THE EFFECTS OF AN OVERHEAD GOAL AND MOVEMENT ANTICIPATION ON LOWER EXTREMITY BIOMECHANICS AND MUSCLE ACTIVATION PATTERNS

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Abstract

Noncontact anterior cruciate ligament (ACL) injury is becoming more common among sport-related injuries. Several “high-risk” movements such as landings and cutting have been associated with ACL injury along with several biomechanical risk factors. These risk factors, specifically at the hip, knee, and neuromuscular abnormalities (unbalanced quadriceps-hamstrings activation ratios) have also been identified as possible causes that lead to ACL injury. Females are nearly 2.5 times more likely to experience an ACL injury while playing volleyball when compared to their male counterparts. While research continues to investigate the reasoning for this injury, the results may not be accurate to what athlete’s experience while playing in a real competition. The purpose of this study was to determine the effects of a physical overhead goal and a subsequent unanticipated lateral cut on the kinematics, kinetics, and muscle activation of the hip and knee of the dominant leg during the landing of a stop vertical jump task (volleyball block). Participants completed four testing conditions – with a target and an anticipated cut following the landing (WTA), with a target and an unanticipated secondary cut (WTUA), and then both anticipated and unanticipated secondary movements with no overhead target (NTA, NTUA). Results indicated that when a target was added to the testing environment, there was a significant decrease in the initial contact (IC) knee flexion angle and an increase in the peak knee extension moment. When comparing anticipated and unanticipated tasks, there were decreases in the IC knee abduction angles and peak knee extension moments in the unanticipated trials compared to the anticipated trials. Significant interactions were also seen in the peak knee abduction moments and the IC hip flexion angles specifically when the target is present. These findings suggest that the more realistic testing environment altered hip and knee landing mechanics, which in turn can be used to help better evaluate the risk factors for ACL injury.

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Nomenclature

A	Anticipated condition
ACL	Anterior Cruciate Ligament
BF	Biceps Femoris
CCI	Co-contraction Index
EMG	Electromyography
GRF	Ground Reaction Force
Hz	Hertz
kg	Kilogram
LEFS	Lower Extremity Functional Scale
MH	Medial Hamstrings (semitendinosus)
m	Meter
m/s	Meters per Second
ms	Millisecond
MVIC	Maximum Voluntary Isometric Contraction
N	Newton
Nm	Newton-meter
NTA	No Target Anticipated condition
NTUA	No Target Unanticipated condition
QD	Quadriceps Dominance
VL	Vastus Lateralis
VM	Vastus Medialis
WTA	With-Target Anticipated condition

WTUA	With-Target Unanticipated condition
Yrs	Years

CHAPTER 1: INTRODUCTION

Background and Rationale

The anterior cruciate ligament (ACL) is one of the most important stabilizing ligaments of the leg (1). An estimated 100,000 to 250,000 ACL injuries occur every year (1, 2), with injury rates becoming more prominent in multidirectional sports and younger populations (3). ACL injuries occur when the load applied to the ACL exceeds the strength of the ligament itself (4, 5). This load can become too great both in sporting activities or during daily tasks (6). In multidirectional sports, both contact and noncontact ACL injuries can occur during game play, with approximately 70% of these injuries occurring in noncontact situations (7, 8). Unfortunately, studies have shown that athletes who suffer an ACL injury have about a 40% chance of developing knee osteoarthritis later in life (9).

Certain dynamic movements in sports have been found to increase the risk of noncontact ACL injury. Noncontact ACL injuries occur during activities that include rapid deceleration, such as landing (8), or sudden changes in direction such as pivoting and cutting (10). Jump landings are one of the most dangerous movement patterns for athletes (11), with injuries most likely occurring almost immediately after initial ground contact (8). Jump landings are considered one of the most dangerous due to the knee being near full extension at initial contact with the ground, increasing the force applied to the ACL due to increased anterior tibial shear force not being absorbed by the muscles around the knee. Additionally, most ACL injuries after landing occur during unilateral foot contact (7, 12, 13). During unilateral landings, there is increased knee abduction during the stabilization phase of the landing which could be due to decreased hip and core strength or lateral tilting of the pelvis to the contralateral side (11). Finally, lateral cutting has also been identified as a more high-risk task due to the increased knee abduction angles when completing the movement (14). While this injury can be very detrimental, several risk factors, both modifiable and non-modifiable, for the injury have been identified.

Kinematic risk factors involved with noncontact ACL injuries have been identified at both the knee and the hip. At the knee, the main modifiable risk factor in the sagittal plane is a flexion angle of less than 30° and, the center of gravity behind the knee (11, 15). During landing, peak loads on the ACL occur just after initial contact when the knee is near full extension (between 0° and 30°), (16). Landing with decreased knee flexion angles has been shown to increase the other main modifiable risk factor at the knee which is high levels of abduction angles (11, 17). Other risk factors that can contribute to increased knee abduction angles and internal knee adduction moments are external tibial rotation and internal rotation at the hip (15). At the hip, another common kinematic variable that has been found in noncontact ACL injuries was decreased hip flexion during landing (7, 11, 18). Decreased hip and knee flexion cause the quadriceps to produce more anterior tibial shear force that is directed to the ACL due to the limited range of motion (ROM) of the hip and knee.

Modifiable risk factors have also been identified in the lower extremity muscle activity. Specifically, altered muscle activation patterns and muscular imbalances have been associated with increased ACL injury risk. Electromyography (EMG) of both the quadriceps and the hamstrings have been analyzed during dynamic tasks to identify if any imbalances in muscle force production are present. Quadriceps dominance (QD), a modifiable neuromuscular imbalance, is defined as the imbalance between knee extensor and flexor strength, recruitment, and coordination (19). The high activation of the quadriceps during an athletic task such as a landing or cutting task can increase anterior tibial shear force, which has been shown to load the

ACL through the patellar tendon (5, 20). The activation of the hamstrings can provide a counterbalancing force to protect against the relatively higher quadriceps force, thus reducing the anterior tibial translation and potentially decreasing ACL strain (21). The quadriceps-to-hamstrings co-contraction index (Q:H CCI) is one way to measure the difference in the activations of both muscle groups. When the Q:H CCI is greater than one, the quadriceps are activating at a greater level than the hamstrings which would increase the amount of anterior tibial shear force that is applied to the ACL. The Q:H CCI has been found to influence the knee flexion angle at initial contact, with higher ratios resulting in decreased initial contact knee flexion angles (22).

Of the modifiable risk factors, common differences have been identified between males and females. Females are nearly 2.5 times more likely to experience an ACL injury in volleyball when compared to males (23). Intrinsic, or nonmodifiable risk factors seen more frequently in females include: high Q-angle, increased joint laxity, decreased ACL size, decreased femoral notch sizes, and increased posterior tibial and meniscal slopes (24-26). Extrinsic, or modifiable risk factors that females exhibit more frequently compared to males include: increased knee abduction angles at initial contact, decreased peak knee flexion angles during dynamic movements, increased quadriceps activity during dynamic movements, and decreased hamstrings activity during dynamic movements (19-21, 27-29).

A primary concern in biomechanics research is the lack of a realistic, game-like environment during data collection. One step being taken to make laboratory testing more realistic is by including external objects in the testing environment. It is possible that including a physical goal (i.e. object) in the testing environment and having the subjects reach for that goal can potentially lead to altered lower extremity joint kinetics and kinematics (30). Mok et al. (31)

found that including on overhead target to a drop vertical jump task increased jump height by 5.8%. The increase in jump height also led to increases in the moments due to a 5% increase in the peak vertical GRF during the landing. Another step in making the testing environment more realistic is by including cognitive tasks to dynamic movements. This technique attempts to create a more game-like environment by having the athlete focus on multiple tasks, rather than just primarily focusing on the experimental task. Finally, including the unpredictability of an unanticipated task is another way to make the testing environment more realistic. This unpredictability requires the athletes to react to a stimulus, similar to what would be experienced during game play. Overall, through understanding the risk factors of ACL injuries it might be possible to reduce the amount of injuries that occur each year by changing athlete training protocols or teaching safer movement techniques.

Statement of the Problem

Previous research on noncontact ACL injury has identified many injury risk factors in athletes across multiple sports (i.e. soccer, basketball, volleyball, football). However, much of the research that has been conducted lacked in representing game-like environments due to the need to be in a laboratory setting for data collection. With this in mind, testing protocols need to find ways to better represent game-like environments that an athlete experiences to better understand the mechanisms of the noncontact ACL injury.

Therefore, the purpose of this study was to determine the effects of a physical overhead goal and a subsequent unanticipated lateral cut on the kinematics, kinetics, and muscle activation of the hip and knee of the dominant leg during the landing of a stop vertical jump task (volleyball block).

Research Hypotheses

Due to the limited amount of studies that have simulated active game-time environments, this study was an exploratory investigation to determine if there were any differences at the knee and hip while performing a stop jump with a subsequent unanticipated task.

It was hypothesized that performing the stop jump task with a physical overhead target condition would decrease the hip and knee flexion angles, increase knee abduction angles, and increase the peak hip and knee extension moments and knee adduction moments. It was further hypothesized that performing a subsequent unanticipated task (UA) immediately after the jump landing would further decrease hip and knee flexion angles while increasing peak hip and knee extensor moments and increase the knee abduction angle and knee adduction moment. Lastly, it was hypothesized that the Q:H CCI would increase during the unanticipated tasks when compared to the anticipated tasks.

Independent Variables

- External focus condition: stop jump with no target, stop jump with a target
- Anticipation of lateral cut: anticipated side cut off the dominant leg after landing with prior knowledge (A), unanticipated posterior 45° side-cut, a second vertical jump, or stick (no movement after landing) landing after landing without prior knowledge (UA)

Dependent Variables

- Kinematic Variables:
    - Sagittal plane joint angles:
        - Initial contact hip flexion
        - Initial contact knee flexion
        - Peak hip flexion

        - Peak knee flexion
    - Frontal plane joint angles:
        - Initial contact knee abduction
        - Peak knee abduction
- Kinetic Variables:
    - Sagittal plane joint moments:
        - Peak hip extensor moment
        - Peak knee extensor moment
    - Frontal plane joint moments:
        - Peak internal knee adduction moment
- EMG Variables:
    - Pre-contact quadriceps-hamstrings co-contraction ratio
    - Post-contact quadriceps-hamstrings co-contraction ratio

Limitations of the Study

- Running shoes of the same style were worn by all participants to ensure continuity of results. However, they do not represent the footwear the athletes wear while playing volleyball. Therefore, the results may not accurately represent true game-time responses.
- The height of the timing gate that triggered the visual stimulus in the unanticipated conditions was set at the same height for all participants. This may have affected the natural blocking motion of some participants because they may have had to exaggerate motions to trigger the stimulus which may have altered landing mechanics.

Delimitations of the Study

- Participants' age ranged from 18 to 30
- Participants must have had played volleyball at a high school level or higher for at least 2 seasons.
- Participants who ever experienced a lower extremity injury that required surgery, and ACL injury, or who had suffered a lower extremity injury in the 6 months prior to testing were excluded.
- Participants must have been recreationally active for 30 minutes at least three times per week, with one of the three sessions involving dynamic movement such as running with jumping or cutting.
- Participants who scored a 71 or below on the Lower Extremity Functional Scale (LEFS) were excluded.

Assumptions of the Study

- All participants were truthful about their injury history and activity levels.
- All participants were healthy at the time of testing.
- The two force platforms (BP600600, Advanced Mechanical Technology, Inc., Watertown, MA, USA) and the twelve-camera infrared motion capture system (Vicon Motion Analysis, Inc., Centennial, CO, USA) were properly calibrated for each participant.

Operational Definition of Terms

- Noncontact ACL injury was defined as no contact of an individual with one or more other persons when the injury occurs.

- Initial contact was defined as the moment when the vertical ground reaction force was greater than 10 Newtons.
- The time frame of reference was from initial contact with the ground to peak kinematic angles.
- Joint moments were reported as internal joint moments, which are the torques of the joints to resist an external torque on the body.
- Joint kinematics and kinetics followed the right-hand rule convention:
- Hip: flexion (+) / extension (-)
- Knee: flexion (-) / extension (+), adduction (+) / abduction (-)

CHAPTER 2: REVIEW OF THE LITERATURE

Introduction

The purpose of this study was to determine the effects of a physical overhead goal and a subsequent unanticipated lateral cut on the kinematics, kinetics, and muscle activation of the hip and knee of the dominant leg during the landing of a stop vertical jump task (volleyball block). This chapter will review the current literature regarding: 1) the anterior cruciate ligament (ACL) anatomy, 2) biomechanics of noncontact ACL injuries, 3) the differences between single leg and double legged landing and cutting lower extremity biomechanics, 4) gender differences influencing ACL injury, 5) the effect of anticipation on lower extremity biomechanics while performing dynamic tasks, 6) and the effect of external focus on lower extremity landing biomechanics.

ACL Anatomy

The ACL is one of two cruciate ligaments of the knee which can be divided into two anatomically functional bundles, the anteromedial and posterolateral bundles (32-35). The two-bundle model has generally been accepted as the best representation to understand ACL function (34). The anteromedial bundle attaches posteriorly and superiorly on the medial surface of the lateral femoral condyle and the posterolateral bundle attaches anteriorly and inferiorly on the lateral condyle (32). The boney attachment of the ACL can range from 11 to 24 mm wide (34), located on the femur at the posterior part of the inner surface of the lateral femoral condyle in a crescent shape (32, 34). From the femoral attachment, the ACL runs anteriorly, medially, and distally to the tibia. The length of the ACL ranges from 22 to 41 mm with a mean length of 38mm and has a width of 7 to 12 mm with a mean width of 11 mm (34, 36). The tibial attachments of the bundles form a triangle directed posteriorly, with the anteromedial bundle attaching on the medial aspect of the intercondylar eminence of the tibia while forming the medial corner of the triangle (32). The posterolateral bundle attaches just lateral to the midline of the intercondylar eminence, forming the posteriorly directed point of the triangle (32).

ACL Injury

ACL injuries are becoming more prominent in multidirectional sports, especially in younger populations (3), with an estimated 120,000 to 250,000 ACL injuries in the United States alone and more than 50% of these injuries occurring in athletes aged 15 to 25 (37-39). At the collegiate level in the United States from the 1988-89 season to the 2003-04 season, approximately 5,000 ACL injuries occurred in the sample which represented about 15% of the population, so it was estimated that about 2,000 ACL injuries happen annually in NCAA athletes each year (40). Over a 10 year period at one clinic in Switzerland, 1580 ACL injuries were treated out of a total of 7769 injuries to the knee area (41). They also reported that 47 of the 78 knee injuries specifically in volleyball, were ACL injuries.

The ACL acts to stabilize the knee joint and prevent anterior translation of the tibia in relation to the femur (36) and hyperextension of the knee (1). The ACL is responsible for about 86 % of the total force resisting anterior draw near full knee extension (34, 36), with a slight decrease in the percentage of force as the knee flexion increases from near full extension (34, 42). ACL injuries occur when the load that is applied to the ACL exceeds the strength of the ligament itself (4, 5), and can occur in sporting activities or during daily tasks (6). In multidirectional sports there are two types of ACL injuries that occur during game play, contact injuries and noncontact injuries. Contact injuries occur when two or more athletes collide with one another with direct contact to the knee increasing the load to the ACL. Noncontact injuries occur when there is no physical contact between athletes or direct contact to the knee and the load on the ACL exceeds its limit causing injury.

