THE EFFECTS OF AN IN-FLIGHT PERTURBATION ON LOWER EXTREMITY BIOMECHANICS DURING DROP LANDINGS

by

JAE POM YOM

(Under the Direction of KATHY J. SIMPSON)

ABSTRACT

Most ACL injury-related researchers have focused increasingly on noncontact injury, because more than 70% of ACL injuries are categorized as such. However, this is a misleading classification because during movements involving a flight, then a landing phase, contact with another person or object (i.e., an unexpected perturbation) during the flight phase may set into motion a cascade of abnormal movements and mechanics that lead to an ACL injury during the subsequent landing phase. However, the possible mechanisms of such an ACL injury situation are not known. Therefore, the purpose of this study was to determine the effects of an in-flight perturbation on lower extremity biomechanics of females displayed during a drop landing.

Seventeen collegiate-age-female recreational athletes performed baseline (BASE) then unexpected perturbation (PERT) or sham drop landings. For PERT trials, the cable attached to a proprietary perturbing machine pulled on one of the participant's shoulders in a lateral direction during the flight phase. High-speed, digital-video motion and ground reaction force (GRF) data were recorded for the landing phase. We compared 95% confidence intervals of PERT – BASE differences scores ($p<0.05$) for lower-extremity joint kinematics and kinetics and GRF, using one sample t-tests.

The results demonstrated that PERT, compared to BASE, exhibited more extended joint positions of the lower extremity at initial contact and greater knee abduction and hip adduction displacements. Additionally, PERT showed greater peak magnitudes of vertical and medial GRF, and knee and ankle extensor moments compared to BASE. Greater knee and hip adductor moments also were exhibited during PERT. We found that unexpected lateral in-flight perturbation leads to abnormal motions and GRF, but induces compensatory frontal plane moments that are beneficial for maintaining stability of the lower extremity from the lateral in-flight perturbation.

INDEX WORDS: Anterior cruciate ligament, joint kinematics, joint kinetics, landing, perturbation

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TABLE OF CONTENTS

LIST OF TABLES

LIST OF FIGURES

CHAPTER 1

INTRODUCTION

Background

Anterior cruciate ligament (ACL) injury is common and costly. For example, 250,000 ACL injuries are estimated to occur to one in 3,000 people in the United States annually (Miyasaka, et al., 1991). Medical expenses for treatments, such as ACL reconstruction and rehabilitation, are approximately $17,000 per injury in the United States. Thus, the annual cost of ACL injury rehabilitation is more than two billion dollars (Childs, 2002; Gottlob, et al., 1999).

Furthermore, for an injured individual, there are several other issues. Recovery from orthopedic surgery and other outpatient treatment takes approximately five months (Button, et al., 2005). ACL injury also may have serious, long-term consequences for the injured individual. Even with surgical repair, the ACL-injured population is exposed to a higher risk of developing osteoarthritis compared with an equivalent, uninjured population (Gillquist & Messner, 1999; Lohmander, et al., 2004; Myklebust & Bahr, 2005). Signs of degenerative progression also occur at a younger age to ACL-injured compared to non-injured individuals (Lohmander, et al., 2004).

Etiology likely involves multiple, interacting factors (Hewett, et al., 2006). Three important factors related to this study are gender, movements and the resulting loading parameters of the ACL. Relative to gender, within injured populations, female athletes' ACL injury rates are two to eight times higher than male counterparts of comparable athletic skill (Arendt, et al., 1999; Hewett, et al., 1999; Malone, et al., 1993).

One of the most common movements during which ACL injury occurs is one in which there is a landing phase after being airborne (Hewett, et al., 1996). It is during this landing phase that ACL injury can occur (Alentorn-Geli, et al., 2009; Boden, et al., 2000; Huston, et al., 2000). It is estimated that 31 % ACL injuries occur during landing phases (Boden, et al., 2000). In general, some of the movement conditions that are related to ACL strain are knee flexion angle; anterior-tibial/posterior-femoral translation; and tibial ab/adduction and axial rotation (Beynnon, et al., 1997; Cerulli, et al., 2003; Silvers & Mandelbaum, 2007; Weinhold, et al., 2007).

As ACL injury etiology is multi-factorial (Griffin, et al., 2006; Hughes & Watkins, 2006; Huston, et al., 2000), the effects of gender on ACL causation also are likely to be complex. However, one potential explanation for the disparity of gender injury rate is variation in landing biomechanics. Females and males display different body orientations at foot contact and subsequent landing kinematics and kinetics (Decker, et al., 2003; Ford, et al., 2003; Ford, et al., 2005; Kernozek, et al., 2005; Malinzak, et al., 2001; Salci, et al., 2004) that are believed to affect ACL loading (Fleming, et al., 2001; Weinhold, et al., 2007).

However, it is likely that some, but not all females are at risk of ACL injuries. Hewett (2005a) observed for a prospective study of 205 female collegiate athletes and reported that ACL injury group, compared to un-injured group, showed different knee posture and loading such as greater knee maximum abduction angle and moment, and greater vertical ground reaction force (GRF_V). Thus, for a better understanding of the role of landing strategies on female ACL injury etiology, it would be more revealing to investigate the variation of landing strategies among females than to compare a group of females to a group of males.

As ACL loading is complex and difficult to measure under realistic conditions, our knowledge is somewhat lacking of ACL injury mechanisms during actual human movement, especially for landing movements. The landing and loading conditions that can produce damage to the ACL during landing movements are mostly known, at present, from cadaver analysis of video clips from commercial television, and modeling studies (Cerulli, et al., 2003; Hewett, et al., 2009; Krosshaug, et al., 2007; Pflum, et al., 2004; Torzilli, et al., 1994).

