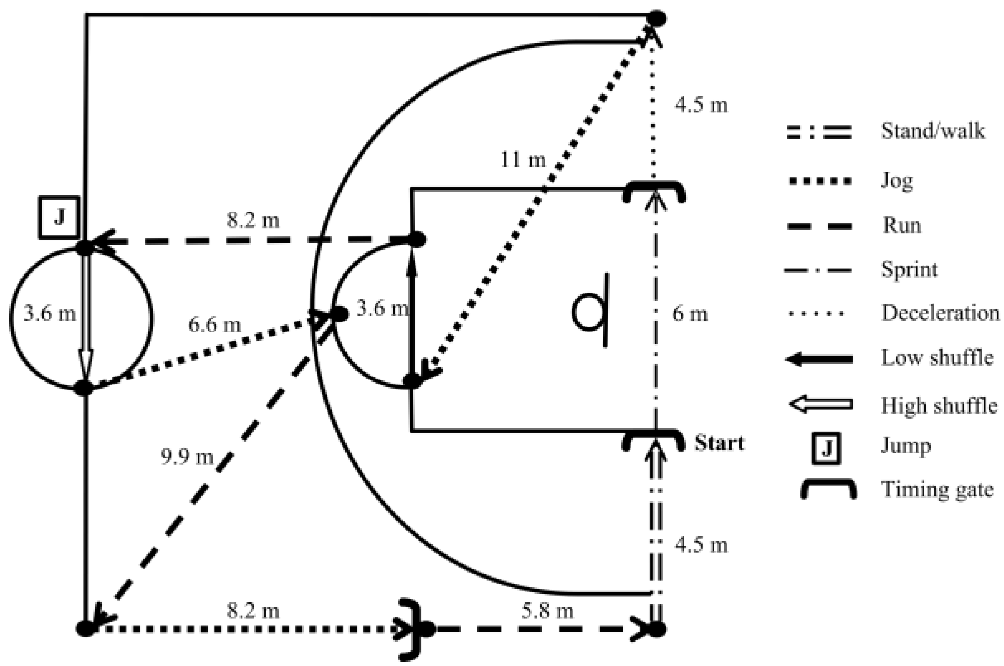


Figure 2.1 BEST Protocol Diagram by Scanlan et al. (2014).[112]

The control session will mimic the timing of the simulation sessions. However, participants will not complete the BEST protocol. Participants will be given the opportunity to perform the warmup prior to each testing session on the control day.

Each session will consist of a five-minute directed dynamic warm-up that includes practice attempts prior to the start of data collection. The practice will include 3 vertical jumps (with 2 practice jumps), 3 drop jumps, 3 drop cuts to the left and to the right, and 3 single leg triple hops for each leg. Data collection will consist of multiple testing periods: prior to the intervention (Pre), at the end of each quarter (Q1, Q2, Q3), and immediately after the last quarter of the game simulation (Post). The Pre- and Post-

test period will include 3 vertical jumps, 3 drop jumps, 3 drop cuts both directions and 3 single leg triple hops for each leg. The Q1, Q2, and Q3 test periods will only include 1 vertical jump and 2 drop jumps. Briefly, the specifics of the testing tasks will be described below.

Vertical jumps: The vertical jump height will be measured using a vertec device. The device will be adjusted to the vertical reach height of the participant where the bottom slat of the vertec is barely touching the tip of the middle finger of the dominant hand of the participant when they are standing directly below the device. The participant will stand with her dominant hand underneath the slats of the vertec and will be instructed to squat down and "jump as high as possible." The point on the device that the participant reaches will be recorded for each trial. The slats will be reset before each trial.

Drop jumps: The participant will be instructed to stand on a 30cm box. She will step off with the foot of her choosing (but be asked to step with that same foot for all subsequent trials) and land on both feet on the ground in a knee bent, squat position. Once making contact with the ground, she will immediately push off the ground, jumping vertical in the air.

Drop cuts: The participants will be instructed to stand on a 30cm box. She will step off with the foot of her choosing (but be asked to step with that same foot for all subsequent trials) and land on both feet on the ground in a knee bent, squat position. Orange cones will be placed seven feet away in both directions. The participant will be instructed to get to the cone as quickly as possible after stepping down from the box.

Once making contact with the ground, she will immediately push into the ground and cut to the left or right. There will be three trials to the left and three to the right, alternating sides between trials.

Single leg Triple hops: A 15-meter stretch of a ruler will be placed on the ground. The individual will start at the starting point on one leg and will jump as far as she can with three broad jumps on that single leg, while keeping her hands placed on her hips. At that third landing, the participant will have to hold her balance and positioning for 10 seconds in order for the trial to count. The total distance reached will be recorded.

Instrumentation and Analysis

Two-dimensional Kinematics: Two HD video cameras will be used to collect two-dimensional videos to input in Dartfish Team Pro software, Version 7.0, which will be used for the kinematic analysis. Both cameras will be utilized during the Drop Jump task. This allows for a frontal view of the movement to assess knee valgus or varus. The secondary camera will provide a sagittal view of the Drop Jump movement to assess changes in ankle and knee angle as well as relative trunk position at different stages of the movement. In particular, the focus will be on six phases for the Drop Jump task: initial stance, initial contact with both feet, lowest point, take off, initial second landing, and stabilizing point.

Only one video camera will be utilized during the Drop Cut task. Given the demands of the task, a side camera position could become a potential safety hazard for the participant. The primary measure for the drop cut task will be the time from the initial landing to initial contact of the first step towards the cone.

Vertical Jump measurement: In order to determine the presence of fatigue following the exercise protocol, a vertec jumping apparatus will be utilized. Vertical jump height is often used to determine anaerobic power. A Vertec is a commonly used functional assessment for force output due to the relatively low cost, the ease of use and consistency, and the low risk of injury for the participant. Most college-aged individuals would have completed a task, such as a vertical jump beforehand or are at least be capable of performing a vertical jump, making this measure applicable for the population. Applying the 20% rule, the participants' vertical jump measurements decrease by at least 20 percent when compared to pre-exercise values.

Statistical Data Analysis: All statistical analysis will be conducted with an alpha level set at, $p \leq 0.05$. There will be a one-way repeated measures analysis of variance for time and a two-way analysis of variance with repeated measures on both factors of intervention and time.

CHAPTER 3

RESULTS

Anthropometric and Protocol Measures

Six active female participants, ages 23.83 ± 3.25 years, completed this study. See Table 1 for demographics. Each of the 6 participants completed a control and an intervention session that consisted of testing periods, four 10-minute quarters, and brief rest periods. There were no dropouts in this study.

The intervention session required participants to complete four 10-minute rounds, or 20 laps, whichever came first, of the BEST protocol. Individual lap times were recorded for all participants as well as the number of laps completed. The overall participant average lap time for the first round was 35.63 ± 1.21 seconds and the overall participant average lap time for the final round was 36.92 ± 1.63 seconds. While there was a general trend towards an increase in time, the difference was not statistically significant. The number of laps completed had a general negative trend from an average of 16.67 ± 1.75 laps completed in the first round to an average of 15.83 ± 2.23 laps completed in the final round. This was found to be statistically significant, $p=0.02$, $\eta^2 = 0.467$ (Figure 3.1).

Rate of Perceived Exertion is a subjective measure on a scale from 6 to 20 for the participant to relate her level of exhaustion and ventilation. A six on this scale represents minimal effort, like that of resting, and a 20 represents an absolute maximal

effort. The participants were asked to relay their RPE at the completion of each round for the control and intervention sessions. The individual RPEs for the control session stayed fairly consistent and did not greatly fluctuate among rounds. The average RPE for all participants was around 7 for all four rounds of the control session. There was a slight positive trend for RPE in the intervention session, with more individual variation in response. The first round RPE was found to be 15.33 ± 0.67 while the final round RPE was 17 ± 0.37. A significant interaction was found for the intervention with a p value of 0.004 (Figure 3.2).