Noncontact ACL injuries make up about 70% of the total amount of ACL injuries that occur each year in the United States alone (8, 43), with females having a four to six-fold greater injury risk than males (1, 44). This type of injury usually occurs during activities that include rapid deceleration, such as landing (8), or sudden changes in direction such as pivoting and cutting (45). With the amount research that has been conducted on noncontact ACL injuries in recent years increasing due to the prevalence of this injury, multiple risk factors have been identified that might help in understanding why these injuries occur. Some of the major modifiable risk factors include knee abduction angle, low hip and knee flexion angles at ground contact, weakness of the knee flexors and hip abductors, and delayed activation of the knee flexors after initial ground contact (11).

However, the exact injury mechanism during jump landing tasks remains unclear. This could possibly be due to the controlled nature of the tasks being performed in the testing environments (i.e. lab) that do not recreate a real "game time" sport environment for the participants (4). Even from observing film of injuries athletes have sustained while in game play, the specific mechanism of noncontact ACL injury has not been identified (46). ACL injuries have commonly been thought to occur during one single instance where the load applied to the tissue surpasses its ultimate strength. While film has not identified the definite mechanism of ACL injury, many similarities in kinematics have been identified through this film. Studies of film of noncontact ACL injury have shown that the foot lands initially with the hindfoot or the entire foot flat at initial contact with the ground, increased knee abduction angles after initial contact, and increased hip flexion after initial contact (47).

Noncontact ACL injuries have continued to be increasingly prevalent, even with the increased research efforts on the subject. However, multiple risk factors have been identified that might help decrease the number of injuries that occur each year. Of the risk factors, there are two main types that are observed in athletes which are non-modifiable and modifiable. Non-modifiable, or intrinsic risk factors cannot be influenced by training to help reduce the risk of ACL injury. Examples of non-modifiable factors include age, gender, hormones, and structural features of the lower extremity that cannot be changed unless surgery is attempted (48). Modifiable, or extrinsic risk factors can be changed through practice of different techniques to modify lower extremity biomechanics or the strengthening and neuromuscular training of the lower extremity. Practicing safer movements or entering a training protocol for better protection of the knee can keep athletes healthier during game play and can also possibly decrease other lower extremity injuries for occurring in athletes.

Biomechanical Factors of ACL Injury

The majority of what is known about the ACL is based on *in vitro* cadaver experiments under various loading conditions (33), which have been performed to try and determine the ultimate tensile strength of the ACL just prior to rupture or loading under different knee kinematic conditions. In 1976, Noyes and Grood (49) reported that the average ultimate load for three femur-ACL-tibia complexes from donors between 16 and 26 years of age was 1730 ± 660 N. Later in 1991, Woo et. al (42) reported that the average ultimate load for the ACL in specimens under 35 years of age was 2160 ± 157 N, which was a 30% increase than what was reported by Noyes and Grood. This data can be used to estimate the ultimate tensile strength of the ACL and determine at what point the ACL is at risk of rupturing.

At the knee, the main modifiable risk factor in the sagittal plane is a flexion angle of less than 30°, and with the center of gravity behind the knee (11, 15). During athletic tasks, landing with decreased knee flexion can cause an increase in the peak posterior ground reaction force (GRF) (50), which causes an external flexion moment of the knee that must then be balanced by an internal extension moment caused by the contraction of the quadriceps (51). This internal extension moment of the knee causes anterior translation of the tibia in relation to the femur and loads the ACL.

At the hip, the most common kinematic variable that has been found in noncontact ACL injuries was decreased hip flexion during landing (18). Hip flexion is influential on knee flexion. Under normal muscle activation, when the hip flexes so does the knee (52). However, if the co-flexion of the hip and knee are impaired in any way, there will be an increase in anterior tibial shear force, placing a greater load directly on the ACL (52). Through video analysis of ACL injury, increased hip flexion angles at initial contact of dynamic movements were observed in injured subjects compared to healthy controls (47). This decreased hip flexion angle may not allow for the hip extensor muscles to absorb the energy of the body effectively and cause the quadriceps to contract at a higher rate to compensate for the hip extensor muscles and produce greater anterior shear force at the knee (47, 53).

During landing, peak loads on the ACL occur at or just after initial contact when the knee is near full extension (between 0° and 30°), (16). During a stiffer landing, the vertical ground reaction force has a greater magnitude, so the tissues of the lower extremity need to compensate for this greater force. Most of the load applied to the knee in the sagittal plane is caused by the internal extension moment. This moment is generated by contraction of the quadriceps and is the main contributor to anterior tibial shear force (54). Cadaveric studies of the ACL have found that the highest levels of ACL loading occur when there are combinations of anterior tibial translation at shallow knee angles in the sagittal plane (7). In other words, the GRF vector during landing is directed at or posterior to the knee, increasing proximal anterior tibial shear force.

In the frontal plane at the knee, the main modifiable risk factor of noncontact ACL injury is high levels of abduction angles (11). Internal knee adduction moments have been shown to increase with the increase of knee abduction angles during athletic tasks. It is likely that increased knee abduction angles and moments are the primary predictors of ACL injury (29). The findings of Pollard et al. (17) supported the theory that when motion is limited in the sagittal plane a strategy of reliance of passive restraints in the frontal plane is used to control and decelerate the body. There has been a relationship of increased hip adduction angle with increased knee abduction angles. Leetun et al. (55) stated that athletes who sustained ACL injury demonstrated decreased hip abduction strength, and this decrease in hip abduction strength may influence the increased knee abduction that is demonstrated during athletic tasks.

The frontal plane kinematics and kinetics also play a large role in the loading of the knee joint. When the knee abduction angle increases during initial contact with the ground, there is a greater internal knee adduction moment. In the frontal plane the quadriceps and hamstrings can help to resist abduction loading of the knee via co-contraction and when both muscles produce internal extension and flexion moments respectively (56, 57). At the same time at the hip, increased knee abduction moments indicate a decrease in the ability of hip and core musculature to absorb the kinetic energy of the body during the deceleration phase of the landing (18). Markolf et al. (5) found that loading to the ACL from anterior shear force, knee abduction/adduction, and internal rotation moments increased as knee flexion angle decreased.

In a similar study, Kiapour et al. (58) found that greater peak ACL loads were observed under higher levels of knee abduction and internal tibial rotation, but that the knee abduction moment had a much greater effect on ACL loading.

In the transverse plane, external tibial rotation and internal rotation at the hip have also been identified as risk factors for ACL injury (15). Increased external tibial rotation moment has been shown to increase loading on the ACL (5, 58). At the hip, increased internal rotation of the hip has been observed during many dynamic tasks, which may result in altered alignment of the lower extremity and place the athlete at a higher risk for ACL injury (59). Increased internal rotation of the hip has been associated with increased loading of the ACL when included with increased hip adduction angles resulting in a medial translation of the knee in regards to the foot (18).

During dynamic tasks, the muscles of the lower extremity must work to absorb the kinetic energy of the body during the deceleration phase of the movement (54). In jump landings specifically, DeVita and Skelly found that soft landings (defined as knee flexion angles greater than 90°) produced lower extremity joint moments that had lower peak values. This means that the muscle tissue around those joints were more efficient at absorbing the kinetic energy of the fall and decreased the total vertical GRF of the landing (60). Norcross et al. (54) added that greater the total energy absorption in the initial impact phase of the landing (first 100 milliseconds (ms) after initial ground contact) was associated with greater peak vertical GRF's, anterior tibial shear force, and internal hip-extension moment, which have been considered risk factors for non-contact ACL injuries.

Neuromechanics of ACL Injury

Altered muscle activation patterns and muscular imbalances have been associated with increased ACL injury risk. Electromyography (EMG) of the quadriceps and hamstrings has been used to determine the onset of both sets of muscles around the knee. The high activation of the quadriceps during an athletic task such as a landing or a cut can cause increased anterior tibial shear force, which has been shown to load the ACL through the patellar tendon (5, 20). However, the activation of the hamstrings can provide a counterbalancing force to protect against the relatively higher quadriceps force, thus reducing the anterior tibial translation and decreasing ACL strain (21). This ratio of contraction between the quadriceps and hamstrings is the Q:H co-contraction ratio. If the quadriceps alone are activated prior to contact, the knee joint will extend rapidly, placing the knee in a dangerous position for landing. However, co-contracting the quadriceps and hamstrings prior to foot contact will allow the quadriceps to be active at touchdown decreasing torque at the knee joint (61). At the same time, proper activation patterns, such as a more balanced Q:H ratio, are necessary to reduce the risk of injury (21).

The pre-activation of the quadriceps and hamstrings plays a major role on the strategy of landing (62). Alves de Britto, et al. (63) investigated the prelanding normalized EMG activations of the vastus medialis, rectus femoris, biceps femoris, and the medial hamstrings in both males and females from different heights. One of the major findings of this study was an increase in quadriceps pre-activation as jump height increased, but no change in hamstrings pre-activation as jump height increased. This data is similar to the work of Chappell et al. (64), who found that hamstring activation was essentially the same for both males and females at the time of landing from a vertical stop-jump task, but females then exhibited lower hamstring activation after the landing compared to males. Specifically, Chappell et al. (64) found that both males and females had an increase in quadriceps EMG about 50 milliseconds prior to landing. Additionally, both males and females had an increase in hamstring EMG activity during the flight phase, but females also had a 12% increase in quadriceps EMG during the flight phase of the jump (64). Both of the studies found that females had an increased pre-activation of the hamstrings, but Chappell et al. (64) found that the difference in pre-activation that females displayed during the flight phase decreased to almost no difference just prior to landing. This is because males had an increased rate of activation in the moment just before landing, whereas females showed pre-activation earlier in the flight phase (64). This differs slightly from what was found by Alves de Britto et al. (63), who found that the medial hamstrings in females was activated at a much greater percentage of the normalized EMG prior to landing compared to males.

One modifiable neuromuscular imbalance is quadriceps dominance (QD) which is defined as the imbalance between knee extensor and flexor strength, recruitment, and coordination (19). QD refers to the stabilization of the knee joint by primarily the quadriceps muscles and it can alter the co-contraction of the hamstrings with the quadriceps. Interestingly, Blackburn and Padua (65) found that increased trunk flexion during a landing led to decreased vertical ground reaction forces and decreased quadriceps activity. A change such as this could allow for athletes to decrease the risk of anterior tibial shear force from the decreased contraction of the quadriceps. These modifiable neuromuscular factors can be influenced by training as well. In a study by Dai and colleagues (27), EMG of the vastus lateralis and biceps femoris was collected on collegiate volleyball players during a stop-vertical jump. After the detraining period represented by the offseason of play, the players had a decrease in prelanding EMG for both muscles. These reductions in the hamstring EMG could be associated with reduced knee flexion angles at initial foot contact during the landing, and with no change in quadriceps activity after initial contact, there would be an increase in anterior tibial shear force because decreased hamstring activity would not be able to counteract the force produced by the quadriceps, placing the player at a higher risk of excessive loading of the ACL (27).

General Sex Differences

Females are nearly 2.5 times more likely to experience an ACL injury playing volleyball when compared to males (23). There are both extrinsic and intrinsic differences between males and females. Some intrinsic risk factors that are exhibited more frequently in females include: high Q angle, increased joint laxity, decreased ACL size, decreased femoral notch sizes, and increased posterior tibial and meniscal slopes (24-26). Extrinsic risk factors that females exhibit more often than males include: higher knee abduction angles at initial contact, a decrease in the peak knee flexion angles during the dynamic movements (20, 29), and higher quadriceps activation and lower hamstring activation than male athletes (19-21, 27, 28).

There has been evidence of differences in lower extremity biomechanics between males and females. The intrinsic risk factors that females exhibit have been researched extensively. The Q angle is measured as the angle between a line connecting the anterior superior iliac spine (ASIS) and the midpoint of the patella, and a line between the midpoint of the patella and the tibial tubercle (66, 67). Because females have a wider pelvis than males, females have a greater Q angle which places them at increased knee abduction angles at rest than males. Females have also been identified as having increased posterior tibial slopes. If the posterior tibial slope is greater, there is an increase in anterior tibial subluxation and internal tibial rotation, which can both strain the ACL (24). Greater lateral meniscal slopes have also been identified in subjects with ACL injury where it cannot restrain excess anterior tibial motion (24). The decreased ACL size in females means that there is a lesser maximal force that can be applied to the tissue before failure compared to males with larger ACL size. The decreased size of the femoral notch can cause impingement of the ACL during rotation of the femur and the tibia, however there is some controversy as to whether the femoral notch size is a risk factor for ACL injury (24).

One influence of increased ACL injury risk may be hormonal differences between males and females. Studies performed by Yu, et al. (68, 69) found results that the acute changes in the hormones during the menstrual cycle could influence ACL metabolism and collagen synthesis. There has been some research that has tried to find a relationship between the time of the menstrual cycle and the incidence of ACL injury. While some studies have found relationships, there is still no clear consensus in the literature that hormonal levels or the time during the menstrual cycle have an impact on ACL injury (38).

Differences between females and males are also observed in extrinsic factors. Females have been shown to display differences in kinematics and kinetics during athletic tasks that place them at higher risk for ACL injury. Females that went on to ACL injury displayed higher knee abduction angles at initial contact, and a decrease in the peak knee flexion angles during the dynamic movements (20, 29). Norcross et al. (70) reported that there was not a difference between genders in sagittal plane mechanics, meaning that energy absorption in the sagittal plane during the initial contact phase does not directly impact the biomechanics in the frontal plane. This was supported by Hewett et al. (29) who found no difference in knee flexion angles between females who experienced ACL injury and those who did not, but did see a large increase in knee abduction angle at contact in those who became injured. This is also in agreement with Pollard and colleagues (59), who observed that females demonstrated higher reliance on the frontal and transverse planes during a dynamic task, specifically greater hip internal rotation, greater internal hip adductor moments and decreased hip flexion angle and internal hip extension moments during early deceleration.

Many studies have investigated co-contraction ratio differences between males and females. Female athletes tend to have higher anterior tibial resultant forces by having a higher quadriceps activation and lower hamstring activation than male athletes (19-21, 27, 28). Females also have increased prelanding quadriceps and hamstrings activity compared to males (64). Increased prelanding quadriceps EMG activity will increase the anterior tibial shear force, and while increased hamstring activity would help to protect the ACL, the increases in hamstring activity for females was less than the males exhibited (64). Females also demonstrate unbalanced medial and lateral Q:H co-contraction ratios with decreased medial co-contraction, resulting in a diminished ability to resist internal adduction and extension loads (19). The imbalanced medial and lateral muscle activation can lead to increased internal knee adduction and knee extension moments can contribute to ACL loading, especially in stiffer landings.

ACL injuries are becoming more prominent in athletics, especially in the female population. Although there is little that can be done to change intrinsic risk factors, extrinsic factors can be modified through practice or training protocols, and understanding each kinematic, kinetic, and neuromechanical risk factor can help to reduce the amount of non-contact ACL injuries that occur each year. It is also important to understand the sex differences, both anatomically and biomechanically, between males and females that could have an influence on the higher incidence rate in females compared to males.