One situation that we predicted would produce abnormal motions and joint kinetics related to ACL injury during the landing phase is one during which a “perturbation” is applied to the athlete while she is in the air. At contact, the performer’s body must be in an appropriate position to attenuate impact forces, reduce the body’s linear kinetic energy and maintain stability without excessively straining or stressing the ACL. However, if a perturbation is applied to the athlete, that is, a teammate or opponent applies force to the athlete during the flight phase, the athlete may land in a nonoptimal position, either reducing the athlete’s ability to achieve these goals or requiring compensatory actions to land safely. We anticipated, therefore, that an abnormal landing would lead to abnormal knee motions that may subsequently disrupt stability and result in abnormal inertial and ground reaction forces (GRF) as well as compensatory knee joint moments. Consequently, these reactions could indicate excessive strain and stress to the ACL during the landing phase. Moreover, the lower extremity may be unable to use typical landing strategies, that is, lower extremity flexion combined with eccentric extensor muscle torques that act to reduce the body’s kinetic energy safely via negative angular muscle work.

It is likely, therefore, that certain perturbations *in-situ* could initiate an ACL injury episode. Indeed, the effects of perturbations on landing biomechanics are thought to be very relevant to ACL injury causation (Shultz, et al., 2008; Shultz, et al., 2010). However, our understanding of the effects of perturbations on landing biomechanics is very limited, as little research exists (Arnett, 2007). Only a few studies have investigated the landing biomechanics of perturbations applied to the athlete as a method to understand the mechanisms of this type of indirect contact ACL injury (Arnett, 2007; Carcia, et al., 2005; Schmitz, et al., 2004; Shultz, et al., 2006; Shultz, et al., 2001). Furthermore, Arnett (2007), to my knowledge, conducted the only research in which the performer was perturbed while in the air. Arnett chose to mimic perturbations in which an athlete is pushed from behind. For Arnett's study, the anterior perturbations were applied to the center of mass (COM) of the body (linear perturbation) and slightly above the COM (rotational perturbation about the medio-lateral axis, i.e., somersault axis), respectively. Interestingly, both perturbation conditions showed significantly less vertical GRF compared to the non-perturbed condition. However, the linear perturbation showed significantly greater peak hip and knee extensor moment and ankle plantarflexor moment compared to the non-perturbed condition. However, there were no differences between perturbation and non-perturbation conditions for lower extremity joint kinematics and kinetics.

While Arnett's (2007) work provided understanding of situations in which athletes get pushed forward while in the air, players may be pushed from an opponent in the sideways direction. Due to musculoskeletal constraints of the lower extremities, movement strategies that can be used during landing to compensate for lateral perturbations are different and more limited compared to an anterior-directed perturbation. Initially, upon landing after a lateral perturbation, there may be a smaller base of support, reducing stability. In addition, either excessive ab/adduction joint movement could occur to control frontal plane motions of the trunk and lower extremity segments or greater frontal plane moments must be generated to counteract/prevent excessive motions. Moreover, if there is abnormal medial-lateral joint alignment at initial ground contact (IC), greater moments due to joint reaction forces would occur, as their moment arms would be longer. Additionally, with less ability to attenuate GRF_V and body inertial forces, these forces would also increase joint reaction force moments. Hence, it is imperative to determine the effects of medio-lateral perturbation on landing biomechanics of women to better understand ACL injury causation (Figure 1).

Purpose of the study

For these reasons, determining the biomechanical effects of medio-lateral types of in-flight perturbations is of urgent importance. Moreover, it is imperative to understand these effects among female athletes. Therefore, the overall purpose of the study was to determine the biomechanical effects of lateral flight-phase perturbations applied to female athletes during drop landings. The perturbation was a lateral pull on the trunk at shoulder height that created a lateral flexion torque applied to the trunk. The specific purpose of study was to determine if applying an unexpected perturbation during the flight phase results in altered lower extremity biomechanics compared to the biomechanics displayed during a typical landing.

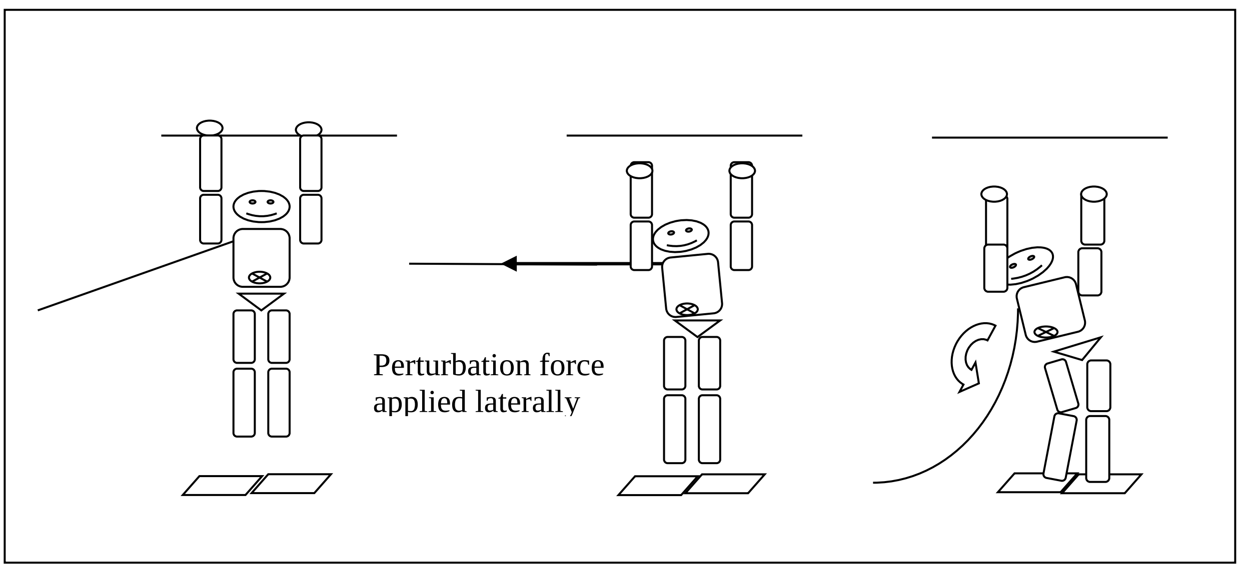

Figure 1. Diagram of the perturbation task. The left picture show the start position of the drop landing; the middle picture show expected body motions after in-flight perturbation; and the right picture represent expected finish position of the drop landing.