Heart rate was a physiological measure taken at one-minute intervals during the four rounds of the control and intervention sessions. The percent of age-predicted maximal heart rate for the round average heart rate indicated the relative intensity of the round. In the control session, there was a negative trend in percent of max heart rate from 50.94 ± 6.05 bpm in the first round to 44.35 ± 3.64 bpm in the final round. The overall percent of max heart rate was higher for the intervention session, while time and heart rate had a positive relationship. The first-round average percent of max heart rate for all participants was 83.12 ± 10.12 bpm while the final round average percent of max heart rate for all participants was 88.38 ± 9.02 bpm. There was a statistically significant interaction with the intervention and percent of max heart rate with a p value of 0.011 (Figure 3.3).

Testing Measures

Study participants completed Pre- and Post-testing consisting of three maximal vertical jumps, 3 single leg triple jumps on each leg, 3 drop jumps, and 3 drop cuts to the

left and to the right. In addition, three intermediate tests consisted of one maximal vertical jump and two drop jumps.

The vertical jump test was used to assess maximal jump height as well as the presence of fatigue. The average vertical jump height in the control session dropped from a Pre-test value of 37.11 cm to a Post-test value of 35.98 cm. The average beginning vertical jump height was lower during the intervention session, 34.76 cm, and reduced to a lesser extent for the intervention Post-test, for an average vertical jump height of 33.94 cm. The BEST did not significantly affect maximal vertical jump height, and no difference was found between control and intervention jump height. However, was significant, the effect of time on vertical jump height with a p value of 0.0007 (Figure 3.4).

Single leg triple jumps are a part of the ACL reconstruction return to play protocol used by athletic trainers. This measurement looks at bilateral symmetry of jump distance with the right and left leg as well as differences in jump distance following exertion. A reduction in twenty percent of maximal distance from Pre-testing values to Post-testing values is indicative of ACL instability and injury risk. There was no statistically significant difference in single leg triple jump as a result of the intervention ($p>0.05$). Time had a significant effect on left leg single leg triple jump distance with a p value of 0.01 (Figure 3.5).

The drop jump is a commonly used exercise to assess changes in landing and jumping mechanics. This movement was analyzed for relative trunk angle, knee angle, and ankle angle at six stages of this movement. Those stages include the initial stance

on the box, the initial contact with both feet on the ground, the lowest point of the squat position, the final take off point from the ground, the initial landing point from the jump, and the stabilizing point or lowest point when returning to the ground. Within each phase, the average joint angles were consistent, i.e. not significantly different, for the control when compared to the intervention at all testing times (Pre-, Break 1, Break 2, Break 3, and Post- testing) for the trunk, knee, and ankle. Only a significant interaction with the intervention for the trunk angle at take-off compared to the control session with a p value of 0.0073 (Figure 3.6, Figure 3.7, and Figure 3.8).

The final testing measure was based from the frontal view of a drop cut. During this movement, the participants stepped off a box and were instructed to touch a cone placed laterally 7 feet away. One cone was placed to the right of the starting location and another placed to the left, while the participant alternated sides between trials. The time to first step towards the right direction was found to be statistically significant with a p values of 0.04, but there was no statistical significance found for this measure towards the left direction (Figure 3.9).

Table 3.1: Participant Demographics

Gender	Female
n	6
Age (years)	23.83 ± 3.25
Height (cm)	164.20 ± 8.33
Weight (kg)	61.97 ± 11.84
Vertical Reach (cm)	215.69 ± 11.62