Biomechanics of Landing and Cutting Tasks

Bilateral Landings

When it comes to noncontact ACL injuries, jump landings are one of the most dangerous situations for athletes (11). Injuries are most likely to occur almost immediately after initial ground contact (8), with peak ACL strain occurring at the same time as the first vertical GRF peak (10 - 14 ms post-contact) (16). Many studies have focused on different directional types of jumps and their impact on lower extremity biomechanics, especially in the sagittal and frontal planes. Taylor et al (7) found that between a sagittal plane double-legged broad jump with a subsequent maximum vertical jump and a frontal plane double-legged lateral jump with subsequent maximum vertical jump, frontal plane landings were stiffer with decreased hip and knee peak flexion angles.

A critical knee flexion angle of less than 30° has been identified during landing which results in a large knee extension moment on the ACL caused by the contraction of the quadriceps during the passive loading phase (16, 18). Shimokochi et al. (8) compared knee flexion angle during upright landing (trunk in a more vertical position), a leaning forward landing, and a self-selected landing. They reported that knee flexion angle at initial contact during upright landings was -26.7±0.5°, -28.9±11.4° during self-selected landings, and -37.5±9.5° during leaning forward landings. This data is consistent with Zahradnik et. al (16) (three landing conditions with initial contact knee flexion angles of less than 30°). The three landing conditions that were performed by female volleyball players were the go, reverse, and rotation conditions. For each of these conditions, there was a knee extension moment of 0.29±0.23 Nm/kg, 0.32±0.24 Nm/kg, and 0.13±0.16 Nm/kg for each of the condition respectively at the time of peak resultant ground reaction force (16). When compared to the other tasks, these three movement patterns showed on average to have greater knee extension moments than the conditions that had greater knee flexion at initial contact. These studies show that altering trunk position during landing to a more forward leaning position can help decrease the vertical GRF and increase the hip extensor moment. Therefore, this altering of trunk position allows the muscles of the hip and core to absorb more energy from the landing. In other words, the torso is further behind the knee and knee extension torque is required to stabilize the knee joint (71).

Zahradnik et al. (72) also showed that in a volleyball blocking task, when a run-back landing was performed, there were increased levels of external knee abduction moments compared to a stick landing or a step-back landing. The peak abduction moment was not observed during the initial landing phase, but when the subject was beginning to actively move away from the net (72). One possible reason for the increased knee abduction moment may be an altered landing pattern that placed the subjects at higher levels of internal tibial rotation at initial contact in an attempt to retreat back to their position quicker than the step-back landing (72).

Unilateral Landings

The majority of noncontact ACL injuries occur during unilateral foot contact (7, 12, 13). Although, there has been little variance in frontal plane lower extremity mechanics during bilateral landings between trials, that is not the case in unilateral landings. In the aforementioned study by Taylor et al. (7), significant interactions between jump direction and landing type (bilateral or unilateral) were present. Unilateral landings in the sagittal plane task had greater hip adduction moments, smaller hip and knee flexion angles, and higher knee internal rotation moments. In the frontal plane, there was a greater external knee abduction moment. Taylor et al. (7) also found that, compared with the bilateral landing in the sagittal plane (forward jump conditions), all other conditions resulted in stiffer landings. Additionally, the unilateral tasks had the greatest joint moments, and the frontal plane movements (lateral jumps) showed more signs of dynamic knee abduction. The increased knee abduction angles in the frontal plane movement is consistent to results published by Sinsurin et al. (12), which compared knee abduction angles with jump direction in a unilateral jump landing. The lateral jump condition had an average knee abduction angle of 8.8°. This is important because knee abduction angle is one of the predominant actions and predictors of ACL injury (54).

During unilateral landings, the strength of the surrounding muscles and hip muscles can be an important predictors of knee kinematics during the deceleration phase of the landing (44). Suzuki et al (44) noted, however, that hip strength influenced knee kinematics in different ways, indicating that hip strength is not the only factor influencing knee kinematics. Norcross et al. (70) concluded that energy absorption could be changed by muscular strength and activation, joint positioning, and the magnitude of joint motion during landing which would allow for less high-risk knee and hip kinematics and kinetics during landing. Increasing energy absorption due to eccentric contractions of the muscles will decrease the load applied directly to the ACL (70).

Landing with a single-leg can also influence the vertical GRF. Ali et al. (50) found that the vertical GRF curve in a single-legged landing from various heights and distances produced a single peak GRF that was much closer to initial contact when compared to a double-legged landing. This difference means that there is a unique difference to the landing patterns observed in single-legged landings compared to double-legged landings, and that findings from studies using strictly double-legged landings may not be comparable to single-legged landings.

Lateral Cutting

Cutting maneuvers are essential for successful performance in many sports (73). The lateral shuffle motion is common in sports such as basketball and volleyball and is used to improve agility on other sports and fitness training (14).

When comparing a lateral shuffle to a 45° side-cut, Zaslow et al. (14) found that a lateral shuffle and side-cut elicited very similar knee abduction angles and moments. However, one difference between the two movements was that the lateral shuffle movement did not produce as much internal hip rotation as the side cut. This means that the lateral shuffle may not be challenging enough to expose deficits in hip control (14). If the lateral shuffle is not the primary cause of the internal tibial rotation, there will be less stress placed on the knee joint, which means that the hip extensor muscles may play a greater role in controlling the body in the sagittal plane during the lateral shuffle.

Gender Differences in Landing and Cutting

One reason that gender differences may exist is because of the influence of hip strength in athletes that are performing multidirectional movement tasks such as jump landings or cutting. Suzuki et al. (44) reported that females who had higher levels of hip extension, hip abduction, and hip external rotation strength demonstrated decreased knee abduction angles during unilateral landing, while males with greater hip extension strength demonstrated increased knee flexion angles during unilateral landing. Previous studies have also identified that females exhibit a greater knee abduction angle at initial contact with the ground and greater peak knee abduction angles during landing (17, 70, 74). Leetun et al. (55) observed through one female athlete who injured her ACL that there was a major deficiency in core strength. The athlete scored well below average on the side bridge core stability test which was found to be significantly correlated to the performance in all other postural muscle tests, and athletes who experienced other lower extremity injuries tended to demonstrate lower core stability measures (55).

With increased knee abduction angles comes increased knee abduction moments. Females display increased knee abduction moments through the later stages of adolescence to adulthood (74). A cadaveric study by Schilaty et al. (75) analyzed both male and female specimens that were loaded to simulate a drop vertical jump based on the kinematic and kinetic variables of healthy, athletic participants. They found that in drop landing situations without the influence of neuromuscular control, female specimens experienced increased knee abduction moments under similar loading conditions to males (75). Because the vertical GRF in all trials were consistent, other mechanisms such as knee abduction angle influence resultant forces and moments at the knee. Overall, with the lack of neuromuscular control females will experience greater knee abduction angles and greater knee abduction moments during a drop landing task (75). This potentially places females at higher risk for ACL injury.

Bilateral and unilateral landing tasks can produce movement patterns that increase the risk for ACL injury, including decreased knee flexion angle at initial contact and increased hip flexion angle at initial contact. Unilateral landings have also been shown to produce greater knee abduction angles, especially in populations with decreased strength around the knee. While these risk factors have been identified, it is important to include un-anticipation into the athletic tasks to try and elicit a realistic response to the task that would represent a game-like situation.

External Goals and Anticipation and its Impact on Landing Biomechanics

External Goals

Including an external physical goal is a step toward making testing in a laboratory more realistic to what actual sport situations would be. Including an extrinsic motivator has been found to improve jump performance, mainly jump height (30). Additionally, it is possible that including a physical goal (i.e. object) in the testing environment and having the subjects reach for that object can potentially lead to a high probability of changing lower extremity joint kinetics and kinematics (30). Mok et al. (31) found that including on overhead target to a drop vertical jump task increased jump height by 5.8%. The increase in jump height also led to increases in the moments due to a 5% increase in the peak vertical GRF during the landing. In the study done by Ford and colleagues (30), including a physical overhead goal resulted in the subjects landing with a more erect trunk posture. This information, along with the previously stated findings of Shimokochi et al. (8), adds to the notion that landing with a more erect trunk causes a decreased internal hip extension moment which means less energy absorption during the landing from the hip musculature (30). With less energy absorbed by the musculature of the hip, there is a greater load that is applied around the knee joint due to increased anterior shear force from quadriceps contraction to absorb the energy of the body. When the vertical GRF and peak internal knee extension moment are increased, there is an increased load on the ACL placing the ligament at a higher risk for injury (54).

Another technique that has been used in landing studies to alter the “setting” or mental environment of a subject is the inclusion of a secondary cognitive task while performing the physical jumping and landing task. Influencing the attention of the subject to not be solely focused on the physical task of jumping and cutting can help create a more game-like, cognitive environment for the subjects.

In most multidirectional sports including a ball, athletes will focus on getting possession of the ball during a jump and will pay little attention to safe landing. When including a secondary cognitive task of counting backwards by ones during a drop jump task with a subsequent vertical jump, Dai et al. (4) reported a significant decrease in the initial knee flexion angle from 24.9±6.4° to 23.0±5.9° when compared to the control “no counting” condition. They also found that there was a significant increase in the vertical (no count: 2.82±0.52 BW, 1’s: 3.10±0.74 BW, 7’s: 3.06 ±0.77 BW) and posterior GRF’s (no count: 0.80±0.22 BW, 1’s: 0.84±0.23 BW, 7’s: 0.86±0.26 BW) in the cognitive conditions when compared to the control condition (4). The inclusion of the cognitive task to the jump landing caused changes to the lower extremity biomechanics in a similar fashion to the changes caused by the inclusion of a physical goal seen by Ford et al. (30) who observed . Furthermore, in a recent study by Almonroeder et al. (76), they found that the influence of additional cognitive demands to a landing task in general resulted in a stiffer landing and greater knee abduction angles in female athletes. However, combining decision-making tasks and a focus of attention did not alter the lower extremity biomechanics to a level that exceeded what each individual cognitive demand had on the lower extremity biomechanics. This means that each cognitive demand influenced the landing mechanics of the subjects, while adding more demands into one condition did not alter the landing mechanics to a greater degree than the single demand. Due to this, including a single cognitive demand may influence differences in landing mechanics during the primary jumping task and may provide a more realistic insight into understanding the mechanisms of game time ACL injuries.

Anticipation While Performing Athletic Tasks

Unpredictability is another aspect of real-time game play that has an impact on noncontact ACL injuries. With the inclusion of an external secondary goal, there is a much greater chance of obtaining results in a laboratory setting that are more realistic to what would happen in a game, increasing external validity of the experiment. Many anticipation biomechanical studies investigating landing mechanics have participants complete the tasks in preplanned actions, where these movement patterns do not accurately represent the dynamic athletic environment (77). However, while working in a laboratory setting with limited space, there is only so far that researchers can go to make the testing environment as realistic as possible. One way to do this is including unanticipated or unpredictable conditions to the testing protocol. Previous studies have stated that knee joint loading was greater during unpredictable conditions when compared to predictable conditions, increasing the risk of ACL injury (78).

Meinerz et al. (77) reported that during an unanticipated land and cut, participants exhibited increased hip-energy absorption (A: -0.84±0.39 W/kg, UA: -1.06±0.49 W/kg), peak internal hip external-rotator moment (A: 49.0±12.4 Nm, UA: 59.9±11.7 Nm), internal hip abductor (A: 57.5±17.7 Nm, UA: 72.1±18.5 Nm) and internal hip extensor moments (A: 168.0±47.0 Nm, UA: 190.6±66.1 Nm), and peak gluteus maximus muscle activation when compared to the anticipated condition. These movement strategies could help to prevent excessive internal rotation of the femur, which in turn would reduce the knee abduction angle and internal knee adduction moment which is thought to decrease the risk of injury (77). These results differ from those gathered by Mornieux et al. (78), who found that during an unpredictable lateral jump to a lateral sliding surface, subjects experienced increased hip joint abduction and increased vertical GRF, as well as reduced knee (anticipated: 26±3.5°, unanticipated: 24±3.7°) and hip flexion (anticipated: 30.6±7.2°, unanticipated: 27.4±7.2°) at initial contact. This finding by Mornieux et al. (78) is in agreement with Sell et al (46). They found that direction and reaction to lateral jumps in either direction influenced knee joint loading in the sagittal plane at the moment of peak posterior GRF (anticipated: -.040±0.045 body weight•height, unanticipated: -0.054±0.039 body weight•height) and with subjects performing the unplanned tasks with greater knee internal rotation moments.

These differences in kinematics and kinetics can be explained by the idea of preplanned motions, or anticipation. There have been suggestions that successful movements during landing are mainly due to preplanned neuromechanical control strategies (77). Both Mornieux et al. (78) and Meinerz et al. (77) concluded that the results of their unpredicted conditions may be due to preprogrammed neuromechanical control strategies that allow for altered landing mechanics to remain safe while landing. This idea of preprogrammed mechanics is important to understand, especially for young athletes and coaches of young athletes. Learning how to land properly can develop those preprogrammed mechanics to reduce injury risk during dangerous landing situations.

Including external goals and anticipation into the experimental tasks may create a more realistic environment while still controlling as many variables as possible. Including external goals may change the cognitive demands that are required to perform a task and allow for a game/practice like reaction. At the same time, including the aspect of unanticipated movements or tasks have been shown to alter the mechanics of dynamic tasks. Including both of these into testing will allow for the possibility of movements that could actually happen and cause ACL injury.

Summary

Noncontact ACL injuries are becoming increasingly prominent in athletics. Providing more realistic ‘game-like’ experimental tasks to the laboratory setting is essential for new information to be discovered about the mechanisms of injury and in the development of prevention steps for athletes and coaches reduce the risk of this detrimental injury. By understanding the risk factors of noncontact ACL injuries, better conclusions can be drawn by data that comes from experiments that provide a more “game-like” environment. At the knee and the hip, there are extrinsic risk factors such as low knee flexion, internal knee extensor moment, increased knee abduction, increased knee abduction moment, high levels of hip flexion, increased hip adduction, external tibial rotation, and hip internal rotation. Neuromechanical extrinsic factors include quadriceps dominance during athletic tasks, the Q:H co-contraction ratio, and delayed or pre-activation of the quadriceps and hamstrings during athletic tasks. There have also been biomechanical differences observed between males and females that place females at increased risk for ACL injury. In landing tasks, unilateral landings produce kinematics and kinetics that are more high-risk than bilateral landings. Experimental tasks that include unanticipated change of direction or cognitive tasks have also been shown to alter landing kinematics and kinetics and may be necessary to fully understand the ACL injury mechanism.

CHAPTER 3: METHODOLOGY

The purpose of this study was to examine how an external physical goal (i.e., a volleyball) included in a stop jump with a subsequent unanticipated task would affect hip and knee kinematics and kinetics in recreationally active volleyball players. This chapter describes the methods used to conduct the study.