Significance of the study

Signed into law in 1972, Title IX was an amendment prohibiting discrimination based on gender within educational institutions receiving federal money. The enforcement of Title IX since that time has led to increasing numbers of female participants in school-based sports programs. For example, female participation at the high school level has significantly increased from 294,015 to 2,908,390 in the last 30 years (NFHS, 2005). The dramatically increased number of females participating in sports in many settings, as well as within educational-related programs, therefore, also has increased the number of female athletes who suffer an ACL injury.

However, this does not totally explain the increased numbers of ACL injuries to women. In the United States, ACL injuries are two to eight times more common in females compared with age- and skill-matched male athletes when injury rates are scaled to athlete-exposure hours (Arendt, et al., 1999; Hewett, et al., 1999; Malone, et al., 1993).

Decreasing the rate of ACL injury is of high importance, as a torn or ruptured ACL can result in costly and lengthy treatment, and a higher risk of osteoarthritis in the affected knee joint (Engebretsen, et al., 1993; Gillquist & Messner, 1999; Lohmander, et al., 2004; Myklebust & Bahr, 2005; Noyes, et al., 1980). In the past 15 years, more than 4,000 articles have been published related to ACL injury, on topics such as ACL injury mechanisms, rehabilitation, and prevention (Uhorchak, et al., 2003). Most of the ACL injury researchers have focused on "noncontact" ACL injury etiology (those injuries not caused by a contact force directly applied to the knee region) because this type of ACL injury accounts for approximately 70% of ACL injuries (Boden, et al., 2000). Therefore, an ACL injury that occurs during the landing phase due to an earlier, flight-phase perturbation would be defined as an indirect contact injury. thus, understanding the mechanisms involved in safe landings following in-flight perturbations may reduce the incidence of indirect ACL injury.

According to the authors of *Research Retreat IV: ACL injuries--the gender bias* consensus statement (Shultz, et al., 2008), a perturbation applied to the player prior to landing may be a very common scenario of ACL injury. They also stated that future ACL research should focus on investigating the mechanics of the underlying injury mechanisms. However, there is little research focused on the effects of a perturbation due to contact with another player (or object) that occurs prior to, but not during the interval of time when the ACL is damaged (Hewett, et al., 2007b). Most of these studies investigated the effects of perturbation on lower extremity kinematics and/or muscle activation while the performer was standing on one leg (Carcia, et al., 2005; Schmitz, et al., 2004; Schultz, et al., 2000; Shultz, et al., 2006; Shultz, et al., 2000; Shultz, et al., 2001). Arnett (2007) conducted the only research in which the performer was perturbed while in the air. His results demonstrated some differences in landing mechanics of perturbation compared to non-perturbation conditions occur. This suggests that this topic requires greater investigation to more thoroughly understand the relationship between perturbations and ACL injury.

The outcomes of this study would provide a more comprehensive understanding of the mechanics created by perturbations leading to indirect contact ACL injuries. As explained earlier, I predicted that the perturbation proposed for this study was likely to produce greater potential for abnormal landing mechanics than the perturbations applied during Arnett’s studies. The effects on an athlete of a perturbation likely are very sensitive to the direction and the location that the perturbation is applied: rotational momentum; the limits of stability; potential muscle synergies; the ability to absorb mechanical energy and impact forces effectively may all be affected. Consequently, the in-flight perturbation may induce abnormal knee motions and loading. Thus, the comprehensive understanding of the lower extremity responses from the lateral perturbation would provide information for new design of the prophylactic lower extremity braces or new training protocol to reduce ACL injury rate.

Conversely, it is crucial to understand typical landing mechanics, as athletes endure many contacts with other players during ‘noncontact’ sports such as basketball, soccer and blocking in volleyball. For the great majority of these situations, these athletes land safely. Hence, by understanding safe and typical landings, recommendations may emerge from those who develop ACL injury prevention guidelines and programs. Last, by understanding perturbation biomechanics among female athletes, we may begin to identify or confirm those factors that contribute to ACL injury risk within the female athlete population.

Predicted Outcomes and Justifications of Outcomes

To further understand the detailed predictions and the justifications supporting these predictions that are explained below, first, the phases of a typical movement relevant to this study (e.g., a basketball rebound) are defined. The phases of interest included two major phases: flight, and landing. The flight phase begins at the instant of take-off from the ground and ends at the instant of initial ground contact. An example of a flight phase is when the athlete is in the air during a defensive basketball rebound. Also, the landing phase begins with ground contact and ends typically when the athlete’s center of mass stops moving downward.

When an ACL injury that occurs during a landing movement is due to an unexpected perturbation force applied during the flight phase, then how does this occur? To answer this, we start with the flight phase. The primary mechanical goal for the athlete often is to accomplish some performance objective (e.g., grabs and keeps possession of the basketball during a defensive rebound). The second mechanical goal of the flight phase is to position the body in preparation for achieving a stable landing and subsequent goal-related actions and/or other movements (e.g., pass off the ball to another player and start running down the court).