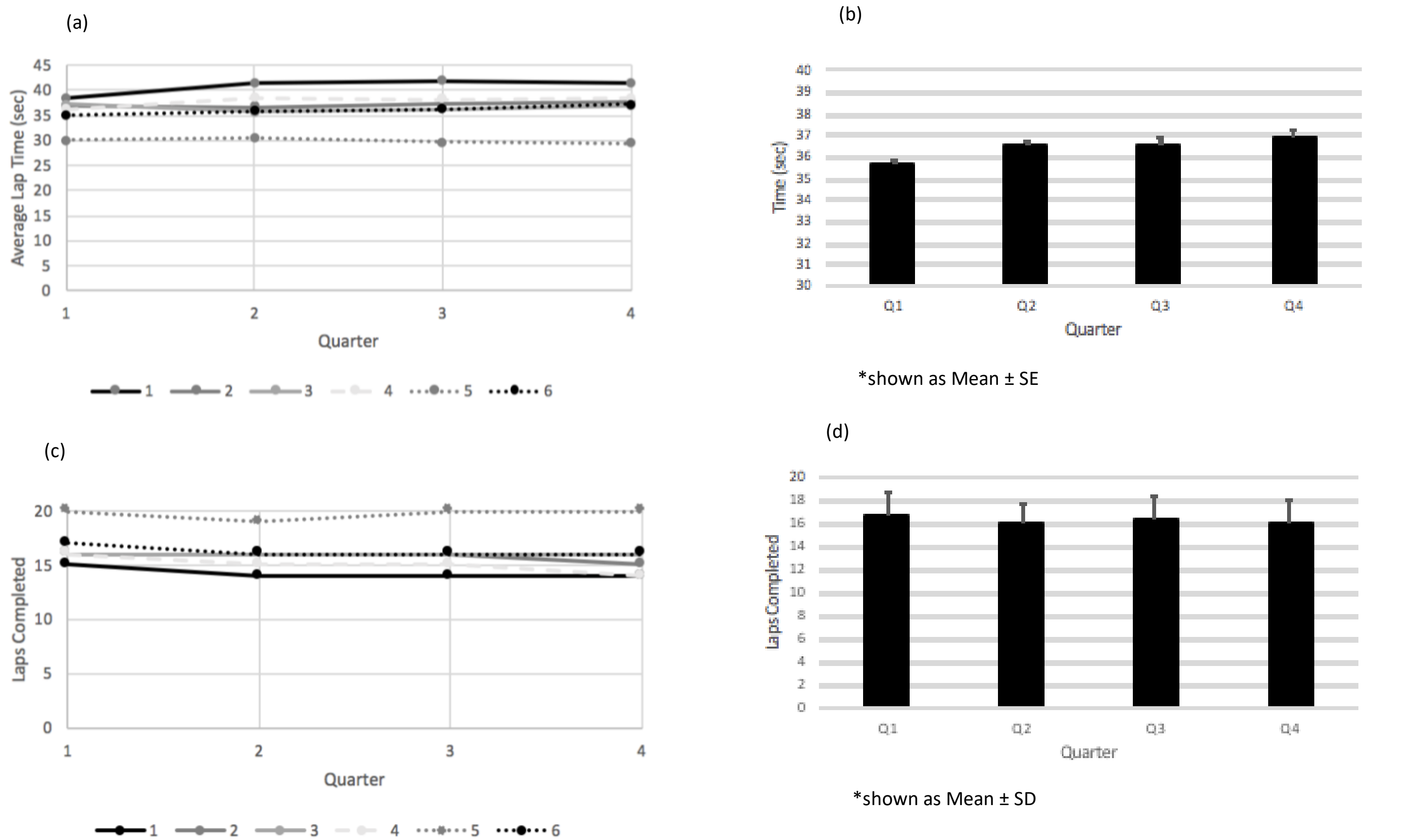
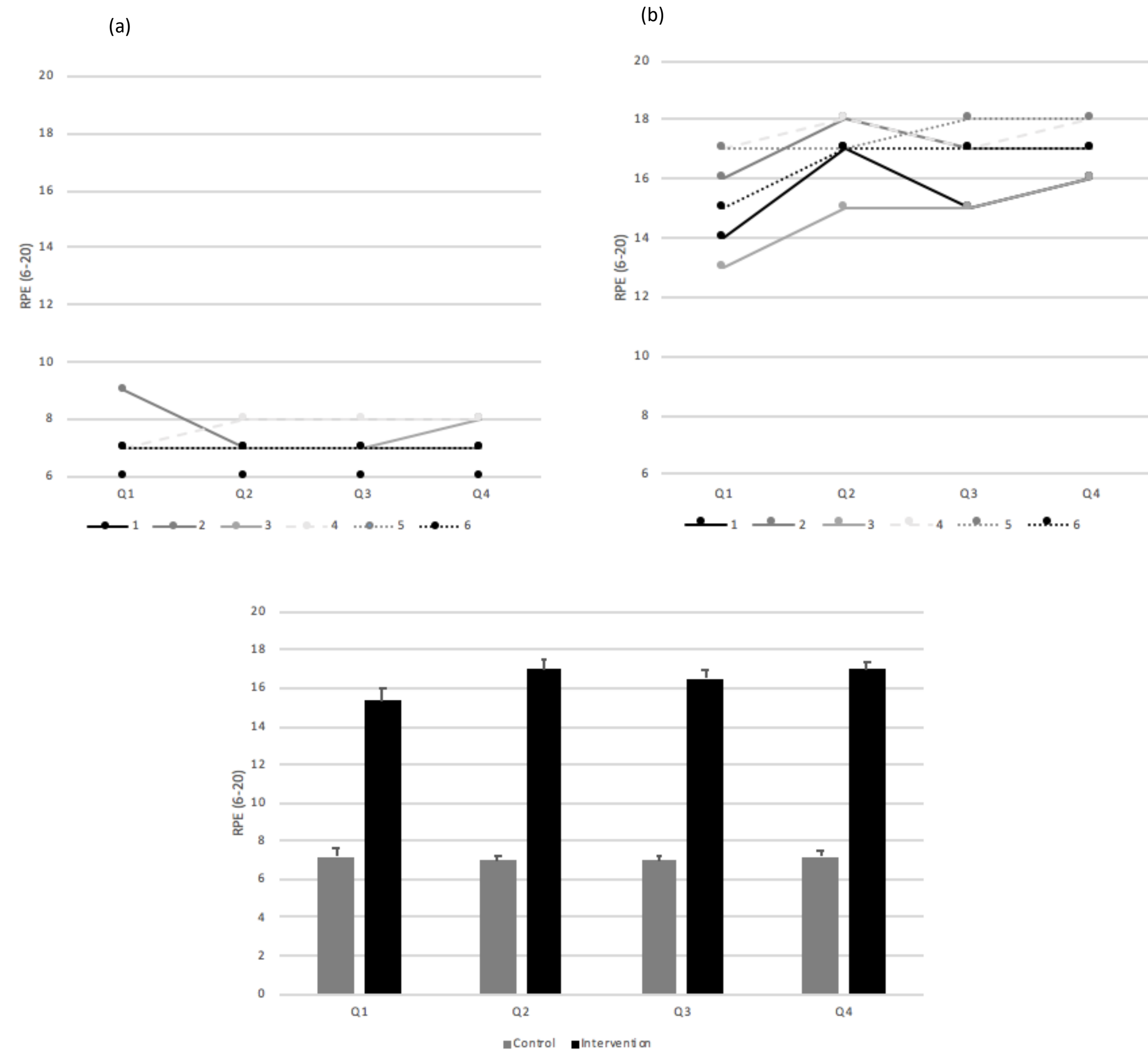
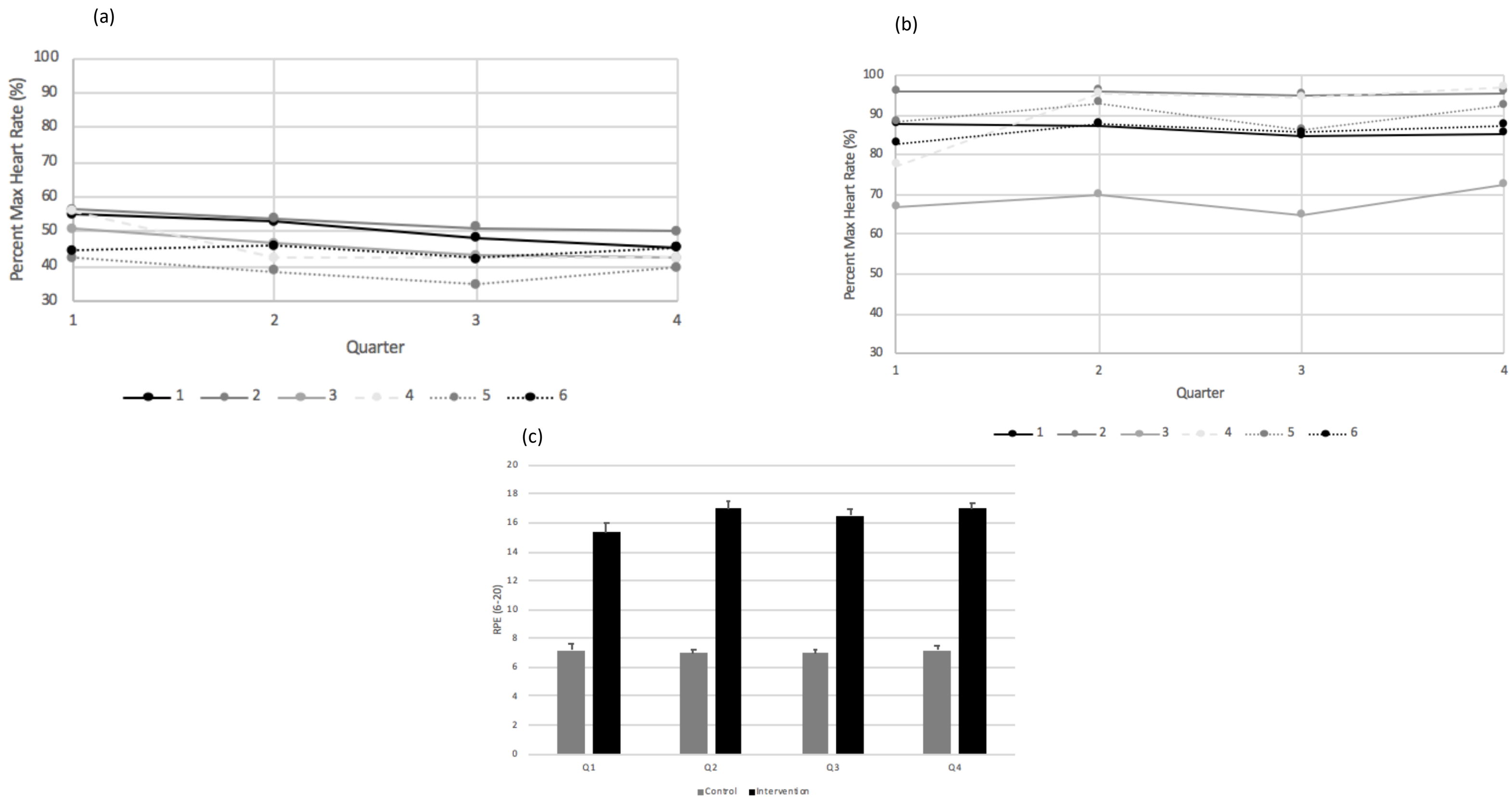


Figure 3.1 Average Lap Time and Number of Laps Completed by Quarter. Figure 3.1a shows the average lap time for each quarter of the BEST protocol for each participant. Figure 3.1b shows the average lap time for each quarter of the BEST protocol for all participants. Average lap times were not different across the 4 quarters ($p>0.05$, $\eta^2 = 0.321$). Figure 3.1c shows the number of laps completed by each participant by quarter. Figure 3.1d shows the average number of laps completed by all participants by quarter. The mean number of laps completed were significantly different across the 4 quarters of the BEST protocol ($p=0.02$, $\eta^2 = 0.467$).