Participants

Thirteen recreationally active females with an age of 18 to 30 were recruited to participate in the study. Each participant needed to have at least two-years of experience playing volleyball at the high school level or higher to be eligible to participate. They also needed to be recreationally active was defined as three or more days per week of 30 minutes or more of moderate to vigorous physical activity, with one of the days consisting of a dynamic activity such as running with jumping or cutting. Participants were excluded from the study if they had a lower extremity injury that required surgery, ever had suffered an ACL injury, or had experienced a lower extremity injury within the six months prior to testing. Participants were introduced to the study procedures and provided written consent before data collection occurred. The informed consent was approved by the University Institutional Review Board prior to testing. Participants completed the Lower Extremity Functional Scale (LEFS) to determine if they have any difficulty in performing different activities. Binkley et al (79) determined that the minimal detectable change at a 90% confidence level was ± 9. Therefore, participants who score 71 and below on the LEFS scale were excluded because they exhibit a clinically meaningful functional change.

Instrumentation

A twelve-camera infrared motion capture system (200 Hz, Vicon Motion Analysis, Inc., Oxford, UK) was used to collect marker coordinate data for each subject. The cameras were calibrated prior to each data collecting session. Each camera collected at least 6000 wand marker counts in the testing area to ensure that each marker on the subject would be collected accurately during the session. The area of the capture volume included above both force platforms to a height that a subject could reasonably reach while jumping with a marker plate on their lower back and to either side of the force platform to collect lateral movements during the unanticipated tasks. Two 60x60 cm force platforms (2000 Hz, BP600600, Advanced Mechanical Technology, Inc., Watertown, MA, USA) were used to measure ground reaction forces. Before each data collection session, each force platform was checked to make sure there is no excess noise that could affect the data. Each force platform was then zeroed before data collection will begin to ensure that the data will be accurate. A single timing gate (63501 IR, Lafayette Instrument Inc., IN, USA) was used to trigger the visual stimulus for the unanticipated conditions.

Wireless EMG sensors were placed on the Vastus Medialis (VM), Vastus Lateralis (VL), long head of the Biceps Femoris (BF), and medial hamstrings (MH) of the dominant leg. Leg dominance was determined by asking the participant which leg they prefer to kick a ball with. A Trigno^{TM} Wireless EMG system and sensors (2000 Hz, Delsys, Inc., Natick, MA, USA) was used because of the high-speed movements that occurred during testing. One sensor was applied to each muscle and the placements of the sensors followed the guidelines from Rainoldi et al. (80), with the sensors being placed on the muscle belly following the muscle fiber direction. The reference line for the vastus medialis was the distance (mm) from the superior medial side of the patella along a line medially oriented at an angle of 50° with respect to the anterior superior iliac spine. The reference line for the vastus lateralis was the distance (mm) along a line from the superior lateral side of the patella to the anterior superior iliac spine, starting from the patella. The reference line for the biceps femoris was the percentage distance from the ischial tuberosity to the lateral side of the popliteus cavity, starting from the ischial tuberosity. Finally, the medial hamstrings reference line was the percentage distance from the ischial tuberosity to the to the medial side of the popliteus cavity, starting from the ischial tuberosity. Sensors were initially positioned after palpation of the muscle during a contraction to determine the muscle size and belly and then adjusted so that the bars of the electrode were perpendicular to the muscle fibers. Muscle activation data was collected at the same time as the marker coordinate data and ground reaction force data using the Vicon Nexus 2.7.1 (Vicon Motion Analysis, Inc., Oxford, UK).

Procedures

Participant Setup

Each potential participant came to the lab for one session to complete the data collection. Each individual was given an informed consent document to read and review with the researcher prior to testing. Participants then changed into spandex shorts with a generic t-shirt and laboratory shoes (Nike, Nike Air Zoom Pegasus 34, USA). Spandex was worn to minimalize the amount of possible movement of the markers to ensure reliability of the data. The same standard lab running shoes were worn by all participants so any changes in the data would not be caused by a difference in the type of footwear. The participants then completed a five-minute warm-up on a treadmill at their own selected pace, followed by stretching until they were satisfied to begin testing. Three maximal effort jump and reach tests were performed to calculate the ball height required for 90% of the maximal jump and reach height using a Vertec Vertical Jump Tester

(Sports Imports Inc., Columbus, OH, USA). Once the average of the three trials was calculated, 90% of that jump height was determined and the middle of the suspended volleyball was placed at that height. If the subject had a jump and reach height that was less than 2.54 m, they were excluded from the study because the ball could not be placed at a height to clear the top of the net. EMG sensors were placed on the four muscles and maximum voluntary isometric contraction (MVIC) data was collected. MVIC tests were done for both the quadriceps and the hamstrings, and this data was used to allow for comparison of the EMG activation levels during the dynamic testing between participants and muscles. An isokinetic dynamometer (Biodex Medical Systems, Inc., Shirley, NY, USA) was used to perform the MVIC trials. Quadriceps MVIC trials were conducted in a seated position with the knee flexed to 90°. Hamstrings MVIC trials were conducted in a prone position with the knee flexed to 30°. Trials began with a three-second ramp-up to 100% of their maximal effort. Maximal contractions were then held for 5 seconds, and three trials were performed on both the quadriceps and hamstrings. EMG MVIC data was pre-amplified and high-pass filtered using a fourth-order, zero lag Butterworth filter with a cutoff frequency of 10 Hz. The signal was also full-wave rectified then low-pass filtered with a cutoff frequency of 10 Hz (81). A one second moving average was then performed on the filtered data and the maximal value was recorded for each trial. The maximum value of the three trials was identified and used to normalize the EMG signal during the motion trials.

Reflective markers were then applied to the participant to define the anatomical structure and identify the segments of the lower extremities and pelvis. The pelvis was defined by the left and right iliac crest markers and left and right greater trochanter markers. Anatomical markers were also placed at the medial and lateral femoral epicondyles on each leg to define the thigh, on the medial and lateral malleoli to define the shank, and on the first and fifth metatarsal to define the foot. Semi-rigid thermoplastic shells with four reflective markers were then attached to Velcro-sensitive neoprene wraps on the pelvis, thighs, and shanks. A semi-rigid thermoplastic shell with four reflective markers was also secured to the lateral heel of each foot using tape to ensure no movement on the shoe. Each of the marker clusters were used to track segment movement during the dynamic movement trials.

First, a static trial was conducted, which consisted of the participant standing still with both feet on one force platform with their arms crossed over their chest and their hands by the opposite shoulder, so all markers are visible to the cameras. Once the static trial was conducted, the anatomical markers were removed so only the tracking markers on the thermoplastic shells remained on the participant. Next, a dynamic range of motion (ROM) trial was then conducted. Participants first kicked their entire leg straight forward, flexing at the hip while keeping the knee at near full extension, and bringing it back to a standing position on the force platform. This was followed by a kick at 45° between forward and lateral to the side of the leg moving, then at 90° or lateral, followed by 45° angle backwards, and finally straight backwards. Then for three times the participant flexed at the hip and at the knee so both joints had a 90° angle followed by extending the knee joint to full extension and then back to 90° flexion. With their knee still bent at 90°, the participant then plantarflexed and dorsiflexed at the ankle three times each. This was followed by three circular motions of the foot around the ankle joint. As soon as the first leg was complete, the participant did the same movements with the other leg. This was then be followed by flexing the trunk forward at the hips three times with both feet on the force platforms. Once the dynamic ROM trial was conducted, the testing data collection began.

Testing Protocol

Prior to testing, each participant was given the opportunity to practice the stop jump and secondary movements of the task. This was be done by participants stepping toward the force plates along the net, stopping on the force plates, performing a bilateral maximal vertical jump while moving the hands up in a blocking motion, and finally, landing with one foot fully in contact with each force platform. Once the participants felt comfortable with the movement and landing on the force platforms, they began to practice the stop jump landing followed by the secondary task. This secondary task was either a 45° posterior/lateral cut to the right, a 45° posterior/lateral cut to the left, no movement from the landing (a stick landing), or a second vertical jump following the landing. When the participant was comfortable with the tasks, the testing space was prepared for the first condition.

Four different testing conditions were performed by each participant with five valid trials per condition. The testing conditions consisted of the following: 1) jumping with no target and an anticipated secondary task (NTA), 2) jumping with no target and an unanticipated secondary task (NTUA), 3) jumping with a ball and net and a secondary anticipated task (WTA), and 4) jumping with a ball and net and a secondary unanticipated task (WTUA). The with-target conditions included a net at the standard 2.24 m that is specified for female volleyball competitions and a ball that was suspended by a rebounder machine with the middle of the ball at 90% of the participants maximum jump and reach height. While jumping in all conditions, participants were required to jump with both hands raised over their heads and extend them forward as if to block the ball. For the anticipated task conditions, participants were told to jump and block the ball, land and cut off the dominant foot at a 45° angle posterior/lateral to the force plates (i.e. right foot dominant cuts off the right foot to the left). For the unanticipated task conditions, a visual cue was given when the participant's hands broke the infrared beam of the timing gate which will be set at 2.33 m from the floor. When the infrared beam was broken, a random number generator (Lab View) produced a value and projected either a left arrow (left cut), a right arrow (right cut), a stop sign (stick landing), or and up arrow (second vertical jump) onto a computer monitor, which was set on a raised platform that was 1.83 m away from the back of the force platforms and at a height of 1.36 m from the floor to the base of the monitor. The unanticipated task trials were deemed valid if the participant immediately moved in the proper direction without stopping after the landing. The only trials that were analyzed in the unanticipated conditions were the landings where the participant cut off their dominant foot in the opposite direction at a 45° angle posterior/lateral to the force plates (i.e. right leg dominant cuts off right foot to the left).

There were eight different condition orders. The order of conditions was ordered in a way that allowed the two with-target conditions to be completed one after another and the two no target conditions to be completed one after another. The need for this was to reduce the amount of preparation time between the conditions for either setting up or removing the net and ball apparatus. A total of at least 50 valid trials were conducted (5 for anticipated, 20 for unanticipated (5 each movement)) and the data collection concluded at that point.

Data Processing and Statistical Analysis

Motion trials were analyzed using Visual3D software suite (v6, C-Motion, Inc., Rockville, MD) which computed kinematic and kinetic data for both legs. The time frame of reference for the processing was during the deceleration phase of the landing after initial contact of the jump landing. Marker coordinate and ground reaction force data were filtered using a fourth-order, zero-lag Butterworth filter with a cutoff frequency of 10 Hz.

A seven-segment skeletal model was produced using the static calibration trial (pelvis, right/left thigh, right/left shank, right/left foot). Kinematic and kinetic data were computed by using the segment coordinates and ground reaction force data. Internally applied, three-dimensional joint kinetics were calculated using a Newton-Euler approach (82), and was then projected to the joint coordinate system (83). All kinetic data were normalized to body mass. Body segment masses were estimated from Dempster et al (84) and segment moment of inertias were estimated using the work of Hanavan (85). Hip joint centers were located at 25% of the distance from the ipsilateral to contralateral greater trochanter marker (86). Knee joint centers were the midpoints between the epicondyle markers (87), and ankle joint centers were the midpoints between the malleoli markers (88).

EMG data were pre-amplified and high-pass filtered using a fourth-order, zero lag Butterworth filter with a cutoff frequency of 10 Hz. The signal was also full-wave rectified and normalized to the MVIC of each muscle over the trial. The full-wave rectified signal was then low-pass filtered with a cutoff frequency of 20 Hz (81). EMG from each trial was expressed as a percentage of the EMG from the MVIC trials (%MVIC).

Quadriceps-to-hamstrings co-contraction index (Q:H CCI) were computed both pre-contact (1) and post-contact (2) for each trial. Q:H CCI was computed as the ratio of average quadriceps activation to average hamstring activation.

$$Pre\text{-}contact\ Q{:}H\ CCI = \frac{(Q_{50ms\ pre-HS})}{(H_{50ms\ pre-HS})} \qquad (1)$$

$$Post\text{-}contact\ Q{:}H\ CCI = \frac{(Q_{HS-100ms\ post})}{(H_{HS-100ms\ post})} \qquad (2)$$

The average quadriceps activation was calculated as the sum of the average VM and average VL activations (3), while the average hamstrings activation was calculated as the sum of the average BF and MH activations (4).

$$Q = (avgVM + avgVL) \quad (3)$$

$$H = (avgBF + avgMH) \quad (4)$$

Data was analyzed using SPSS (v25.0, SPSS Inc., Chicago, IL, USA) to examine the differences between the lower extremity biomechanics for each condition. 2 (WT vs NT) x 2 (A vs UA) repeated-measures analyses of variance (ANOVAs) were performed. All tests were performed with a significance value of $\alpha = 0.05$. If significant main effects were found, the partial eta squared value was reported. If a significant target×anticipation interaction was identified, the partial eta squared from the ANOVA was reported. A pairwise comparison was then run on the variable and the significant difference was reported with the *p*-value and Cohen's d value of the interaction.

CHAPTER 4: THE EFFECTS OF AN OVERHEAD GOAL AND MOVEMENT ANTICIPATION ON LOWER EXTREMITY BIOMECHANICS AND MUSCLE ACTIVATION PATTERNS

Abstract

Noncontact anterior cruciate ligament (ACL) injury is becoming more common among sport-related injury. Several "high-risk" movements such as landings and cutting have been associated with ACL injury along with several biomechanical risk factors. Females are nearly 2.5 times more likely to experience an ACL injury while playing volleyball when compared to their male counterparts. While research continues to investigate the reasoning for this injury, lab-based studies may not accurately represent game-time biomechanics in volleyball athletes. The purpose of this study was to determine the effects of a physical overhead goal and a subsequent unanticipated lateral cut on the kinematics, kinetics, and muscle activation of the hip and knee of the dominant leg during the landing of a stop vertical jump task (volleyball block). Participants completed four testing conditions – with a target and an anticipated cut following the landing (WTA), and an unanticipated secondary cut (WTUA), and then both anticipated and unanticipated secondary movements with no overhead target (NTA, NTUA). Significant interactions were present for peak knee abduction moments and the IC hip flexion angles specifically when the target is present. Additional results indicated that when a target was added to the testing environment, there was a significant decrease in the IC knee flexion angle and an increase in the peak knee extension moment. There were decreases in the IC knee abduction angles and peak knee extension moments in the unanticipated task compared to the anticipated task. These findings suggest that the more realistic testing environment altered hip and knee landing mechanics, which in turn can be used to help better evaluate the risk factors for ACL injury in volleyball players.

Introduction

The anterior cruciate ligament (ACL) is one of the most important stabilizing ligaments of the knee (1). An estimated 100,000 to 250,000 ACL injuries occur every year (1, 2), with injury rates becoming more prominent in multidirectional sports and younger populations (3). ACL injuries occur when the load applied to the ACL exceeds the strength of the ligament itself (4, 5). This load can become too great both in sporting activities and during daily tasks (6). In multidirectional sports, both contact and noncontact ACL injuries can occur during game play, with approximately 70% of these injuries occurring in noncontact situations (7, 8). Noncontact ACL injuries occur during activities that include rapid deceleration, such as landing (8), or sudden changes in direction such as pivoting and cutting (10). Furthermore, females are nearly 2.5 times more likely to experience an ACL injury in volleyball when compared to males (23).