When a perturbation occurs during flight, a force is applied to the player. For when the defensive player gets pushed by an opponent while in the air, then that push is a perturbation force. Therefore, several mechanical factors related to this perturbation force will directly influence the linear and rotational effects that the perturbation would have on the player's subsequent mechanics: the location on the body (point of application) where the perturbation was applied, line of action of the perturbation force, the direction that the force is applied along that line of action, and the magnitude of the perturbation force applied to the player. Consequently, these factors affect the magnitude and direction of the amount of linear momentum gained due to the perturbation. Moreover, due to the direction and point of application that these perturbations are applied, torque(s) also are generated by the perturbation force. Therefore, the perturbations also cause the performer to start rotating about the trunk's anterior-posterior axis.

The other major factor that directly influences the effects that the perturbing force has on the linear and rotational motion of the performer is the length of time that the perturbation is applied to the performer. Therefore, the magnitude of the perturbation is the *impulse* applied to the performer. Simplistically, this can be defined as *the average perturbation force multiplied by the time that the perturbation force is applied to the trunk during the flight phase*. As the perturbation force will cause torque(s) to be applied to the performer in this study, rotational impulse also is created during the flight phase and can be expressed as: *torque multiplied by the time that the perturbation torque is applied to the trunk during the flight phase*.

Thus, we can determine how the performer's momenta would be changed due to the linear and angular perturbation impulses. According to the impulse-momentum principle (assuming a rigid object), the magnitude of an impulse applied to an object or a performer causes that amount of change of the object's momentum. For this study, the performer would gain linear momentum in the direction that the perturbation force was applied, and angular momentum (rotational motion). Based on the principles of the conservation of linear and angular momentum (assuming negligible air resistance), during the flight phase, the magnitudes of linear and rotational momenta of the body remain the same until contact with another object (e.g., the ground) or person (perturbation) occurs. When perturbation impulses are applied, the performer linear and rotational momenta increase, and the performer cannot alter the magnitude of horizontal linear and angular momenta until ground contact. Compared to a typical flight phase, during a perturbation situation, the player's increased linear and rotational momenta were expected, thusly, to influence body positioning at landing and resulting landing mechanics.

For the subsequent landing phase, the mechanical goals are to reduce body's downward vertical velocity to zero while landing safely; possibly to accomplish a performance objective (pass the ball to a teammate); and prepare to perform the next movement (e.g., running down the court). To optimally achieve the mechanical goals, contacting the ground with both feet and flexing the lower extremity are performed during the landing phase. Also, the body's vertical kinetic energy can be reduced via internal muscle work and optimal stability attained. In terms of muscle contributions to energy absorption, this is done primarily through negative muscle work by extensor muscles acting across the lower extremity joints while the lower extremity joints are moving through flexion. Concurrently, this landing strategy also should attenuate vertical impact forces and internal inertial forces, hence, reduce the potential for abnormal and/or excessive tissue loading to be applied to structures of the lower extremity.

Among the axis directions, anatomically, the flexion/extension (FL/EXT) axes of the lower extremity are better designed to raise and lower the body. Therefore, not only can flexing the joints of the lower extremities help absorb kinetic energy, but joint flexion increases the body's stability by lowering the COM.

To prevent excessive forces from being applied to one leg, during initial contact with the ground, both feet should contact the ground simultaneously, with the body's center of gravity located approximately midway between the right and left foot. This would likely allow both feet to apply approximately the same amount of force to the ground during the subsequent landing phase, and, thus, the corresponding vertical ground reaction force (GRF_V) magnitudes applied to each leg would be more equivalent. In addition, to achieve create optimal stability and minimize abnormal joint motions and non-compressive bone loading, the segments of a given leg should be in alignment so that they would move in the same sagittal plane during joint flexion about the lower extremity. Likely, lower extremity flexion also reduces the effective mass involved in the collision of the body with the ground. Consequently, this landing strategy should minimize the potential for abnormal joint reaction moments, hence the need for lower extremity joint moments being generated in compensation, particularly about the frontal plane axes. As a result, the landing strategy should prevent abnormal loading to musculoskeletal and soft tissues within a particular leg (Beynnon, et al., 1997; Norcross, et al., 2010; Silvers & Mandelbaum, 2007).

However, for drop landings during which the perturbation is applied, I anticipated several biomechanical consequences, due to the additional angular and lateral linear momenta that the performer would have at contact. First, the performer would be applying greater lateral forces to the ground compared to the normal landings. This would cause increased medial GRF (GRF_M) acting on the body due to Newton's law of action-reaction. Additionally, at contact, because the perturbation force pulled the trunk laterally, the medio-lateral location of the COM of the body (COM_{BODY}) also would be shifted laterally toward the leg closer to the perturbation source (ipsilateral leg). Thus, more downward inertial force would be applied to the ground by the ipsilateral leg and less by the contralateral leg during the subsequent landing phase. Hence, the ipsilateral leg would experience the greater increase of GRF_V of the two legs. Greater GRF_M also would exhibit on the ipsilateral leg in order to returning laterally shifted COM_{BODY} to neutral landing position.

The second major mechanical consequence of the perturbation impulse expected would be increased greater knee abduction (ABD) angle and/or net muscle moments acting on the distal femur. When the trunk is pulled laterally by the perturbation, it was anticipated that the trunk would pull the pelvis and proximal femur laterally, causing the femur to rotate in the frontal plane, such that the hip distal femur would be adducted at initial ground contact (IC). Hence, the adducted femur would lead to an increased hip adduction (femoral adduction relative to the pelvis) angle at IC. Consequently, at the knee joint, participants would exhibit an increased knee abduction angle (tibial abduction relative to the femur) at IC. Ankle inversion also would be exhibited due to the lateral shifted COM_{BODY} at IC, compared to BASE.