*shown as Mean ± SE

Figure 3.2 Rate of Perceived Exertion (RPE) by Quarter. RPE is a subjective measure to indicate how strenuous the participants felt the quarter based on their perception of their exhaustion levels on a scale from 6 to 20, where 6 indicates resting with minimal effort and 20 indicates absolute maximal effort. Figure 3.2a shows the RPE for each quarter of the control session for each participant. Figure 3.2b shows the RPE for each quarter of the intervention session for each participant. Figure 3.2c shows the control and intervention averages of the RPE by quarter for all participants. There was a significant interaction between the Intervention and Time for RPE where RPE was greater in the game simulation than during the control session, however RPE decreased in the control condition while RPE increased during the exercise condition ($p=0.004$, $\eta 2 = 0.0057$).



*shown as Mean ± SD

Figure 3.3 Percent Max Heart Rate per Quarter. Figure 3.3a shows the control session average heart rate per quarter as percent of the age-predicted maximal heart rate for each participant. Figure 3.3b shows the exercise session average heart rate per quarter as percent of the age-predicted maximal heart rate for each participant. Figure 3.3c shows the control and intervention session average percent of max heart rate for all participants by quarter. There was a significant interaction between the Intervention and Time for percent of maximal heart rate where percent of maximal heart rate increased in the game simulation and decreased during control ($p=0.011$, $\eta 2 = 0.0113$).

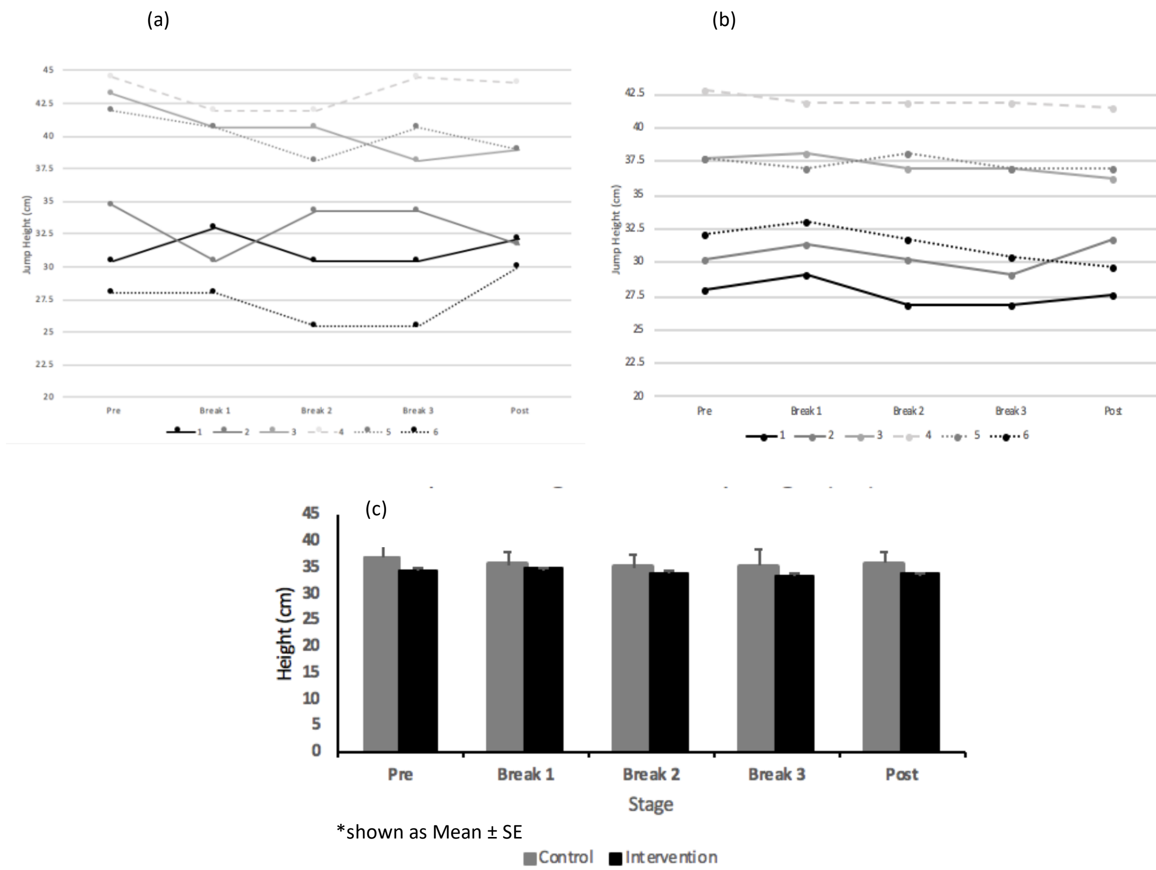


Figure 3.4 Vertical Jump Height. Figure 3.4a shows the control vertical jump height for each participant at each testing period. Figure 3.4b shows the intervention vertical jump height for each participant at each testing period. Figure 3.4c shows the average vertical jump height for all of the participants at each testing period for the control and intervention sessions. The test was used as an index of fatigue. Depicted is the individual jump height averages during the different testing periods. No significance for the intervention on vertical jump height ($p>0.05$, $\eta 2< 0.1$) occurred, but there was a significant effect of time on vertical jump height ($p=0.0007$, $\eta 2 = 0.019$).

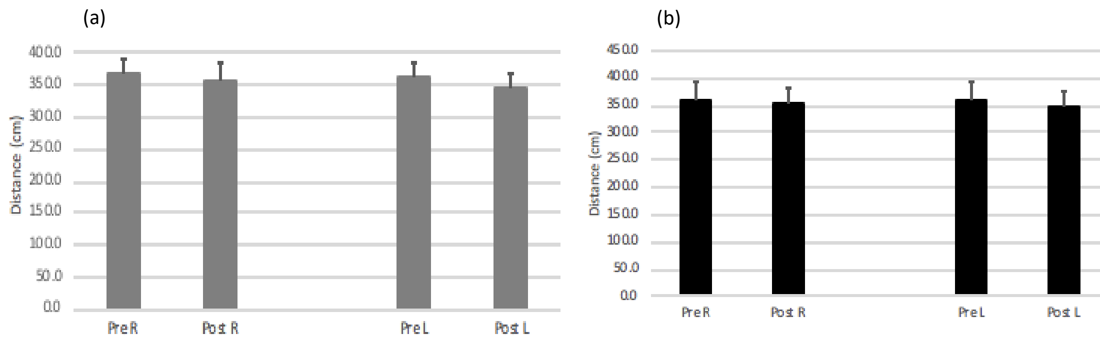


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Figure 3.5 Average Single Leg Triple Jump Distance. Figure 3.5a shows the control single leg triple jump average distance for all participants on the right and left leg from the Pre- and Post-testing. Figure 3.5b shows the intervention single leg triple jump average distance for all participants on the right and left leg from the Pre- and Post-testing. The single leg triple jump gives insight into bilateral discrepancies as well as ACL risk assessment based on reduction in performance and is also considered a fatigue assessment. There was no significant effect for the intervention on this measure ($p>0.05$, $\eta2 < 0.1$), but there was a significant effect of time for the left leg single leg triple jump distance ($p=0.01$, $\eta2 = 0.28$).