Risk factors involved with noncontact ACL injuries have been identified at both the knee and the hip. At the knee, an initial contact flexion angle of less than 30°, high abduction angles, and the center of gravity behind the knee (11, 15). During landing, peak ACL loads occur just after initial contact, when the knee is near full extension (16). The angle between the patellar tendon and the tibial shaft decreases as knee flexion increases which results in decreased amount of force that is directed anteriorly relative to the tibia (54, 89). When the knee is near full extension the knee flexor musculature cannot provide a large amount of posterior force to the tibia thus increasing anterior tibial shear forces. Decreased hip flexion during landing has also been identified as an ACL injury risk factor due to the coupled nature of the hip and knee (7, 11, 18).

Injury risk factors have also been identified in the lower extremity muscle activity. Specifically, altered muscle activation patterns and muscular imbalances have been associated with increased ACL injury risk. Quadriceps dominance, a modifiable neuromuscular imbalance, is defined as the imbalance between knee extensor and flexor strength, recruitment, and coordination (19). The high activation of the quadriceps during an athletic task such as a landing or cutting task can increase anterior tibial shear force, which has been shown to load the ACL through the patellar tendon (5, 20). The activation of the hamstrings can help to provide a counterbalancing force to protect against the relatively higher quadriceps force, thus reducing the anterior tibial translation and potentially decreasing ACL strain (21). The quadriceps-to-hamstrings co-contraction index (Q:H CCI) is one way to quantify the difference in the activations of both muscle groups, and has been found to influence knee flexion angle at initial contact, with higher ratios resulting in decreased initial contact knee flexion angles (22).

A primary concern in biomechanics research is the lack of a realistic, game-like environment during data collection. One step being taken to make laboratory testing more realistic is by including external objects in the testing environment. Including the unpredictability of an unanticipated task is another way to make the testing environment more realistic. This unpredictability requires the athletes to react to a stimulus, similar to what would be experienced during game play while reacting to a ball. Besier et al. (90) concluded that unanticipated side-cutting movements can cause increases in the frontal plane internal knee adduction moments by up to twice the magnitude of the same pre-planned movement.

The purpose of this study was to determine the effects of a physical overhead goal and a subsequent unanticipated lateral cut on the kinematics, kinetics, and muscle activation of the hip and knee of the dominant leg during the landing of a volleyball block. It was hypothesized that performing the volleyball block landing task with a physical overhead target condition would decrease the hip and knee flexion angles, increase knee abduction angles, and increase the peak hip extension, knee extension and knee adduction moments. It was further hypothesized that performing a subsequent unanticipated task immediately after the volleyball block landing would further decrease hip and knee flexion angles, increase knee abduction angles, and increase peak hip extension moments, knee extension moments and knee adduction moments. Lastly, it was hypothesized that the Q:H CCI would increase during the unanticipated tasks when compared to the anticipated tasks.

Methods

Participants

Thirteen recreationally active females (age: 21.3 ± 2.7 yrs, height: 1.75 ± 0.06 m, mass: 67.80 ± 8.97 kg) were recruited to participate in the study. Each participant needed to have at least two-years of experience playing volleyball at the high school level or higher to be eligible to participate. In addition, they needed to be recreationally active defined as three or more days per week of 30 minutes or more of moderate to vigorous physical activity, with one day having to consist of a dynamic activity such as running with jumping or cutting. Participants were excluded from the study if they had a lower extremity injury that required surgery, ever had an ACL injury, or had experienced a lower extremity injury within the six months prior to testing. Participants provided written consent upon arrival to the lab, with the informed consent being approved by the University Institutional Review Board prior to testing. Participants then completed the Lower Extremity Functional Scale (LEFS) to determine if they have any difficulty in performing different daily activities. Binkley et al (79) determined that the minimal detectable change at a 90% confidence level was ± 9. Therefore, participants who scored 71 and below on the LEFS scale were excluded because they exhibited a clinically meaningful functional change.

Experimental Procedures

A twelve-camera motion capture system (200 Hz, Vicon Motion Analysis, Inc., Oxford, UK) was used to collect three-dimensional marker coordinate data for each subject. Two 60x60 cm force platforms (2000 Hz, BP600600, Advanced Mechanical Technology, Inc., Watertown, MA, USA) were used to measure ground reaction forces (GRF). Dominant leg Vastus Medialis (VM), Vastus Lateralis (VL), Biceps Femoris (BF), and medial hamstrings (MH) were recorded with a TrignoTM Wireless EMG system (2000 Hz, Delsys, Inc., Natick, MA, USA). Leg dominance was determined by asking the participant which leg they prefer to kick a ball with. One sensor was applied to each muscle following the guidelines from Rainoldi et al. (80). Muscle activation data was collected at the same time as the marker coordinate data and ground reaction force data using Vicon Nexus 2.7.1 (Vicon Motion Analysis, Inc., Oxford, UK). A single timing gate (63501 IR, Lafayette Instrument Inc., IN, USA) was used to trigger the visual stimulus for the unanticipated conditions.

Prior to testing, maximal voluntary isometric contraction (MVIC) tests were done for both the quadriceps and hamstrings with an isokinetic dynamometer (Biodex Medical Systems, Inc., Shirley, NY, USA). Quadriceps MVIC trials were conducted in an upright seated position, with the hip and knee flexed to 90°. Hamstrings MVIC trials were performed in a prone position with the knee flexed at 30°. Trials began with a three-second ramp-up to 100% of their maximal effort. Maximal contractions were then held for 5 seconds, and three trials were performed on both the quadriceps and hamstrings. EMG MVIC data was pre-amplified and high-pass filtered using a fourth-order, zero lag Butterworth filter with a cutoff frequency of 10 Hz. The signal was also full-wave rectified then low-pass filtered with a cutoff frequency of 10 Hz (81). A one second moving average was then performed on the filtered data and the maximal value was recorded for each trial and the maximum value of the three trials was the value that was used as the MVIC value to normalize movement trial EMG.

Reflective markers were then applied to the participant to define lower extremities and pelvis segments. The pelvis was defined by the left and right iliac crest markers and left and right greater trochanter markers. Anatomical markers were also placed at the medial and lateral femoral epicondyles on each leg to define the thigh, on the medial and lateral malleoli to define the shank, and on the first and fifth metatarsal to define the foot. Semi-rigid thermoplastic shells with four reflective markers were then attached to Velcro-sensitive neoprene wraps on the pelvis, thighs, and shanks. A semi-rigid thermoplastic shell with four reflective markers was also secured to the lateral heel of each foot using tape to ensure no movement on the shoe. Each of the marker clusters were used to track segment movement during the dynamic movement trials. A static trial was conducted, followed by a dynamic range of motion (ROM) trial. Once the dynamic ROM trial was collected, the movement trial data collection began.

Testing Protocol

Prior to testing, participants performed three maximal jump and reach trials using a Vertec Vertical Jump Tester (Sports Imports Inc., Columbus, OH, USA). Once the average of the three trials was calculated, 90% of that jump height was determined and the middle of a suspended volleyball was placed at that height using a rebounder machine. Participants were excluded if they had a jump and reach height that was less than 2.54 m as the ball would not be able to clear the top of the net (2.24 m) when set at 90% of the participants maximum jump and reach height.

The beginning to all trials was the same during the testing session. The participant would begin by standing on the side of the force platforms that they felt most comfortable approaching the jump from. They would then step toward the force platforms first with the foot furthest from the platforms towards followed by the other foot. Once the participant was on the force platforms, they would perform a vertical jump with their arms raised over their heads and then anteriorly extend their arms as if to reach over the net to block the ball. Four different testing conditions were performed that consisted of 1) jumping with no target and anticipating the secondary task (NTA), 2) jumping with no target and an unanticipated secondary task (NTUA), 3) jumping with a ball and net and an anticipated secondary task (WTA), and 4) jumping with a ball and net and an unanticipated secondary task (WTUA). The with-target conditions included a net at the standard 2.43 m that is specified for female volleyball competitions and a ball that was suspended by a rebounder machine with the middle of the ball at 90% of the participants maximum jump and reach height. Because of the set-up required for the with-target conditions, testing condition order was randomized where with-target conditions were completed consecutively as well as the no target conditions being completed consecutively.

For the anticipated conditions, participants were told to jump and block the ball, land and cut off the dominant leg (i.e. right leg dominant cuts off the right leg to the left) away from the net at a 45° angle. For the unanticipated task conditions, a visual cue was given when the participant's hands broke the infrared beam of the timing gate which was set just in front of (the same side as the participant), and above the top of the net. When the infrared beam was broken, a custom Lab View program presented either a left arrow (left cut at a 45° angle), a right arrow (right cut at a 45° angle), an up arrow (second vertical jump), or a stop sign (stick landing) onto a computer monitor. The unanticipated task trials were deemed valid if the participant immediately moved in the proper direction without stopping after the landing. Prior to data collection, participants were given the opportunity to practice the stop jump until they were comfortable with the tasks.

Data Processing and Statistical Analysis

Motion trials were analyzed using the Visual3D software suite (v6, C-Motion, Inc., Rockville, MD) which computes kinematic and kinetic data for both legs. Hip joint centers were located at 25% of the distance from the ipsilateral to contralateral greater trochanter marker (86). Knee joint centers were located at the midpoints between the epicondyle markers (87), and ankle joint centers were placed at the midpoints between the malleoli markers (88). Marker coordinate and ground reaction force data were filtered using a fourth-order, zero-lag Butterworth filter with a cutoff frequency of 10 Hz. Internally applied, three-dimensional joint kinetics were calculated using a Newton-Euler approach (82), and were transformed to the joint coordinate system (83). All kinetic data was normalized to body mass. Initial contact during the landing was determined by finding the first frame when the force was greater than 10 N.

Motion trial EMG data was pre-amplified and high-pass filtered using a fourth-order, zero lag Butterworth filter with a cutoff frequency of 10 Hz. The signal was then full-wave rectified. The full-wave rectified signal was then low-pass filtered with a cutoff frequency of 20 Hz. EMG from each trial was expressed as a percentage of the maximum EMG from the MVIC trials (%MVIC).

Quadriceps-to-hamstrings co-contraction index (Q:H CCI) were computed both pre-contact (1) and post contact (2) for each trial. Q:H CCI were computed as the ratio of average quadriceps activation to average hamstring activation.

$$Pre\text{-}contact\ Q{:}H\ CCI = \frac{(Q_{50ms\ pre-HS})}{(H_{50ms\ pre-HS})} \tag{1}$$

$$Post\text{-}contact\ Q{:}H\ CCI = \frac{(Q_{HS-100ms\ post})}{(H_{HS-100ms\ post})} \tag{2}$$

The average quadriceps activation was calculated as the sum of the average VM and average VL activations (3), while the average hamstrings activation was calculated as the sum of the average BF and MH activations (4).

$$Q = (avgVM + avgVL) \tag{3}$$

$$H = (avgBF + avgMH) \tag{4}$$

The dependent variables evaluated in this study included: initial contact and peak hip kinematics in the sagittal plane, peak internal moments at the hip in the sagittal plane, initial contact and peak knee kinematics in the sagittal and frontal planes, peak internal moments at the knee in the sagittal and frontal planes, and pre-contact and post-contact Q:H CCI. Kinematics and kinetics were evaluated during the deceleration phase of the landing, while EMG data were evaluated from 50 ms pre-contact to 100 ms post-contact. For each participant, all dependent variables were represented as the mean ± standard deviation of the five trials for each condition.

Statistical Analysis

Data was analyzed using SPSS (v25.0, SPSS Inc., Chicago, IL, USA) to examine the differences between the lower extremity biomechanics for each condition. 2 (WT vs NT) x 2 (A vs UA) repeated-measures analyses of variance (ANOVAs) were performed. All tests were performed with a significance value of $\alpha = 0.05$. Partial eta squared values were reported if significant main effects or interactions were found. If a significant target×anticipation interaction was identified, paired samples t-tests were performed, and Cohen's d (1992 Cohen paper) effect sizes were calculated.

Results

All participants in this study were right leg dominant. Jump heights were very similar between conditions (WTA: 0.41 ± 0.05 m, WTUA: 0.41 ± 0.06 m, NTA: 0.41 ± 0.06 m, NTUA: 0.42 ± 0.06 m), with individual jump heights listed in Table 6.

Electromyography

There was a significant target×anticipation interaction for pre-contact Q:H CCI (p = 0.027, η_p^2 = 0.345) (Table 3). Post hoc pairwise comparison t-tests determined that there were no significant differences between target conditions (NTA significantly different than NTUA (p = 0.187, d = 0.30), WTA significantly different than WTUA (p = 0.776, d = 0.06)) and no significant differences between anticipation conditions (NTA significantly different than WTA (p = 0.779, d = 0.05), NTUA significantly different than WTUA (p = 0.231, d = 0.38)). Neither the target×anticipation interaction, nor the main effects of target or anticipation were significant for post-contact Q:H CCI.

Kinematics and Kinetics

With regards to IC kinematics, there was a significant target main effect for the IC knee flexion angle (p = 0.013, η_p^2 = 0.413) (Table 4), and a significant anticipation main effect for the IC knee abduction angle (p = 0.001, η_p^2 = 0.609). IC knee flexion angles during with-target conditions demonstrated a 1.7° decrease when compared to the NT conditions, while the IC knee abduction was 1.2° greater during the anticipated condition when compared to the unanticipated condition. A significant target×anticipation interaction in the IC hip flexion angle (p = 0.006, η_p^2 = 0.345) was also observed. Post hoc tests determined there was an anticipation effect during the with-target condition specifically (p = 0.017, d = 0.45). The WTA condition had on average a 3.4° increase in IC hip flexion angle compared to the WTUA condition.

For peak kinematics there was a significant target×anticipation interaction for the peak knee abduction angle (p = 0.043, η_p^2 = 0.300) (Table 5). Post hoc results indicated there was an anticipation effect when the target was present (p = 0.007, d = 0.54). Participants demonstrated on average 2.3° greater peak knee abduction angles during the WTA task compared to the WTUA task. There was also a significant anticipation effect for the peak knee flexion angle (p = 0.034, η_p^2 = 0.324). The unanticipated condition on average had 2.0° greater knee flexion compared to the anticipated condition.

Finally, there was a significant anticipation main effect for peak hip extension moment (p = 0.017, η_p^2 = 0.389) and a significant target main effect for peak knee extension moment (p = 0.001, η_p^2 = 0.617). Peak hip extension moment in the anticipated conditions was on average 0.25 Nm/kg greater when compared to the unanticipated conditions, while the with-target conditions had on average a 0.16 Nm/kg greater knee extension moment when compared to the NT conditions (Table 5).

Discussion

This study was meant to allow the participants to be as natural as they could and make the environment more realistic. By including the net and ball to the environment, we were able to observe differences in the IC hip flexion angle and the peak knee extension moments. By including a secondary unanticipated task, we also observed differences in the IC knee abduction angle, peak knee flexion and abduction angle, and peak hip extension moments. While some of the values for these variables may not have been what was hypothesized, they are still useful in helping to determine how the playing environment can affect potential ACL injuries.

Our first hypothesis of including the physical goals in the environment would negatively affect ACL injury risk factors was partially supported. When including the overhead target into the testing environment, we found significant decreases in the IC knee flexion angle, but no differences in the IC hip flexion angle or peak hip and knee flexion angles. Other studies have suggested that including the overhead goal would potentially increase knee range of motion in females (91), however we only found significant changes in the IC knee flexion angle between the two target conditions. Decreases in the knee flexion angles during landing has been shown to be a risk factor for ACL injury (11, 15) and this data shows that adding a target did negatively influence this specific risk factor.