In addition, during landings after the perturbation was applied, the flexion/extension landing strategy to minimize potential abnormal lower extremity joint alignment and motion would be compromised. Consequently, this would be decreased the lower extremity sagittal plane displacements.

In summary, I anticipated that, compared to non-perturbation drop landings, the perturbation landings would display greater peak medial and vertical GRFs. Furthermore, less knee flexion angular DISP and greater tibial abduction angular DISP and hip adduction DISP would be shown between initial ground contact and the end of the landing phase. In addition, the perturbation drop landing would show greater lower extremity extensor net muscle moments and knee adductor moment compared to the non-perturbation drop landings.

Hypotheses

Compared to typical landings (BASE), for the perturbation condition (PERT), the leg of interest would exhibit during the landing phase:

Increased ankle inversion, knee abduction, hip adduction IC angles.

Increased ankle inversion, knee abduction, and hip adduction displacements.

Decreased ankle, knee, hip joint flexion displacements.

Increased peak vertical and medial GRF.

Increased lower extremity flexor/extensor net muscle moments.

Increased lower knee adductor net muscle moment

CHAPTER 2

REVIEW OF LITERATURE

This literature review consists of four major parts: 1) Anatomical normal knee, 2) ACL injury mechanism, 3) gender bias of ACL injury and 4) biomechanical roles of trunk during drop landing.

Knee anatomy

The most complex and largest joint in the body is the knee joint. The knee joint consists of two joints in one: the patellofemoral, and tibiofemoral joints with the C-shaped menisci. The patellofemoral joint consists of the patella and the anterior femoral plateaus. Also, the tibiofemoral joint consists of proximal tibia and the distal femur (Hughes & Watkins, 2006). The C- shaped menisci are located in the tibial articular surfaces. The tibiofemoral joint acts as a modified hinge which allows flexion and extension. When the knee is partly flexed, some rotation is possible. However, when it is fully extended, rotation is strongly restricted by the bicondylar structure of the knee joint and ligaments with the C- Shaped menisci (Marieb, 2004).

Major ligaments

Ligaments hold bone to bone and provide stabilization and guiding of joints (Childs, 2002). The knee joint has four major ligaments, anterior cruciate ligament (ACL), posterior cruciate ligament (PCL), medial collateral ligament (MCL), and lateral collateral ligament (LCL). The ligaments provide joint stability and restricted range of motion in the knee joints.

The ACL and PCL are called intracapsular ligaments because they are placed in the notch between the femoral condyles. The ACL attaches from the anterior intercondylar of the tibia to the posterior femur on the medial side of its lateral condyle. On the other side, the PCL is attached to the posterior intercondylar area of the tibia and passes anteriorly, medially, and upward to attach to the lateral side of the medial femoral condyle (Marieb, 2004).

The MCL attaches from the medial femoral condyle to the middle of the medial condyle of tibia. Also, the LCL is located on the lateral side knee joint, just as its the name indicated. The origin is the middle of the lateral femoral condyle and the lateral head of the fibula. The function of the ligaments is to prevent tibial abduction and adduction when the knee is extended (Marieb, 2004).

<u>The function of the cruciate ligaments</u>

As I mentioned above, the cruciate ligaments pass through the intercondylar notch and cross each other. The ACL and hamstrings work together to prevent anterior displacement of the tibia relative to the femur by resisting forward movement of the tibial plateaus (Frank & Jackson, 1997; Marieb, 2004; Nordin & Frankel, 2001). On the other hand, the PCL works with quadriceps to help prevent posterior displacmenet of the tibia and femur by restricting backward movement of the tibial plateaus (Hughes & Watkins, 2006). The major function of the cruciate ligaments is to provide stabilization of the knee joint, and prevent abnormal movements (Nordin & Frankel, 2001).

<u>Muscles and menisci of the knee joint</u>

The knee joint has two major muscle groups (quadriceps and hamstrings) as knee extensor and knee flexors. The quadriceps is the powerful knee extensor muscle group and it arises from four separate heads (rectus femoris (RF), vastus lateralis (VL), vastus medialis (VM), and vastus intermedius (VI)). The quadriceps muscle extends the knee joint and stabilizes knee joint.

The hamstring muscle group is located in the posterior thigh and it is divided among the biceps femoris, semitendinosus, and semimembranosus. These muscles have the same origin on the ischial tuberosity and they act like knee flexors (Marieb, 2004). Also, they cross both the hip and knee joints and may also act as hip flexors.

The menisci, crescent-shaped wedges of fibrocartilage, are located between the condyles of the femur and tibia(Hewett, et al., 2007a). Approximate 50 % to 85 % of compressive load of the knee is transmitted through the menisci. The major function of the menisci is absorbing the compressive load to protect the articular surfaces and help prevent rocking of the femur on the tibia (Ahmed & Burke, 1983; Marieb, 2004).

Differences between ruptured ACL knee and normal knee

There are a considerable number of studies on alteration of ACL ruptured knees because anterior-posterior and rotational stabilization is the major function of the ACL. Muneta et al. (1996) used lateral radiographs to define the extension angle of the healthy knee and the unilateral ACL ruptured knee while the participants were lying supine. They reported that the injured knees' extension loss was statistically significant. In addition, a knee with absence of ACL showed significantly less knee stiffness compared with a normal knee at lower levels of forces with non-weight bearing situation (Markolf, et al., 1984). On the other hands, Kvist et al. (2007) reported that ACL-deficient knees show less tibial translation during activity, even though the knee shows greater static tibial translation.