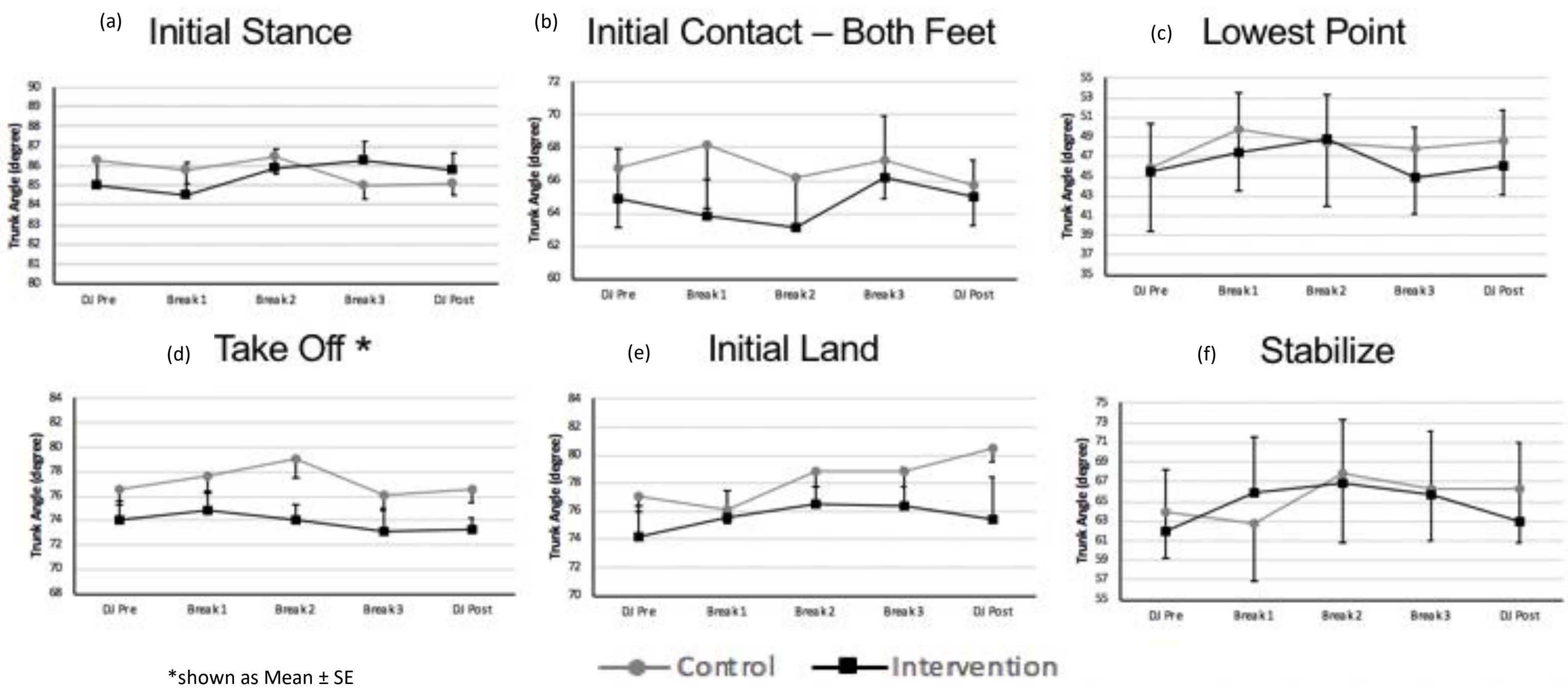

Figure 3.6 Drop Jump Relative Trunk Angle. Figure 3.6a shows the average trunk angle during the initial drop jump stance for the control and intervention testing points. Figure 3.6b shows the average trunk angle during the initial contact with the ground of both feet for the control and intervention testing points. Figure 3.6c shows the average trunk angle during the lowest point for the control and intervention testing points. Figure 3.6d shows the average trunk angle during take-off for the control and intervention testing points. This was found to be significant, indicated with the *, with a p value of 0.0073 and a $\eta2$ of 0.36. Figure 3.6e shows the average trunk angle during initial landing of the drop jump for the control and intervention testing points. Figure 3.6f shows the average trunk angle during the stabilizing point for the control and intervention testing points. With the exception of the take-off phase, there was no significant changes between the intervention and control for relative trunk angle.

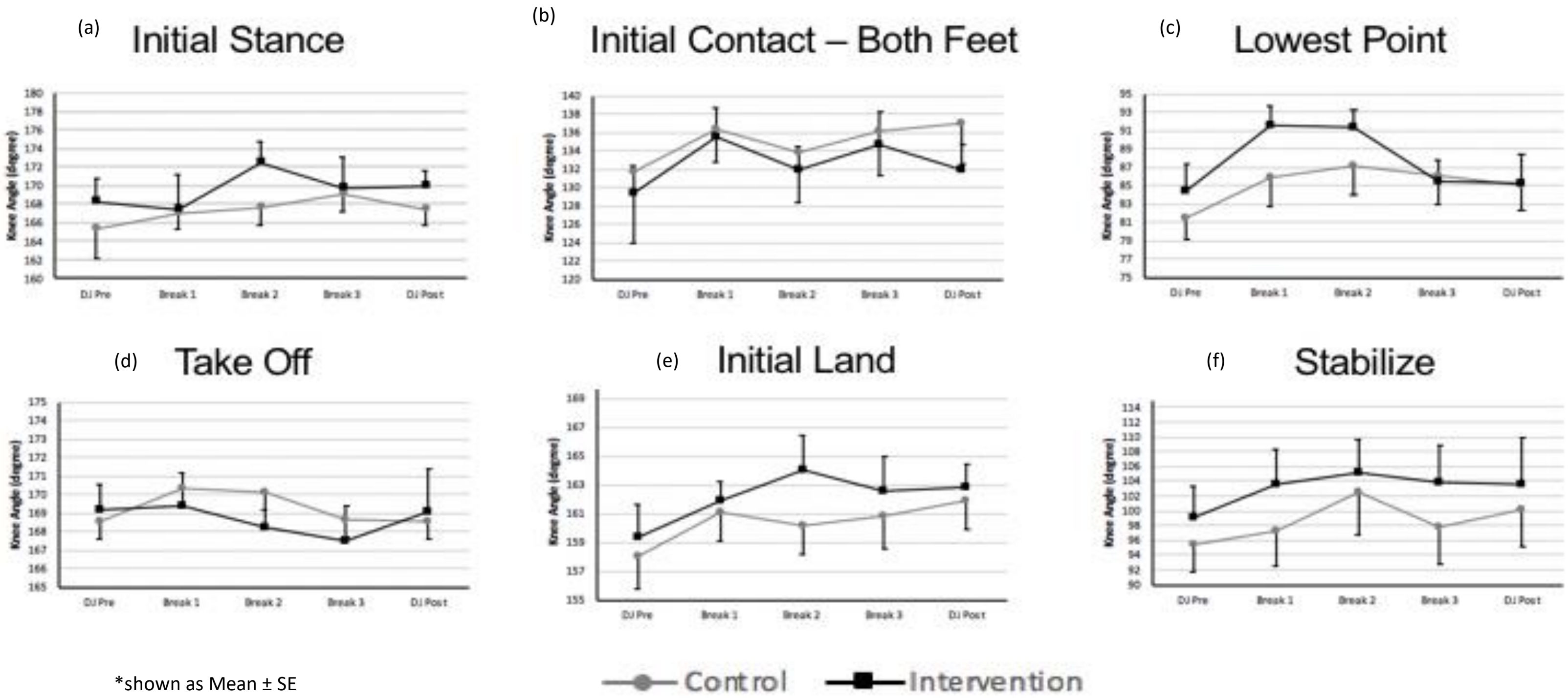

Figure 3.7 Drop Jump Knee Angle. Figure 3.7a shows the average knee angle during the initial drop jump stance for the control and intervention testing points. Figure 3.7b shows the average knee angle during the initial contact with the ground of both feet for the control and intervention testing points. Figure 3.7c shows the average knee angle during the lowest point for the control and intervention testing points. Figure 3.7d shows the average knee angle during take-off for the control and intervention testing points. Figure 3.7e shows the average knee angle during initial landing of the drop jump for the control and intervention testing points. Figure 3.7f shows the average knee angle during the stabilizing point for the control and intervention testing points. There were no significant changes between the intervention and control for the knee angle ($p>0.05$, $\eta2 < 0.1$).