Adding the target to the environment also increased the peak knee extension moment. This is in agreement with Mok et al. (31), who reported an increased magnitudes of hip and knee moments when a target was being used (similar differences but moments were not normalized). With these findings, we can see that when the participants were in the testing environment that was more realistic with the targets, there were some negative changes in ACL injury risk factors.

Our second hypothesis of the unanticipated task negatively influencing ACL injury risk factors was not supported. By adding the unanticipated task to make the blocking task more realistic, we observed decreases in both IC hip flexion and knee abduction angles, increased peak knee flexion, decreased abduction angles, and decreased peak hip extension moments during unanticipated movements. Results indicated that unanticipated trials had an average of 3.4° less hip flexion at initial contact compared to the anticipated trials when a target was present. This finding contradicts those of Meinerz et al. (77), who found a trend of increasing hip flexion angle at initial contact while doing a drop landing task followed by a side cut (anticipated: 35.6 ± 12.0°, unanticipated: 36.7 ± 10.9°). When comparing our initial contact angles to other studies that have included an unanticipated task (78), similar trends were apparent in the decrease of the IC hip flexion angle from the anticipated to unanticipated conditions. The decreased hip flexion angles that were observed may impact the ability of the hip extensor muscles to effectively absorb the energy of the body causing more of the force to be applied to the passive tissues (i.e. ligaments and bones) (47, 53). Decreased hip flexion angles have been associated with ACL injury due to the coupled nature of the knee and hip (18).

While IC knee flexion angles were not significantly different between anticipated tasks in this study, it appears that the decreased IC hip flexion angle was accompanied by decreased IC knee flexion angles during the unanticipated trials. Participants also exhibited 2° greater peak knee flexion in the unanticipated trials compared to the anticipated trials. This increase in knee flexion angle was also observed by Sell et al. (46) who had an average of 3.1° increase during a cutting task, whereas Meinerz et al. (77) found no difference in the peak knee flexion angles (anticipated: 47.9 ± 7.3°, unanticipated: 48.3 ± 7.1°). Regarding peak knee abduction angles, differences were observed only when the target was present. Females were on average 2.3° less abducted during the unanticipated condition compared to the anticipated condition.

The peak hip extension moment had differing trends between anticipation condition to those of Meinerz et al. (77). In the current study there was a prominent decrease in the magnitude of the hip extension moment from anticipated to unanticipated conditions. While not many studies have specifically evaluated the differences in moments at the hip while doing unanticipated tasks, there is much more data on the moments of the knee. Increases in sagittal knee joint loading has been shown by Sell et al. (46) during an unanticipated landing task, while Meinerz et al. (77) found a significant decrease in the knee extension moment during unanticipated conditions.

Many of the frontal plane measures from this study differed from the results of previous research. Sell et al. (46) reported a significant increase in the peak knee adduction moment when adding an unanticipated task. Whereas in the current study we found no differences in the peak knee adduction moments between the anticipated tasks. Additionally, the frontal plane knee kinematics in the current study are actually reducing the known risk-factors when performing the unanticipated trials compared to the anticipated trials. While most of the loading that occurs to the ACL comes through the sagittal plane and anterior tibial translation, there has still been a connection between larger knee adduction moments an increased risk for rupturing the ACL (29).

Since there were no significant differences in the pre-contact or post-contact Q:H CCI, we can reject our third hypothesis that the unanticipated task would increase the Q:H CCI. While we found no significant differences, data on Table 3 shows that there was greater pre-contact activation in the anticipated conditions and greater post-contact activation in the unanticipated conditions. While we did not examine the relationship between the pre- and post-contact activation, we can see the idea of quadriceps dominance in both the pre- and post-contact CCIs. Because of the high level of activity of the quadriceps during this time around the landing in all of the conditions, there could still be an increased amount of anterior tibial translation during the deceleration phase of the landing that is placing the ACL at risk of injury during this specific block landing task.

There were a few limitations that should be noted for the current study. First, participants were only recreationally active may not have had the same level of training or may have had a longer time of not participating in the sport prior to the data collections. The differences in skill level may have influenced some of the different landing mechanics between participants. Another limitation is that some of the participants may not have had the same position specific training as others performing the same tasks. Since there was no position requirement with this study, athletes who may not have played blocker as much in their career were able to participate and their data may differ from a participant who has played blocker their entire career. Finally, fatigue may have been a factor in some of the participants data collections. Because the visual stimulus was completely random, the total number of trials completed was not consistent between participants. During testing, we would try to prevent fatigue from becoming a factor by allowing the participant to take however much time they needed between trials (at least 30 seconds), as well as water breaks whenever they desired during the testing session. In unanticipated conditions that went above the 20-trial mark, we would check with the participant to make sure they were still comfortable with the task and if they wished to continue until 5 valid trials cutting off the dominant leg were collected.

Conclusion

Our findings suggest that the more realistic testing environment altered hip and knee landing mechanics. In the case of a volleyball block, including a net and a ball in the testing environment allows for a more accurate evaluation of the risk-factors for ACL injury in athletes. While differences were observed in both target and anticipation testing conditions it is important to note that a majority of the differences were seen between the different anticipation conditions meaning that the quick reactions similar to decisions made during game play may have more of an impact on landing mechanics than just the inclusion of a physical goal into the testing environment.

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APPENDICES

APPENDIX A. SUBJECT DATA

Table 1. Participant Demographics: mean ± STD.

Height (m)	**Weight (kg)**	**Age (yrs)**	**Total Playing Experience (yrs)**	**High School/College Playing Experience (yrs)**
1.75 ± 0.06	67.80 ± 8.97	22.9 ± 2.7	7.7 ± 2.3	6.3 ± 1.9

Table 2. Individual Participant Demographics

Participant	Height (m)	Mass (kg)	Age (yrs)	LEFS	Total Playing Experience (yrs)	High School/College Playing Experience (yrs)
1	1.84	71.71	21	80	10	8
2	1.65	58.29	21	78	9	5
3	1.78	66.04	19	80	6	6
4	1.86	78.64	21	77	10	8
5	1.69	66.49	20	80	8	7
6	1.76	75.96	20	80	7	4
7	1.67	49.11	19	80	6	4
8	1.73	75.79	26	77	8	3
9	1.73	66.77	27	80	2	2
10	1.81	81.50	23	80	8	4
11	1.72	60.31	19	80	8	4
12	1.73	64.58	18	77	7	4
13	1.77	66.25	21	80	11	4

APPENDIX B. CHAPTER 4 TABLES

Table 3. EMG 2x2 ANOVA Main Effects

		Anticipated	**Unanticipated**	**Target Effect *p*-value**	**Anticipation Effect *p*-value**	**Target*Anticipation *p*-value**
Pre-Contact Q:H CCI †	**Target**	2.046 ± 2.421	2.167 ± 1.482	0.580	0.590	0.027
	No Target	2.155 ± 2.134	1.617 ± 1.396			
Post-Contact Q:H CCI	**Target**	1.996 ± 2.141	3.017 ± 2.487	0.082	0.098	0.273
	No Target	1.862 ± 1.447	2.079 ± 2.217			

†: significant target×anticipation interaction

Table 4. Kinematics 2x2 ANOVA Main Effects

		Anticipated	**Unanticipated**	**Target Effect *p*-value**	**Anticipation Effect *p*-value**	**Target*Anticipation *p*-value**
IC Hip Flexion Angle (°)† [(a)]	**Target**	23.7 ± 9.2	20.3 ± 5.7	0.986	0.145	0.006
	No Target	22.1 ± 7.2	21.9 ± 6.0			
IC Knee Flexion Angle (°)*	**Target**	-12.0 ± 5.7	-10.5 ± 4.1	0.013	0.089	0.908
	No Target	-13.8 ± 6.2	-12.1 ± 5.3			
IC Knee Abd. Angle (°) ^	**Target**	-3.2 ± 3.1	-1.9 ± 3.5	0.909	0.001	0.696
	No Target	-3.1 ± 2.9	-2.0 ± 3.1			
Peak Hip Flexion Angle (°)	**Target**	53.9 ± 21.9	53.9 ± 11.3	0.100	0.139	0.115
	No Target	54.8 ± 14.6	58.6 ± 13.9			
Peak Knee Flexion Angle (°) ^	**Target**	-63.8 ± 8.1	-65.5 ± 9.5	0.323	0.034	0.697
	No Target	-64.3 ± 9.3	-66.6 ± 9.1			
Peak Knee Abd. Angle (°) ^†[(a)]	**Target**	-11.4 ± 4.5	-9.1 ± 4.1	0.908	0.017	0.043
	No Target	-10.7 ± 4.3	-9.7 ± 4.6			

*: significant target main effect
^: significant anticipation main effect
†: significant target×anticipation interaction (a= WTA is significantly different than WTUA)

Table 5 . Peak Kinetics 2x2 ANOVA Main Effects

		Anticipated	**Unanticipated**	**Target Effect *p*-value**	**Anticipation Effect *p*-value**	**Target*Anticipation *p*-value**
Peak Hip Extension Moment (Nm/kg) ^	**Target**	-2.18 ± 0.49	-1.80 ± 0.58	0.064	0.017	0.199
	No Target	-2.27 ± 0.49	-2.16 ± 0.49			
Peak Knee Extension Moment (Nm/kg) *	**Target**	2.08 ± 0.51	2.16 ± 0.51	0.001	0.707	0.209
	No Target	1.97 ± 0.52	1.95 ± 0.54			
Peak Knee Add. Moment (Nm/kg)	**Target**	0.40 ± 0.24	0.36 ± 0.20	0.602	0.821	0.051
	No Target	0.39 ± 0.24	0.41 ± 0.22			

*: significant target main effect
^: significant anticipation main effect

Table 6. Individual Jump Heights (m)

Participant	NTA	NTUA	WTA	WTUA
1	0.56 ± 0.01	0.55 ± 0.01	0.52 ± 0.02	0.56 ± 0.00
2	0.47 ± 0.01	0.47 ± 0.02	0.45 ± 0.02	0.44 ± 0.01
3	0.39 ± 0.01	0.38 ± 0.02	0.35 ± 0.01	0.37 ± 0.01
4	0.51 ± 0.01	0.49 ± 0.00	0.50 ± 0.01	0.48 ± 0.02
5	0.40 ± 0.00	0.44 ± 0.01	0.42 ± 0.00	0.42 ± 0.02
6	0.41 ± 0.01	0.42 ± 0.01	0.39 ± 0.01	0.39 ± 0.01
7	0.41 ± 0.01	0.44 ± 0.01	0.43 ± 0.01	0.43 ± 0.01
8	0.38 ± 0.01	0.39 ± 0.01	0.39 ± 0.01	0.39 ± 0.01
9	0.38 ± 0.02	0.39 ± 0.02	0.43 ± 0.01	0.43 ± 0.01
10	0.34 ± 0.02	0.35 ± 0.01	0.35 ± 0.01	0.35 ± 0.01
11	0.33 ± 0.02	0.32 ± 0.01	0.32 ± 0.01	0.32 ± 0.01
12	0.39 ± 0.01	0.40 ± 0.01	0.39 ± 0.01	0.38 ± 0.01
13	0.38 ± 0.02	0.42 ± 0.03	0.40 ± 0.01	0.40 ± 0.01

APPENDIX C. CHAPTER 4 FIGURES

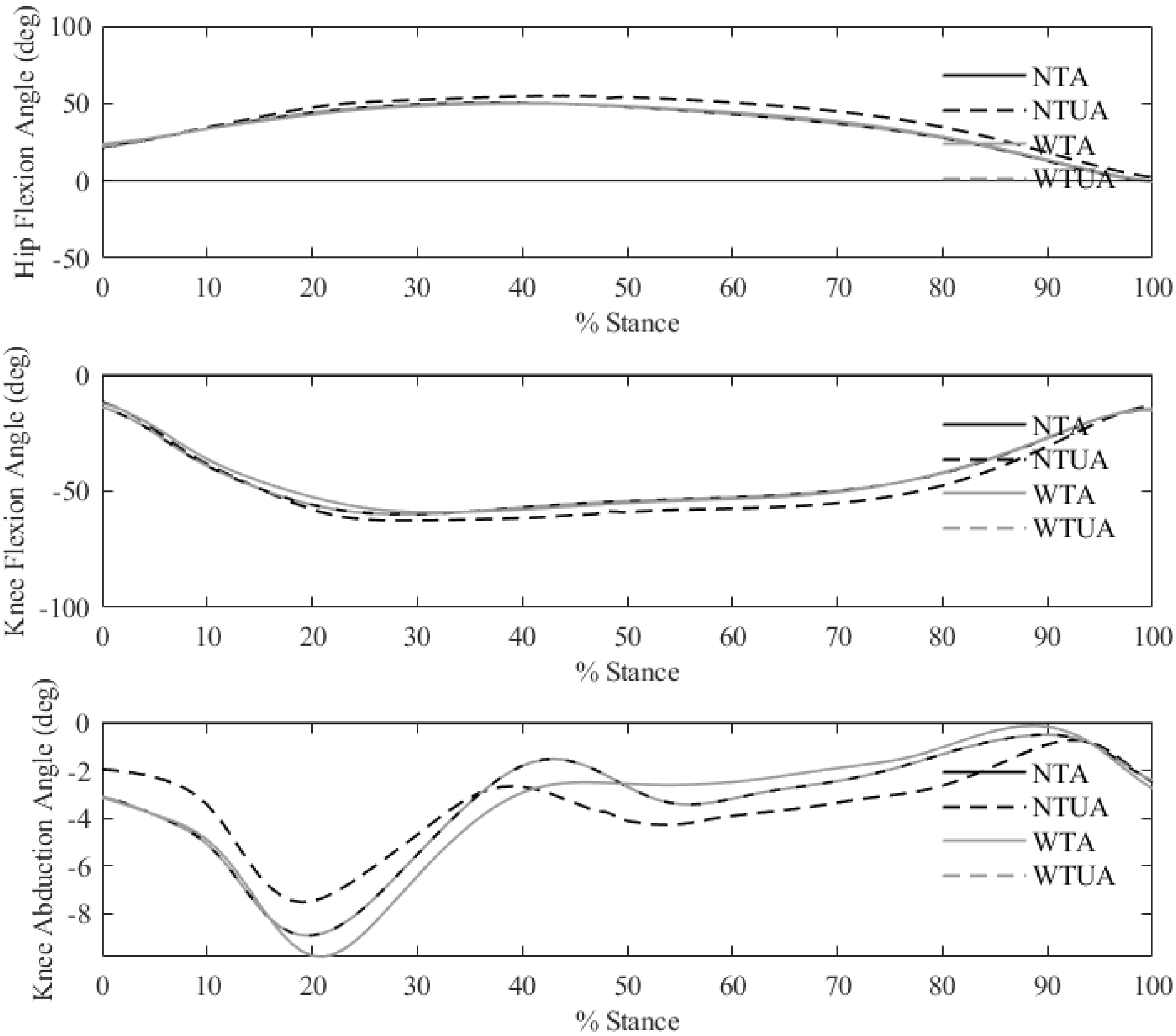

Figure 1. Kinematic Ensemble curves. Comparison of the knee and hip kinematics between the four different testing conditions from initial contact to final push off from the force platform.