ACL injury mechanisms

Several researchers divided and characterized the ACL injury factors as an intrinsic and extrinsic (Arendt & Dick, 1995; Chappell, et al., 2005; Hewett, et al., 2006; Huston, et al., 2000). Even though, there are slightly different definitions between intrinsic and extrinsic injury factors by researchers, main characteristics are matched each other. For example, neuromuscular, anatomical (Q-angle, femoral notch), and hormonal factors belong the intrinsic in ACL injury mechanism. On the other hand, extrinsic mechanisms of injury consist of landing characteristics, motion perturbations, and shoe-surface interaction (Arendt & Dick, 1995; Hewett, et al., 2006; Huston, et al., 2000).

We can also describe ACL injury mechanisms as non-contact, direct contact, and indirect contact among the extrinsic mechanisms. The non-contact ACL injury means that the injury person is not hit by another person or object when the injury occurs (Colby, et al., 2000). Usually, the non-contact ACL injury is occurred with a sudden deceleration prior to a landing or change of direction (Boden, et al., 2000). Based on the systematic video analysis studies and data review studies, several researchers reported that large proportion, more than 70 percent, of ACL ruptures occur by a result of non-contact injuries (Boden, et al., 2000; Myklebust, et al., 1998; Noyes, et al., 1983). For examples, Noyes et al. (1983) checked ACL ruptured 103 high school and collegiate sports team athletes. They reported that 80 of 103 ACL ruptures were occurred by non-contact injury. Myklebust et al. (1998) reported that 89 percent ACL injury occurred in non-contact situation. In addition, Boden et al. (Boden, et al., 2000) interviewed and reported that 72 percent of patients were non-contact ACL ruptured in their study. However, when ACL injuries occur with external forces, it is categorized as a direct or indirect contact ACL injury. The direct contact injuries are caused by directly applied an external force in the injured knee and is the proximate cause of the injury (Hewett, et al., 2007b) as seen in tackling in American Football . On the other hand, the indirect contact injuries mean that ACL is injured by the external force applied on upper extremity such as being out of balance due to pulled or pushed by another player and it is not the proximate cause of the injury (Hewett, et al., 2007b).

Gender bias of ACL injury

Although total numbers of ACL incidence of male athletes are still significantly greater than females, most studies, related to ACL injury, have reported female had a higher (four to eightfold) ACL injury rate than male athletes (Agel, et al., 2005; Arendt & Dick, 1995; Ferretti, et al., 1992; Malone, et al., 1993). Especially, Agel et al. (2005) reported 682 ACL injuries (514 women, 168 male) that occurred in basketball and 586 ACL injuries (394 women, 192 men) in soccer based on NCAA Injury Surveillance System (ISS) for 1990 to 2002. Furthermore, female athletes of soccer and basketball players have sustained a significantly higher rate of ACL injuries than did male soccer and basketball players. Female high school and collegiate basketball and soccer players were shown statistical significant higher injury rates than male (Arendt & Dick, 1995; Borowski, et al., 2008).

The higher incidence rate of ACL injuries among female athletes has been documented (Arendt, et al., 1999; Arendt & Dick, 1995; Hewett, et al., 1999). There are potential risk factors that may increase female ACL injury rates and those factors can be divided by anatomical (structural), and biomechanical-neuromuscular risk factors.

Anatomical risk factors

First, structural alignment (anatomical: Quadriceps femoris angle, femoral notch characteristics, patella tendon tibia shaft angle, and joint laxity) of the lower extremity may contribute to the disparity of ACL injury rates between males and females (Hughes & Watkins, 2006; Huston, et al., 2000; Silvers & Mandelbaum, 2007). The Quadriceps femoris angle (Q angle) is considered as one of the anatomical risk factors that contributes to differential ACL injury rates among males and females. The static Q angle is determined by measuring the acute angle produced by two lines. The first line is drawn through anterior superior iliac spine and the midpoint of the patellar and the second drawn through the midpoint of the patella and the tibial tubercle (Huston, et al., 2000). Several studies have attempted to find the relationship between the Q angle and the gender discrepancy of ACL injury rates. These studies have consistently reported that young adult women have greater Q angles compared with young adult men. In addition, these studies have also found that the greater Q angle is a contributing factor to the knee injuries by altering lower extremity kinematics (Csintalan, et al., 2002; Heiderscheit, et al., 2000; Horton & Hall, 1989; Livingston, 1998; Mizuno, et al., 2001; Moul, 1998; Shambaugh, et al., 1991; Woodland & Francis, 1992; Zelisko, et al., 1982). According to Shambaugh et al. (1991), the mean Q angles of athletes sustaining knee injuries (4 of the 45 athletes) were significantly larger than the mean Q angles for the athletes who were not injured (injured group:14±2.5° vs. uninjured group: 10±3.5). Also, Horton and Hall(1989) observed that mean Q angles in females were 4.6° greater than mean Q angles in males. In contrast, other studies not only have reported that the static Q angle was not related to ACL injury rates but also, it can possibly change with an isolated quadriceps contraction (Guerra, et al., 1994; Hahn & Foldspang, 1997; Myer, et al., 2005a). As a result, we see that only the static Q angle cannot simply explain the disparity of ACL injury rates between males and females.