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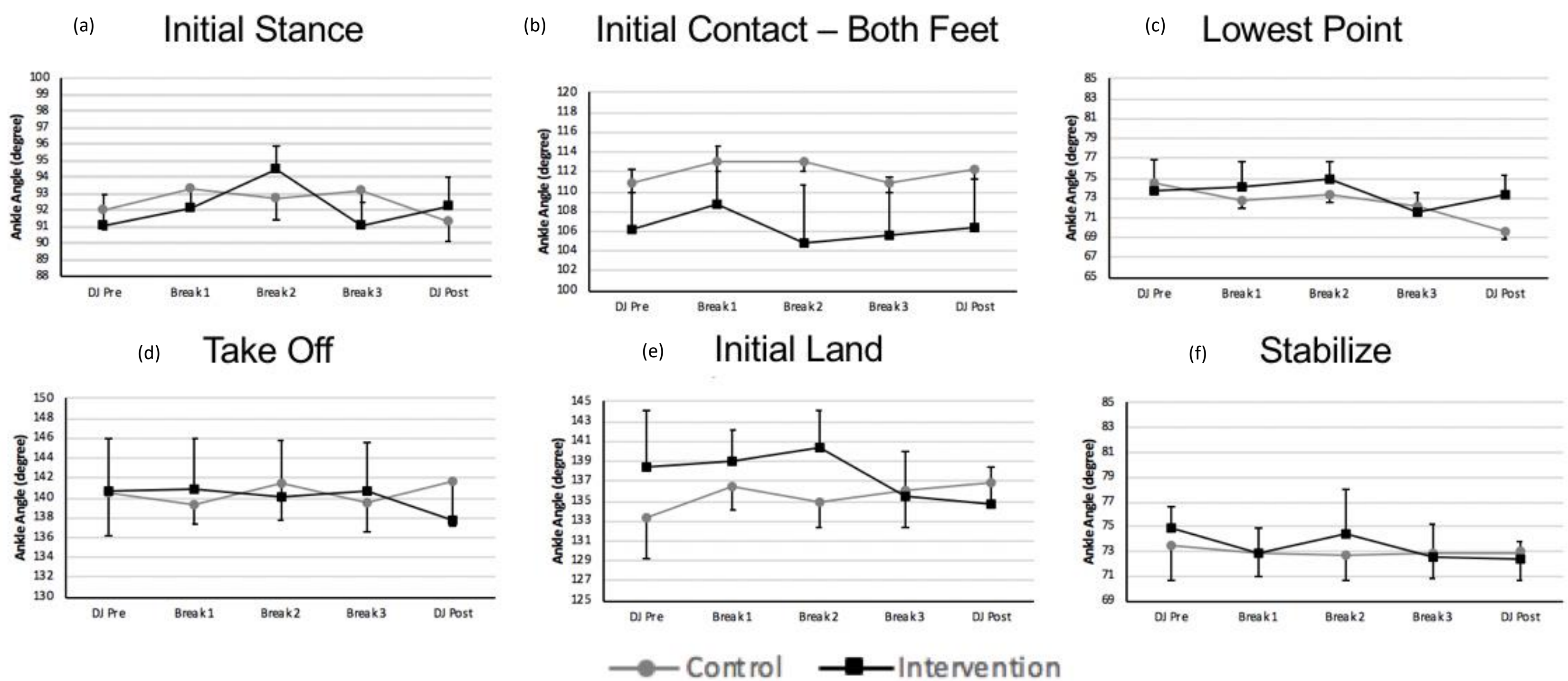

*shown as Mean ± SE

Figure 3.8 Drop Jump Ankle Angle. Figure 3.8a shows the average ankle angle during the initial drop jump stance for the control and intervention testing points. Figure 3.8b shows the average ankle angle during the initial contact with the ground of both feet for the control and intervention testing points. Figure 3.8c shows the average ankle angle during the lowest point for the control and intervention testing points. Figure 3.8d shows the average ankle angle during take-off for the control and intervention testing points. Figure 3.8e shows the average ankle angle during initial landing of the drop jump for the control and intervention testing points. Figure 3.8f shows the average ankle angle during the stabilizing point for the control and intervention testing points. There were no significant changes between the intervention and control for the ankle angle ($p>0.05$, $\eta 2 < 0.1$).

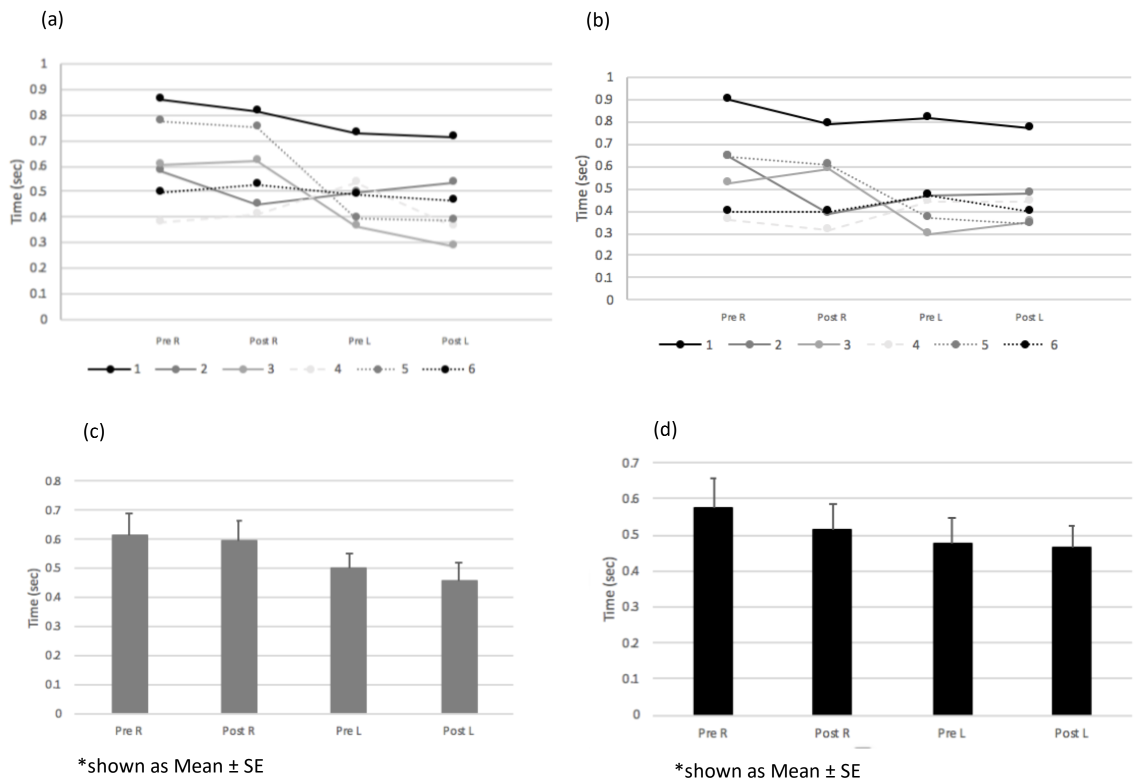

Figure 3.9 Drop Cut Time to First Step. Figure 3.9a shows time to first step for the drop cut for each participant towards the right and left direction from the Pre- and Post-testing of the control session. Figure 3.9b shows time to first step for the drop cut for each participant towards the right and left direction from the Pre- and Post-testing of the intervention session. Figure 3.9c shows average time to first step for the drop cut for all participants towards the right and left direction from the Pre- and Post-testing of the control session. Figure 3.9d shows average time to first step for the drop cut for all participants towards the right and left direction from the Pre- and Post-testing of the intervention session. The initial contact of a step following the landing of both feet on the ground was considered the time it took to reach the first step. This measurement is an indication of reaction time. There was a significant effect of the intervention for the step towards the right direction ($p=0.04$, $\eta 2= 0.25$) but no significant effect of intervention towards the left direction ($p>0.05$, $\eta 2 < 0.1$).

CHAPTER 4

DISCUSSION

The purpose of this study was to determine the effects of basketball game related stress on lower extremity landing mechanics for active females. Contrary to the hypothesis, the BEST protocol did not elicit signs of fatigue in these physically active female participants even though the participants showed signs of strenuous physical exertion during the BEST protocol. Furthermore, the BEST protocol did not influence kinematics when compared to baseline measures. Participant self-management of her exertion, in order to complete the task and perform at a high level for testing, resulted in insufficient levels of fatigue for the selected testing measures. The demands of the BEST were insufficient to impose fatigue.