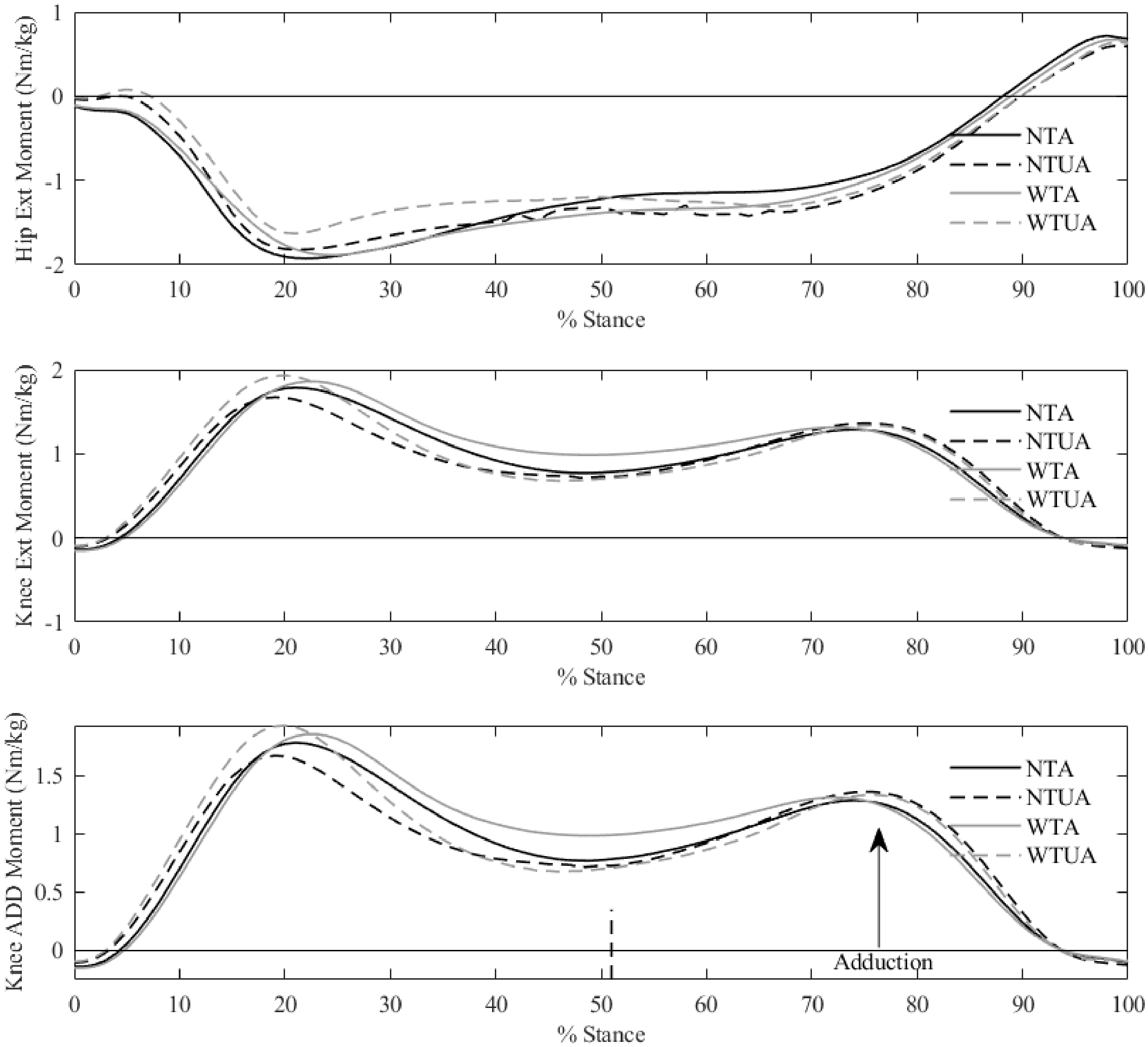

Figure 2. Kinetic ensemble curves. Comparison of the knee and hip kinematics between the four different testing conditions from initial contact to final push off from the force platform.

APPENDIX D: INFORMED CONSENT

Informed Consent

The Effects of an Overhead Goal and Movement Anticipation on Lower Extremity Biomechanics and Muscle Activation Patterns

INTRODUCTION

You are invited to participate in a research study conducted in the University of Tennessee Biomechanics Lab (HPER 136). The purpose of this study is to determine the effects of a physical overhead goal and a subsequent unanticipated lateral cut on leg biomechanics and muscle activations in healthy, recreationally active women.

ELIGIBILITY

To participate in this study, you must be between the ages of 18 and 30 and be currently recreationally active. We define recreationally active as being physically active at least 3 days per week for a minimum of 30 minutes each session. You must have also had at least two years of volleyball experience playing at the high school level or higher. You must NOT have: undergone surgery for a lower extremity injury (e.g., ligament rupture, meniscus repair, bone fracture), suffered an anterior cruciate ligament (ACL) injury, or suffered a lower extremity injury in the past six months.

INFORMATION ABOUT PARTICIPANTS' INVOLVEMENT IN THE STUDY

You will come into the Biomechanics Lab for one session, which will last approximately 2 hours.

You will come into the Biomechanics Lab, where you will read the informed consent document and given time to ask any questions pertaining to the study. You will also complete the Lower Extremity Functional Scale to ensure that you are qualified to participate. A score of 71/80 or lower on the scale will exclude you from the study. You then will change into Spandex shorts and a generic t-shirt, which will be provided. Height and weight will be taken, followed by a 5-minute warmup walk/jog and stretching of the lower extremities.

Anatomical and tracking markers will be placed on you, which will be used to track movement data throughout the session. You will be asked to perform a stop jump (blocking motion) followed by an immediate cut off the dominant foot for anticipated conditions. The same stop jump movement will take place in the unanticipated tasks with the inclusion of cutting off the non-dominant foot and a landing with no movement after. Five successful trials for each condition will then be completed by the participant. The session will conclude once these trials are collected.

RISKS

While this study involves jump landings, you will be performing a lateral cut immediately after contact with the ground decreasing the amount of time to react safely. Therefore, there is still some risk of injury. Although the likelihood of these risks are low, they are still known risks associated with the procedure.

BENEFITS

There will be no direct benefits to you. The data collected from you will help provide a better understanding of how a more realistic environment influences knee biomechanics and leg muscle activation and how this affects known risk factors of ACL injury. The data collected may also provide information on how to alter walking to reduce potential injury risk.

CONFIDENTIALITY

The information collected in this study will be kept confidential. You will be identified by a given number. Data will be stored securely, both in a password-protected computer desktop and in a locked drawer in the Biomechanics lab. Information will be available only to persons conducting the study unless participants specifically give permission in writing to do otherwise. No reference will be made in oral or written reports which could link you to the study.

CONTACT INFORMATION

If you have questions at any time about the study or the procedures, (or you experience adverse effects because of participating in this study,) you may contact the researcher, Michael O'Dwyer, at mo39dwye@vols.utk.edu, or Dr. Joshua Weinhandl at jweinhan@utk.edu. If you have questions about your rights as a participant, you may contact the University of Tennessee IRB Compliance Officer at utkirb@utk.edu or (865) 974-7697.

PARTICIPATION

Your participation in this study is voluntary; you may decline to participate without penalty. If you decide to participate, you may withdraw from the study at any time without penalty and without loss of benefits to which you are otherwise entitled. If you withdraw from the study before data collection is completed your data will be permanently deleted from our system, and any forms with identification information will be shredded.

CONSENT

I have read the above information. I have received a copy of this form. I agree to participate in this study.

Participant's Name (printed) ____

Participant's Signature ________________________________ Date __________

IRB NUMBER: UTK IRB-18-04887-XP
IRB APPROVAL DATE: 03/04/2019
IRB EXPIRATION DATE: 12/09/2019

APPENDIX E: LOWER EXTREMITY FUCTIONAL SCALE

Lower Extremity Functional Scale

We are interested in knowing whether or not you are having any difficulty at all with the activities listed below. Please provide an honest answer for each activity.

KEY

0 - Extreme difficulty or unable to perform activity
1 - Quite a bit of difficulty
2 - Moderate difficulty
3 - A little bit of difficulty
4 - No difficulty

	Extreme	Quite a bit	Moderate	Minimal	None
Today, ***do you*** or ***would you*** have any difficulty at all with:	0	1	2	3	4
1. Any of your usual work, housework or school activities	☐	☐	☐	☐	☐
2. Your usual hobbies, recreational or sporting activities	☐	☐	☐	☐	☐
3. Getting into or out of the bath	☐	☐	☐	☐	☐
4. Walking between rooms	☐	☐	☐	☐	☐
5. Putting on your shoes or socks	☐	☐	☐	☐	☐
6. Squatting	☐	☐	☐	☐	☐
7. Lifting an object, like a bag of groceries from the floor	☐	☐	☐	☐	☐
8. Performing light activities around your home	☐	☐	☐	☐	☐
9. Performing heavy activities around your home	☐	☐	☐	☐	☐
10. Getting into or out of a car	☐	☐	☐	☐	☐
11. Walking 2 blocks	☐	☐	☐	☐	☐
12. Walking a mile	☐	☐	☐	☐	☐
13. Going up or down 10 stairs (about 1 flight)	☐	☐	☐	☐	☐
14. Standing for 1 hour	☐	☐	☐	☐	☐
15. Sitting for 1 hour	☐	☐	☐	☐	☐
16. Running on even ground	☐	☐	☐	☐	☐
17. Running on uneven ground	☐	☐	☐	☐	☐
18. Making sharp turns while running fast	☐	☐	☐	☐	☐

19. Hopping	☐	☐	☐	☐	☐
20. Rolling over in bed	☐	☐	☐	☐	☐

APPENDIX F: IRB APPROVAL LETTER

THE UNIVERSITY OF
TENNESSEE
KNOXVILLE

March 04, 2019

Michael Joseph O'Dwyer,
UTK - Coll of Education, Hlth, & Human - Kinesiology Recreation & Sport Studies

Re: UTK IRB-18-04887-XP
Study Title: THE EFFECTS OF AN OVERHEAD GOAL AND MOVEMENT ANTICIPATION ON LOWER EXTREMITY BIOMECHANICS AND MUSCLE ACTIVATION PATTERNS

Dear Michael Joseph O'Dwyer:

The UTK Institutional Review Board (IRB) reviewed your application for **revision** of your previously approved project, referenced above.

The IRB determined that your application is eligible for **expedited** review under 45 CFR 46.110(b)(2). The following revisions were approved as complying with proper consideration of the rights and welfare of human subjects and the regulatory requirements for the protection of human subjects:

Approved Revisions
-Change eligibility criteria from ages 18-25 to 18-30

Approved Documents
-UTK Knoxville Main Campus IRB Application – Version 1.2
-The Effects of an Overhead Goal and Movement Anticipation on Lower Extremity Biomechanics and Muscle Activation Patterns - Version 1.2
-MO recruitment email 02 - Version 1.2

Approval does not alter the expiration date of this project, which is 12/09/2019.

In the event that subjects are to be recruited using solicitation materials, such as brochures, posters, web-based advertisements, etc., these materials must receive prior approval of the IRB. Any revisions in the approved application must also be submitted to and approved by the IRB prior to implementation. In addition, you are responsible for reporting any unanticipated serious adverse events or other problems

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involving risks to subjects or others in the manner required by the local IRB policy.

Finally, **re-approval** of your project is required by the IRB in accord with the conditions specified above. You may not continue the research study beyond the time or other limits specified unless you obtain prior written approval of the IRB.

Sincerely,

Colleen P. Gilrane

Colleen P. Gilrane, Ph.D.
Chair

APPENDIX G: INDIVIDUAL RESULTS FOR SELECT VARIABLES

Table 7. Individual Pre-contact Q:H CCI

Participant	NTA	NTUA	WTA	WTUA
1	0.556 ± 0.446	0.692 ± 0.251	1.137 ± 0.971	2.844 ± 1.834
2	5.060 ± 3.758	3.560 ± 3.782	2.296 ± 3.207	1.462 ± 1.286
3	0.329 ± 0.168	0.764 ± 0.722	1.056 ± 1.369	0.740 ± 0.791
4	0.381 ± 0.123	0.254 ± 0.117	0.227 ± 0.110	1.808 ± 1.027
5	0.211 ± 0.125	0.659 ± 0.661	0.945 ± 0.666	0.609 ± 0.253
6	0.217 ± 0.060	0.196 ± 0.114	0.118 ± 0.033	0.563 ± 0.370
7	3.806 ± 2.581	3.971 ± 2.432	1.508 ± 0.781	2.315 ± 1.224
8	2.217 ± 1.091	1.494 ± 1.069	2.216 ± 1.289	3.768 ± 1.238
9	6.104 ± 3.145	1.408 ± 1.095	8.876 ± 1.289	4.965 ± 2.598
10	1.068 ± 1.821	0.506 ± 0.237	0.277 ± 0.096	1.096 ± 1.409
11	4.081 ± 2.558	3.429 ± 1.640	3.995 ± 0.065	3.057 ± 1.073
12	3.778 ± 0.928	3.213 ± 1.741	3.750 ± 2.442	4.157 ± 1.581
13	0.207 ± 0.064	0.873 ± 0.673	0.196 ± 0.067	0.781 ± 0.639

Table 8. Individual Post-contact Q:H CCI

Participant	NTA	NTUA	WTA	WTUA
1	1.564 ± 0.339	1.506 ± 0.353	1.389 ± 0.511	2.400 ± 0.772
2	2.944 ± 2.617	1.821 ± 0.636	1.884 ± 0.395	5.737 ± 3.867
3	0.329 ± 0.127	0.772 ± 0.273	0.273 ± 0.050	0.904 ± 0.242
4	0.630 ± 0.136	0.497 ± 0.221	0.445 ± 0.189	0.695 ± 0.809
5	0.889 ± 0.226	0.920 ± 0.518	0.691 ± 0.394	1.133 ± 1.057
6	1.087 ± 0.163	1.453 ± 0.939	1.230 ± 0.695	3.836 ± 0.679
7	2.280 ± 1.199	8.320 ± 3.410	7.871 ± 1.844	9.202 ± 2.357
8	1.212 ± 0.259	1.941 ± 0.952	1.579 ± 0.296	1.570 ± 0.060
9	5.239 ± 3.086	1.408 ± 1.095	3.635 ± 2.193	4.856 ± 3.940
10	0.572 ± 0.524	0.490 ± 0.261	0.292 ± 0.151	1.155 ± 1.461
11	2.276 ± 1.711	4.414 ± 1.566	2.674 ± 1.051	3.438 ± 2.276
12	3.924 ± 0.953	2.136 ± 1.205	3.754 ± 1.731	3.469 ± 0.798
13	1.258 ± 0.647	1.351 ± 0.911	0.233 ± 0.125	0.825 ± 0.912

Table 9. Individual IC Hip Flexion Angles (°)