Several studies have been conducted to define the relationship between the femoral notch characteristics and the incidence of ACL injuries (Anderson, et al., 1987; Shelbourne, et al., 1998; Uhorchak, et al., 2003). Usually, those studies were initiated with the assumption that the stresses in a narrow intercondylar notch with smaller ACL will be greater for a given applied load. In a prospective study, Shelbourne et al. (1998) reported that the mean of intercondylar notch width of the femur was narrower in women. They also found that the mean notch width for the ACL injured group was significantly narrower compared with the non-ACL injury group. Uhorchak et al. (2003) reported similar results, demonstrating that females had narrower notch widths compared to males (15.6±2.9 vs. 17.7±3, respectively, $p \leq .05$), and the women who had less than 13 mm of notch width were at 16.8 times greater risk ratio than those with a larger notch width.

Notch width index (NWI) constitutes a different way to evaluate and measure femoral notch differences. The NWI is determined by the ratio of the width of the intercondylar notch to the width of the distal femur at the level of the popliteal groove. Several researchers have reported that there is no correlation between notch width index (NWI) and ACL injury rate (LaPrade & Burnett, 1994; Souryal & Freeman, 1993). In a prospective study of 213 collegiate athletes, LaPrade and Burnett (1994) concluded that there was no significantly statistical difference between gender and NWI and ACL injury rates. However, they found that femoral intercondylar notch stenosis and ACL injuries were correlated. Also, in a prospective study of 902 high school athletes, Souryal and Freeman (1993) showed similar conclusions this relationship.

The patella tendon-tibia shaft angle (PTTSA) is another potential anatomic risk factor of noncontact ACL injuries because the greater the PTTSA, the greater the anterior shear force applied to the proximal end of the tibia, resulting in potential strains on the ACL. The PTTSA is defined as the angle in the sagittal plane between the line of action of the patellar ligament and the long axis of the tibia (Hughes & Watkins, 2006). Nunley et al. (2003) measured 7 different knee flexion angles (from 0° to 90° flexion, at 15° intervals) and reported that mean of females' PTTSA was 3.7° greater than mean of males' PTTSA (*p*=0.00) over the 0~90° rage of knee flexion. The differences in PTTSA result in a 13.2% increase in the anterior shear force applied to the proximal end of the tibia(Nunley, et al., 2003). In addition, a number of studies have reported that most non-contact ACL injuries occurred between 20° of knee flexion and almost fully extension (Boden, et al., 2000; McNair, et al., 1990; Olsen, et al., 2004). When combined with the PTTSA and ACL injured knee position, the greater PTTSA may lead to increasing risk of ACL injury in females than in males. However, those anatomical gender differences were measured in static and non-weight bearing situations.

In addition, female's greater knee joint laxity is considered as potential risk factor that may contribute the greater ACL incidence rate. Several studies which were related with joint laxity revealed that females showed greater knee anterior and rotational laxity compared with males (Hsu, et al., 2006; Rozzi, et al., 1999; Uhorchak, et al., 2003). However, the passive joint laxity is not enough to understand the gender disparity of the ACL injury rate. Consequently, only the anatomical gender differences could not explain why the ACL incidence rates are difference among males and females. Hence, comprehensive understanding of biomechanical and neuromuscular ACL risk factors is demanded for ACL etiology.

Biomechanical and neuromuscular risk factors

Most ACL injuries occur in dynamic situations; therefore, it is necessary to investigate the biomechanical and neuromuscular ACL risk factors. Gender differences of lower extremity biomechanics including joint kinematic, kinetic and muscle characteristics during the landing phase (stance phase) may contribute to a disparity of incidence of ACL injuries between male and female athletes (Anderson, et al., 2001; Arendt & Dick, 1995; Ireland, 2002; Shelbourne, et al., 1998).

First, several researchers have reported that gender differences of lower extremity (sagittal or frontal plane) were exhibited during landing phase (Decker, et al., 2003; Ford, et al., 2003; Ford, et al., 2005; Kernozek, et al., 2005; Kernozek, et al., 2008; Lephart, et al., 2002; Malinzak, et al., 2001; McLean, et al., 1999; Salci, et al., 2004). According to Decker et al. (2003) and Salci et al. (2004), female groups demonstrated more erect lower extremity landing position (sagittal plane) at initial ground contact (IC) compared with male groups. Also, Ford et al. (Ford, et al., 2003; Ford, et al., 2005) reported that females exhibited greater total valgus knee motion and angles than male athletes and only females showed significant side-to-side differences between their dominant and non-dominant side in maximum valgus knee angle. The discordance of kinematics between males and females may predispose females to non-contact ACL injures. However, the kinematic results from each plane did not conclusively show why females had higher ACL injury incidence rate than males.

On the other hand, a few researchers investigated landing characteristics of gender differences in the frontal and sagittal planes. Malinzak et al. (2001) reported that female performed running, side-cutting, and cross cutting maneuvers showed significantly less knee flexion and more knee valgus during the stance phase. Also, Kernozek et al. **(**2005**)** compared gender differences in frontal and sagittal plane kinematics during drop landings. They reported that males and females exhibited similar joint positions at IC in both planes. However, females exhibited significantly greater ankle dorsiflexion, foot pronation, and knee varus-valgus range of motion (ROM).

In addition, in a prospective study, Hewett et al. (2005a) reported that injured and uninjured female athletes demonstrated different joint kinematics. The ACL injured group exhibited 8.4° greater knee abduction angles at initial ground contact and 7.6° greater at maximum knee abduction angle demonstrated during stance phase. The maximum knee flexion angle was 10.5° less in injured than in uninjured athletes at landing. However, at IC knee flexion angle was not significantly different among between the groups. On the other hand, Ford et al. (2005) and McLean et al. (1999) found no difference in knee flexion angles between males and females. As a result, we see that females generally exhibited less knee flexion and more valgus during the landings or cutting maneuvers. Even though kinematic differences between males and females have been reported, this is insufficient to explain the gender disparity of ACL injury rates.