Protocol Measures

The measure of lap time and number of laps completed shows the ability of the participants to complete the BEST protocol to its true intent. The BEST was designed for the participants to complete four ten-minute rounds of 20 laps, for an average of 30 seconds per lap. At the completion of the 20 laps or 10-minute period, whichever occurred first, the quarter was terminated. One participant was able to complete the 20 laps during the 10 min period for three of the four quarters. None of the other five participants were able to complete this task. One possible explanation for the difficulty of the participants to meet the 30 second time to complete a lap, is that, the BEST was

designed and validated based on male basketball athletes.[111] While it is not currently established, there could be a different expectation towards the total laps completed and the time to complete each lap for females based off of the load of a female basketball match. However, since one of the six participants was able to complete the 20 laps in the allotted time, the gender discrepancy does not make it impossible for females to complete this protocol.

There was a general positive trend for the average lap time as the participants completed additional quarters and a slight negative trend for the average laps completed per quarter. While the average lap time per quarter did not result in significant change, there was a significant reduction in the number of laps completed per quarter (Fig. 3.1). The decline in laps completed from the first quarter to the second quarter was responsible for this significant finding, there was no significant difference in laps completed for the third and fourth quarter from the second quarter. Given the definition of fatigue as an exercise induced reduction in power outputs reversible with rest, the change in laps completed does not seem to represent fatigue and is more likely physical exertion. In part, laps completed would not be a power output but rather a representation of the exercise used to induce the fatigue. Also, if this measure were to suggest fatigue, the laps completed in the third and fourth quarter would have been less than the second quarter.

The measures of lap time and laps completed are related. As the lap times increase, the participants will not be able to complete as many laps in the 10-minute period. Possible explanations for the reduction in lap time include physical exertion and

changes in motivation. While there were no measurable amounts of fatigue with the BEST, it is possible that levels of physical exertion as the rounds progressed resulted in this change. In addition, this protocol took place over an hour and twenty-minute time period. Once the duration of the session progressed there could have been a reduction in motivation to maximally exert oneself or in order to achieve better final measures, the participants - while knowing that there would be Post-testing measures - would have exerted reduced levels of effort, i.e. self-paced, in order to reserve energy.

The significant difference in RPE score and percent of maximal heart rate for each quarter when comparing the control to the intervention suggests the physical exertion for the participants. As shown in Fig. 3.2 and Fig. 3.3, a dramatic difference occurred in these measures when comparing the control to the intervention session. Identifying the success of the intervention in creating physical exertion is imperative to show that the BEST was capable of creating a subjective and an objective physiological response in the participants.

Fatigue has been suggested as a risk factor for ACL rupture in female athletes. Previous studies have linked fatigue with negative alterations in landing kinematics, muscular coordination, and proprioception. Different measures of fatigue have been utilized including changes in EMG values relative to MVIC, power outputs, and vertical jump height. In this study, vertical jump height was utilized given its applicability to basketball athletes. However, there was not a significant change in vertical jump ($p>0.05$, $\eta2< 0.1$), but there was a significant effect of time on vertical jump height ($p=0.0007$, $\eta2 = 0.019$). This suggests that the duration of the testing sessions was

primarily responsible for the variation in vertical jump height and the addition of the intervention did not result in a significant change in vertical jump height. The BEST protocol is predominantly closed chain exercises with linear movement patterns anteriorly or laterally, so a vertical jump test might not be an accurate depiction of the fatigue based on the exercises performed.

A secondary measure of fatigue used for this study was the single leg triple jump. ACL related injuries are most common during single leg tasks and the use of the single leg triple jump as a return to play measure for knee instability and ACL risk made it an ideal choice for this study design. The study population was generally active college-aged females. With the exception of one participant, the participants were novice to the single leg triple jump task. They were asked to keep their hands on their hips during this jump in order to reduce the use of upper body movements for momentum and stabilization so the lower extremity effects would be more pronounced. The acceptance of a single leg triple jump distance required a stuck landing on the third jump that was held for 10 seconds, in order to determine a controlled maximal jump distance. Three failed trial attempts on one leg resulted in a missed trial recording. The average distance for the pre- and post- testing was analyzed. The novelty of the task led to higher variation in jump distances between trials. No significance was found in the change in jump distance following the intervention when compared to the control ($p>0.05$, $\eta 2 < 0.1$). There was significance in the change in left leg triple jump distance based on time ($p=0.01$, $\eta 2 = 0.28$). This result suggests that the physical exertion the participants' reached by the BEST protocol was not enough to elicit significant levels of fatigue and

that the longer duration of the control and intervention session is the primary explanation for the reduction in jump height.

Potential self-modifications or imbalances would affect the validity of this measure for the study population. All participants were right hand dominant, so the assumption of right foot dominance would suggest the possibility of left footed reduction in coordination and stability during the single leg triple jump task. Being a non-dominant side, the left side has the potential to be less stable and have an increased variation in jump distance. Also, in order to avoid a "failed" attempt by not sticking a landing, the participants were more likely to complete a more conservative jump following a missed attempt. In addition, since this measure exclusively focused on distance, there could have been changes in muscular coordination or compensations to produce the observed effects. For example, a 2006 study that tested single leg hop landings in 13 healthy men found that following fatigue there was a shift towards ankle compensations in order to compensate for knee weakness.[95]

Participant Descriptive Analysis

There were some individual differences in the trends noticed for lap time, vertical jump, heart rate, and the time to reach the first step for the drop cuts. The first participant (1) had slower lap times for her later laps as the number of laps and quarters increased. This coincided with a decrease in average heart rate for the later quarters when compared to the first quarter. There is an error of about 12 beats per minute for the age predicted max heart rate. So, this or any other participant/s percent of max heart rate calculation may be in error. However, the error for each subject would be

consistent and not effect relative intensity between quarters. Her testing measures resulted in a minor decline in vertical jump height and a faster time to reach her first step for the drop cut. She seemed to self-pace the final rounds in order to reduce intensity allowing for less decline in vertical height measures and a reduction in time to first step.

The second participant (2) was the only participant to see an increase in vertical jump height following the BEST protocol. She had a slight increase in average lap time for later quarters than her initial quarter's lap time and had an overall slight reduction in heart rate from the first quarter to the last quarter. However, her relative intensity was the highest of the participants. Her lowest quarter average heart rate was still 95.6% of her age predicted heart rate max. She had a drastic reduction in time to first step towards the right direction, which could be attributed to a change in technique. Given her testing measures only improved with increased exercise, there could be genetic or other factors that impact her heart rate measures, so it was less representative of her relative intensity for the task. In addition, her Pre-testing measures might have been impacted by an insufficient warm-up or submaximal effort.

The third participant (3) had the lowest average heart rates per quarter for the experimental session. While the other five participants had average max heart rates of 80-85%, her highest quarter average heart rate was only 72.3% of her age predicted max heart rate. She consistently completed 16 laps with only a slight decline in pace and had a slight decline in vertical jump height from the initial measures to final measures. Her time to first step increased, suggesting her physical exertion impacted reaction

timing. Her testing results reflect the general trends of the group, so It is most likely that the heart rate measure was inaccurate during this trial or that the individual has an abnormal exercise heart rate.

The fourth participant (4) followed general trends of what was expected in her measures. As she completed more quarters, she progressively completed less laps (i.e. had an increase in lap time). She had an increase in average heart rate during the course of the quarter as well as with increasing quarter number). She had a slight decline in vertical jump height and only minor changes in response time for the drop cut.

The fifth participant (5) was the only one to complete 20 laps in the 10- minute round, which she achieved in three of the four quarters. Her lap time skewed the group average for lap time to be shorter as well as increased the average laps completed. She also had an increase in heart rate as the quarters progressed, a minor reduction in vertical jump height, and a slight improvement in drop cut time to first step time.