Participant	NTA	NTUA	WTA	WTUA
1	16.6 ± 3.6	18.3 ± 3.1	12.0 ± 1.9	15.5 ± 4.1
2	23.3 ± 9.4	26.2 ± 5.2	27.8 ± 8.0	22.5 ± 4.6
3	26.3 ± 5.4	27.5 ± 5.5	35.8 ± 5.8	28.6 ± 4.9
4	28.1 ± 3.1	24.0 ± 2.4	33.9 ± 2.7	26.0 ± 4.4
5	24.6 ± 6.6	23.8 ± 6.3	27.9 ± 2.5	23.1 ± 6.6
6	24.6 ± 3.8	28.7 ± 6.2	17.7 ± 4.8	18.1 ± 2.6
7	21.3 ± 5.5	16.5 ± 7.1	14.9 ± 4.5	11.3 ± 1.6
8	6.0 ± 4.6	15.6 ± 4.1	8.6 ± 2.5	11.2 ± 2.9
9	22.9 ± 3.2	26.0 ± 4.2	24.0 ± 2.2	24.6 ± 4.3
10	9.3 ± 5.2	7.5 ± 4.7	14.2 ± 4.4	15.0 ± 5.6
11	28.9 ± 2.5	21.0 ± 2.6	30.3 ± 3.8	20.7 ± 5.7
12	29.2 ± 5.0	25.7 ± 2.7	33.8 ± 2.7	26.0 ± 1.1
13	26.2 ± 7.6	24.2 ± 5.2	27.2 ± 3.8	21.1 ± 4.1

Table 10. Individual IC Knee Flexion Angles (°)

Participant	NTA	NTUA	WTA	WTUA
1	-9.9 ± 1.8	-8.5 ± 2.1	-6.4 ± 0.8	-6.2 ± 1.6
2	-13.7 ± 2.9	-7.8 ± 1.7	-10.5 ± 1.8	-7.0 ± 2.1
3	-14.8 ± 1.9	-8.4 ± 2.9	-8.2 ± 2.4	-10.0 ± 2.8
4	-18.3 ± 2.4	-12.6 ± 3.8	-18.3 ± 4.8	-10.8 ± 4.3
5	-16.6 ± 3.7	-14.6 ± 3.9	-16.9 ± 4.6	-13.3 ± 2.2
6	-18.1 ± 8.8	-21.7 ± 13.3	-11.9 ± 5.7	-11.9 ± 3.5
7	-9.6 ± 4.8	-14.1 ± 5.8	-12.4 ± 4.2	-10.2 ± 2.9
8	-0.6 ± 2.6	-6.3 ±3.3	-1.0 ± 2.0	-4.7 ± 2.1
9	-15.1 ± 2.5	-15.9 ± 1.9	-15.7 ± 1.2	-14.7 ± 2.7
10	-8.4 ± 2.6	-6.6 ± 1.7	-8.1 ± 2.2	-7.3 ± 1.1
11	-10.7 ± 2.7	-6.0 ± 2.8	-9.3 ± 3.1	-6.6 ± 2.0
12	-26.1 ± 2.7	-19.9 ± 4.1	-23.1 ± 3.6	-18.6 ± 2.6
13	-17.0 ± 4.7	-14.8 ± 1.1	-14.2 ± 2.0	-14.7 ± 1.7

Table 11. Individual IC Knee Abduction Angles (°) (Abd (-), Add(+))

Participant	**NTA**	**NTUA**	**WTA**	**WTUA**
1	0.4 ± 0.4	1.9 ± 0.9	1.4 ± 0.7	3.9 ± 0.9
2	-1.7 ± 0.7	-0.5 ± 1.1	-2.4 ± 1.0	0.9 ± 0.7
3	-3.2 ± 0.8	-0.7 ± 1.0	-1.8 ± 0.3	-0.6 ± 0.5
4	2.1 ± 1.6	2.3 ± 0.8	0.9 ± 1.3	2.3 ± 1.2
5	-4.1 ± 1.2	-3.3 ± 2.0	-5.1 ± 0.8	-5.6 ± 1.5
6	-4.0 ± 4.2	-3.1 ±1.7	-6.6 ± 1.3	-2.4 ± 2.1
7	-4.4 ± 0.4	-5.3 ± 1.0	-5.8 ± 0.3	-5.3 ± 0.7
8	-7.0 ± 1.1	-5.6 ± 0.9	-7.4 ± 0.9	-5.7 ± 0.8
9	-1.7 ± 1.1	1.1 ± 1.5	-0.1 ± 1.1	1.6 ± 1.2
10	-0.9 ± 0.5	-1.2 ± 0.5	-2.1 ± 0.8	-2.2 ± 0.8
11	-4.1 ± 0.9	-1.4 ± 0.6	-2.7 ± 0.9	-1.4 ± 0.5
12	-9.1 ± 1.6	-8.3 ± 2.1	-8.0 ± 0.6	-7.8 ± 1.7
13	-2.7 ± 1.1	-1.6 ± 0.8	-1.3 ± 0.3	-1.8 ± 0.7

Table 12. Individual Peak Hip Flexion Angles (°)

Participant	NTA	NTUA	WTA	WTUA
1	50.8 ± 6.4	53.2 ± 4.5	39.5 ± 2.7	46.7 ± 4.8
2	72.3 ± 2.3	71.2 ± 6.7	63.2 ± 9.3	58.2 ± 14.4
3	53.0 ± 3.6	53.4 ± 5.9	53.9 ± 7.6	53.6 ± 7.4
4	78.2 ± 2.0	83.1 ± 3.0	74.8 ± 4.6	74.3 ± 3.9
5	72.2 ± 2.9	73.4 ± 3.3	73.8 ± 2.7	65.9 ± 6.5
6	48.2 ± 2.8	61.3 ± 7.1	42.7 ± 2.2	49.6 ± 5.2
7	55.7 ± 4.2	51.8 ± 5.3	44.8 ± 1.8	44.7 ± 5.0
8	30.0 ± 9.8	46.4 ± 8.8	40.0 ± 6.2	34.0 ± 3.3
9	50.4 ± 6.8	62.9 ± 3.6	56.6 ± 4.1	62.4 ± 4.1
10	30.5 ± 7.6	27.4 ± 4.4	34.9 ± 5.7	37.0 ± 2.9
11	64.9 ± 4.9	62.1 ± 1.7	58.6 ± 3.2	58.7 ± 3.6
12	52.6 ± 8.3	63.6 ± 2.5	63.0 ± 4.3	59.0 ±7.2
13	53.2 ± 8.7	52.6 ± 9.4	54.6 ± 5.3	56.0 ± 5.8

Table 13. Individual Peak Knee Flexion Angles (°)

Participant	**NTA**	**NTUA**	**WTA**	**WTUA**
1	-72.4 ± 5.3	-70.8 ± 3.0	-71.2 ± 3.0	-71.3 ± 4.8
2	-66.7 ± 4.2	-71.7 ± 8.1	-63.8 ± 5.9	-67.8 ± 11.3
3	-63.1 ± 4.3	-60.5 ± 4.7	-58.6 ± 5.6	-62.0 ± 5.5
4	-68.3 ± 4.3	-67.9 ± 2.9	-70.4 ± 2.7	-71.0 ± 5.4
5	-74.9 ± 6.3	-73.2 ± 5.9	-69.0 ± 1.8	-68.5 ± 10.1
6	-63.8 ± 5.4	-69.4 ± 6.8	-57.5 ± 4.6	-67.6 ± 4.6
7	-71.0 ± 5.9	-73.2 ± 5.8	-69.1 ± 4.8	-68.9 ± 5.9
8	-43.3 ± 10.9	-53.9 ± 14.5	-50.1 ± 7.7	-44.1 ± 8.0
9	-72.7 ± 3.7	-80.3 ± 2.1	-76.4 ± 5.0	-82.4 ± 3.1
10	-51.7 ± 4.9	-47.1 ± 1.8	-51.9 ± 4.5	-53.8 ± 4.7
11	-55.1 ± 1.5	-59.5 ± 2.1	-59.4 ± 3.3	-56.9 ± 2.0
12	-70.3 ± 4.0	-72.2 ± 2.3	-71.1 ± 7.7	-70.8 ± 4.8
13	-62.5 ± 6.3	-66.6 ± 9.9	-61.0 ± 9.8	-66.3 ± 5.0

Table 14. Individual Peak Knee Abduction Angles (°)

Participant	NTA	NTUA	WTA	WTUA
1	-15.8 ± 0.9	-16.8 ± 2.3	-15.5 ± 1.7	-13.5 ± 1.4
2	-16.4 ± 3.4	-13.1 ± 1.4	-16.9 ± 1.4	-10.0 ± 2.8
3	-11.0 ± 2.7	-6.3 ± 1.3	-8.1 ± 1.7	-6.3 ± 1.5
4	-4.3 ± 1.4	-6.2 ± 2.6	-6.4 ± 1.9	-6.9 ± 1.2
5	-11.5 ± 1.5	-12.8 ± 3.0	-14.1 ± 3.1	-13.7 ± 0.3
6	-8.5 ± 3.2	-4.6 ± 1.6	-12.4 ± 0.8	-4.3 ± 2.4
7	-13.6 ± 1.2	-13.5 ± 3.0	-13.2 ± 0.5	-13.4 ± 1.5
8	-12.2 ± 2.5	-10.0 ± 1.2	-12.0 ± 3.2	-8.8 ± 2.2
9	-3.3 ± 1.0	-0.8 ± 1.1	-2.4 ± 2.1	-1.2 ± 1.6
10	-7.9 ± 2.1	-7.5 ± 0.9	-9.3 ± 1.9	-7.7 ± 1.5
11	-7.9 ± 1.8	-9.8 ± 2.2	-10.9 ± 1.3	-8.1 ± 1.6
12	-16.8 ± 3.4	-15.3 ± 2.7	-18.6 ± 1.2	-15.7 ± 2.1
13	-9.6 ± 0.4	-9.6 ± 1.5	-8.4 ± 1.0	-8.3 ± 1.9

Table 15. Individual Peak Hip Extension Moments (Nm/kg)

Participant	NTA	NTUA	WTA	WTUA
1	-2.11 ± 0.12	-2.13 ± 0.18	-1.89 ± 0.37	-1.73 ± 0.15
2	-3.23 ± 0.46	-4.06 ± 1.70	-3.09 ± 0.28	-2.00 ± 0.86
3	-2.02 ± 0.35	-2.07 ± 0.31	-1.88 ± 0.27	-2.28 ± 0.42
4	-2.87 ± 0.20	-2.76 ± 0.20	-2.76 ± 0.17	-2.45 ± 0.26
5	-2.88 ± 0.28	-2.09 ± 0.45	-2.41 ± 0.33	-2.31 ± 0.65
6	-2.19 ± 0.24	-2.31 ± 0.70	-2.04 ± 0.07	-1.21 ± 0.70
7	-2.26 ± 0.24	-1.34 ± 0.58	-1.60 ± 0.20	-1.25 ± 0.32
8	-1.64 ± 0.46	-2.12 ± 0.31	-1.68 ± 0.25	-1.40 ± 0.13
9	-2.59 ± 0.27	-2.49 ± 0.18	-2.44 ± 0.26	-1.75 ± 0.49
10	-2.03 ± 0.35	-2.21 ± 0.49	-2.56 ± 0.38	-2.85 ± 0.43
11	-1.69 ± 0.27	-0.68 ± 0.26	-1.41 ± 0.67	-0.72 ± 0.19
12	-1.76 ± 0.45	-1.74 ± 0.23	-2.15 ± 0.15	-1.60 ± 0.35
13	-2.21 ± 0.44	-2.05 ± 0.36	-2.38 ± 0.10	-1.80 ± 0.35

Table 16. Individual Peak Knee Extension Moment (Nm/kg)

Participant	NTA	NTUA	WTA	WTUA
1	2.66 ± 0.21	2.25 ± 0.35	2.63 ± 0.31	2.79 ± 0.55
2	2.06 ± 0.80	2.32 ± 0.61	2.33 ± 0.68	2.64 ± 0.71
3	1.83 ± 0.27	2.14 ± 0.55	1.71 ± 0.54	2.46 ± 0.57
4	1.62 ± 0.50	1.45 ± 0.50	1.63 ± 0.28	1.73 ± 0.31
5	1.25 ± 0.25	1.15 ± 0.43	1.34 ± 0.32	1.37 ± 0.29
6	1.90 ± 0.16	1.82 ± 0.67	1.68 ± 0.37	2.09 ± 0.50
7	2.33 ± 0.43	2.18 ± 0.77	2.31 ± 0.13	2.28 ± 0.22
8	1.26 ± 0.32	1.59 ± 0.38	1.86 ± 0.25	1.93 ± 0.34
9	2.35 ± 0.63	1.87 ± 0.69	2.37 ± 0.19	1.86 ± 0.36
10	2.20 ± 0.35	2.09 ± 0.06	2.42 ± 0.33	2.26 ± 0.39
11	1.17 ± 0.27	1.11 ± 0.12	1.34 ± 0.42	1.28 ± 0.26
12	2.50 ± 0.23	2.27 ± 0.32	2.43 ± 0.42	2.49 ± 0.24
13	2.58 ± 0.39	3.09 ± 0.81	2.95 ± 0.50	2.91 ± 0.35

Table 17. Peak Knee Adduction Moment (Nm/kg)

Participant	**NTA**	**NTUA**	**WTA**	**WTUA**
1	0.91 ± 0.19	0.93 ± 0.17	0.74 ± 0.09	0.64 ± 0.20
2	0.75 ± 0.26	0.67 ± 0.09	0.86 ± 0.20	0.53 ± 0.19
3	0.21 ± 0.09	0.24 ± 0.08	0.19 ± 0.11	0.19 ± 0.07
4	0.08 ± 0.19	0.23 ± 0.17	0.34 ± 0.19	0.29 ± 0.18
5	0.28 ± 0.08	0.41 ± 0.10	0.60 ± 0.10	0.70 ± 0.23
6	0.17 ± 0.11	0.17 ± 0.13	0.13 ± 0.19	0.08 ± 0.20
7	0.18 ± 0.10	0.36 ± 0.23	0.10 ± 0.06	0.16 ± 0.05
8	0.49 ± 0.26	0.32 ± 0.16	0.48 ± 0.37	0.22 ± 0.08
9	0.42 ± 0.14	0.38 ± 0.07	0.23 ± 0.17	0.35 ± 0.12
10	0.26 ± 0.19	0.27 ± 0.09	0.20 ± 0.15	0.28 ± 0.25
11	0.31 ± 0.12	0.34 ± 0.06	0.38 ± 0.15	0.27 ± 0.09
12	0.47 ± 0.10	0.37 ± 0.10	0.52 ± 0.09	0.36 ± 0.06
13	0.51 ± 0.10	0.68 ± 0.19	0.45 ± 0.13	0.61 ± 0.18

VITA

Michael Joseph O’Dwyer was born in Towson, MD in 1995 to Tom and Joyce O’Dwyer. Michael graduated from Towson High School in 2013, but prior to that achieved his Eagle Scout award in 2010. He graduated from Coastal Carolina University in 2016 with a B.S. in Exercise and Sport Science. After graduation, he completed a M.S. in Kinesiology with a concentration in Biomechanics at the University of Tennessee, Knoxville in Spring 2019. He will begin his professional career following Summer 2019 graduation.