The sixth participant (6) followed the group's general trends. She had an increase in average lap time as well as average heart rate with the final quarter when compared to the first quarter. She had a reduction in vertical jump of one inch and only minor improvements in drop cut reaction time.

Having only six participants is a limitation of this study since an abnormality of one individual has a greater impact on the group's averages. The most impacted was lap time since participant 4 was quicker than the rest of the group and heart rate where one participant was slightly higher, and another was significantly lower than the rest of the participants. Given the small sample size and minor differences in individual responses,

there were no obvious trends for the measures of heart rate, average lap time, vertical jump, and time to the first step of the drop cut.

Testing Measures

The drop jump task was utilized to address changes in jumping and landing mechanics with emphasis on the relative trunk angle, knee angle, and ankle angle. The drop jump task was broken down into six phases: initial stance, initial contact with both feet on the ground, the lowest point of the squat down, the take-off point, the initial landing from the jump, and the point of stabilization or the second lowest point reached.

Dartfish 2D videography software was used to determine the trunk, knee, and ankle angles at each of the six drop jump phases using sagittal recordings. The trunk angle at the take-off phase of the drop jump was the only significant effect of the intervention found from the kinematic data (Fig. 3.6). Following the physical exertion of the BEST, the participants had a tendency to elevate their trunk to a lesser degree than pre-testing measures. Possible explanations include a change in motivation from performing a maximal jumping to performing a submaximal jump or a reduction in postural control following physical exertion where it was more challenging to keep an upright chest position during take-off following the BEST.

While no significant changes in the other phases of the relative trunk angle, knee angle, or ankle angle were found, it is important to note that the angles measured are representative of the task. The expected angles for the trunk, knee, and ankle for the different demands of the drop jump phases were as anticipated. For example, while in

the initial stance on top of the box, the legs are outstretched and the knee angle around 170 degrees and an ankle angle around 90 degrees is what would be expected.

The drop jump is a commonly used task in fatigue protocols since there is force absorption following the step off the box and a jump task. In this study, measures had limited results, but other studies have found, especially in females, drop jump landing following a closed chain exercise protocol had increased flexion velocity, knee joint abduction angles, and delayed lateral hamstring activation.[38] The limitation of my study to focus on the sagittal view for joint angles means it would not have been able to detect knee abduction angles effectively. In addition, most noted changes in females were either joint velocity based or related to changes in the muscular coordination in the completion of the task.[97] Using a 2D system, a researcher would not be able to recognize those more complex changes, logically, there could have been other changes that occurred that were not represented by my measures.

When compared to a single-leg landing task, a double leg jump is more stable. Regarding the level of exertion as a result of the BEST, it is possible that the drop jump task was too controlled in order to see major changes. For example, one study tested 25 NCAA female athletes' unanticipated and anticipated single-leg landings following a squat and jump sequence-based protocol.[17] Fatigue resulted in significant increases in initial contact hip extension and internal rotations, and in peak amounts of knee abduction, internal rotation, and ankle supination angles. These findings were more pronounced in the unanticipated landings, which suggest that fatigue and decision making when combined inhibit performance and could result in an overload in central

control mechanisms, leading to injury. Since the drop jump task was a more controlled movement that was planned, the testing measure chosen could have been too simple to detect the extant of changes that occurred as a result of the BEST.

No significant effect for timing or intervention was observed for the time it took to reach the first step of the drop cut towards the left direction ($p>0.05$, $\eta 2 < 0.1$). However, there was a significant effect of the intervention on the time it took to reach the first step towards the right direction ($p=0.04$, $\eta 2=.25$). Cutting is a common task completed in basketball games and often results in a change of direction with deceleration or pivoting with the knee near full extension while the foot is planted, which are mechanisms of ACL injury. This measure was chosen to determine shifts in reaction time upon landing to reaching the cone as well as simulate the motion involved in the sport of interest, basketball.

A delay in timing for the first step would have suggested a reduction in processing following the BEST, suggesting signs of central fatigue. However, the significant findings towards the right direction found a reduction in time it took to complete this action. All participants were unfamiliar with this task, which would allow for a motor learning curve which could explain the improvement of the measures following the exercise protocol. The participants, with increased time, would be able to determine a more effective way to complete the task, thus making the final testing faster than the initial testing. In addition, this was the final measure taken during the testing sessions. It is probable that participants had increased motivation to complete the task more quickly in the Post-testing period given that was the final task they had to

complete. This would have been more pronounced during the intervention session, since the intensity of the rounds was greater so there was more incentive to want to be done with the session. Finally, the greater level of physical activity in the exercise protocol compared to the control might have allowed for the musculature to be more responsive due to greater stimulation. Since the motor neurons have been communicating to the muscles for a greater variety and intensity of tasks during the intervention session, the ability to communicate in order to complete the cut task would be improved.

Another factors to consider as to the changes performing the drop cut task would be the shift in muscular coordination in order to complete the task as well as the changes in ground reaction forces. One study that focused on the changes in cutting following a handball match found a selective decrease in hamstring neuromuscular activity during a side cutting task.[144] This seemed to suggest that an impairment in ACL agonist muscle (the hamstrings) during cutting occur in response to the fatigue. In part, our limited results could have been due to the scope of our measure of the time it took to reach the first step or due to the absence of fatigue and factors related to changes.

Limitations

The predominant limitation of this study stems from low statistical power as a result of a small sample size of only six participants. This small number, in turn, has resulted in a low effect size, $\eta 2 < 0.1$, for several measures. The generalizability of this study should be limited to highly motivated college-aged active females either seeking or at a higher education level, and the results of this study cannot be extrapolated to a

basketball specific population, male population, or specifically low socioeconomic or educated population. There was some novelty to the movements which could have affected the results; however, this should have been limited due to the random assignment of the session order. Another consideration would be motivation as the intervention session continued, as addressed above in terms of lap time and total laps completed while knowing there would be additional testing as well as the limitations of shoe-surface interaction on a relatively slick basketball floor. The floor interaction could have limited the ability of the participants to more aggressively complete the course or testing measurements, i.e. drop cut- and to a lesser extent- single leg triple jump.

Conclusion

The purpose of this study was to determine the effects of basketball game related stress on landing mechanics in active females. It is evident that the BEST intervention was successful at creating physical exertion for participants when compared to the control session based on both the subjective RPE scoring and based on the physiological marker of heart rate, which were found to be significant. While there was physical exertion, no clear evidence of fatigue resulted from the measures of vertical jump and single-leg triple hop. As far as kinematic changes, the only significant finding was the relative trunk angle during the take-off phase of the drop jump. This change in trunk angle for take-off could be linked to decreased motivation for a chest up form or by a reduction in postural stability following the exertion. Thus, I conclude that one completion of the BEST did not elicit fatigue or substantial changes in joint angles.

Future Directions

While the number of significant findings is limited, this study was able to provide insight as to other factors to consider with basketball game related stress and ACL injury risk. First, given the limitation of such a small sample size, a larger sample size will be more suggestive of significance and trends. Having six participants causes the results of one participant to more heavily impact the overall results. In terms of measurements, while the 2D videography angles from Dartfish offered insight as to the joint angles using a 3D measurement system will allow for more subjectivity in the results and offer a more valid measure. In addition, having an increased focus not only on the angles but also on the velocity of the angles will be more representative of the body's ability to respond quickly to result in the same joint angles. This focus would suggest a decline in the central nervous system to communicate effectively with the muscles. Given that ACL injuries do occur at such a high rate for this population, investigating how accumulation of basketball game related stress, i.e. playing multiple games, can impact levels of fatigue and motor coordination is vital to address this issue. Finally, there should be more investigation of both upper and lower extremity effects on accuracy tasks following basketball game related stress. For example, is the participant capable of controlling the exact location of her feet during different tasks? As time of activity increases, is there a trend toward increased variation in the ability to reach a touch point? Does that amount of variation or reduction in motor control increase the likelihood of game type situations where an ACL injury would occur?

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