Second, females' greater vertical and posterior ground reaction forces (GRFs) are considered as one of the biomechanical risk factors that contribute to higher ACL injury rates among females (Hewett, et al., 2005a; Kernozek, et al., 2005; Schmitz, et al., 2007). Kernozek et al. (2005) reported that females exhibited significantly greater peak vertical and posterior ground reaction force than males which may be considered risk factors for sustaining non-contact ACL injuries (Meyer & Haut, 2005). Also, Schmitz et al. (2007), Hewett et al. (2005a), and Lephart et al. (2002) reported that females demonstrated shorter time to peak hip and knee flexion during the landings. According to the impulse-momentum principle, if a force is applied over an interval of time, the applied impulse will result in a change in the momentum of the system. Hence, the shorter time to peak hip and knee flexion results in greater force acting on the system.

In addition, co-activation of two major lower extremity muscle groups, hamstrings and quadriceps, may provide dynamic stability to protect the knee joint. Deficits of lower extremity dynamic neuromuscular control may contribute to the higher ACL injury incidence rate among female athletes (Hewett, et al., 2005b). Most studies which used electromyography (EMG) reported that female athletes exhibited different neuromuscular activation compared with male athletes (Malinzak, et al., 2001; Myer, et al., 2005a; Urabe, et al., 2005; White, et al., 2003; Zazulak, et al., 2005). Those studies reported greater peak quadriceps activity and lower hamstring activity in females than males during landings. The increased quadriceps and decreased hamstrings activation in females may increase anterior tibial translation force and load under dynamic situations because the quadriceps muscles pull the tibia and stresses the ACL at small knee flexion angles. In descriptive laboratory (cadaveric) studies, More et al. (1993) and Withrow et al. (2006) also reported that increased quadriceps force strongly influenced the relative strain on the anteromedial bundle of the ACL.

Training effects of ACL injury

Neuromuscular and dynamic balance training can modify those biomechanical ACL injury factors. In cohort studies, several researchers have reported that plyometric and balance training provide knee stabilization and improve lower extremity alignment, which possibly results in decreased incidence of knee injury in female athletes (Hewett, et al., 1999; Myer, et al., 2005b; Myer, et al., 2005c; Noyes, et al., 2005). Hewett et al. (1999) reported that untrained female athletes had a 3.6 times higher incidence of knee injury than trained female athletes and 4.8 times higher than untrained male athletes. Myer et al. (Myer, et al., 2005b; Myer, et al., 2005c) reported that plyometric training increased IC knee flexion and maximum knee flexion during the drop vertical jump and balance training increased maximum knee flexion during the medial drop landing. Furthermore, the trained female athletes demonstrated significantly increased knee flexion-extension ROM and decreased knee valgus (28%) and varus torques

(38%) during the 6-week interval. In addition, Noyes et al. (2005) reported that neuromuscular-trained female athletes had statistically greater normalized knee and ankle separation distances than males. However, they could not show any results regarding gender issues in training effects because the studies recruited only female athletes or recruited males as a control group.

Biomechanical roles of trunk during drop landing

The relation between ACL injuries and only lower extremity biomechanics has shown limited understanding of ACL injury mechanism. As a result, in the past five years, researchers have begun to investigate biomechanical influences of the proximal knee segment or trunk to the knee joint (Arnett, 2007; Blackburn & Padua, 2008a; Hewett, et al., 2009; Kulas, et al., 2008; Schmitz, et al., 2004; Zazulak, et al., 2007).

Lower extremity biomechanics and trunk position during drop landing

Trunk and/or proximal knee segment may alter alignment of the lower extremity kinematics and kinetics during landing movements (Blackburn & Padua, 2008b, 2009; Hewett, et al., 2009; Kulas, et al., 2008). The trunk segment consists of approximately 35.5 percent of whole body mass, and it contributes GRFs during landing (Kulas, et al., 2008; Lees, 1981). Blackburn & Padua (2008b, 2009) investigated the effects of trunk flexion on lower extremities during drop landing. They reported that the trunk flexed landing induced greater knee and hip flexion angles during the landing (Blackburn & Padua, 2008b). In the follow-up report, the flexed trunk landing showed less vertical and posterior GRFs, also less mean quadriceps' EMG amplitude than preferred landing condition (Blackburn & Padua, 2009). The flexed trunk landing seems to potentially reduce impact and shearing forces imparted to the ACL because greater vertical and posterior GRF and erect landing position are considered as ACL loading factors (Yu, et al., 2006).

On the other hand, Kulas et al. (2008) investigated the effects of increased trunk loading on lower extremity biomechanics during drop landings. They reported that increased trunk loading (10% body weight) induces increasing knee and ankle flexion angles regardless of trunk position adaptations (Kulas, et al., 2008). However, the trunk extensor group showed approximately two times greater percentage of knee extensor and ankle plantarflexor efforts.

Lower extremity biomechanics and unexpected perturbation

Although unexpected external forces (perturbation) are commonly applied to athletes during sport events, there are only a few studies that have investigated the effects of a trunk or hip perturbation on lower extremity biomechanics during landing (Arnett, 2007) or standing (Carcia, et al., 2005; Schmitz, et al., 2004; Shultz, et al., 2006; Shultz, et al., 2000; Shultz, et al., 2001). All of the hip perturbation studies with one-leg standing were used a lower extremity perturbation device (LEPD) to produce either internal rotation (IR) or external rotation (ER) perturbation. Carcia et al. (2005) and Shultz et al. (2001; 2000) reported that order of muscle recruitments was similar between sexes (gastrocnemius-hamstrings-quadriceps). Females, however, have shown faster quadriceps response time compared with males (Carcia, et al., 2005). This could be one of the reasons that females have shown greater ACL injury rates because quadriceps muscles pull the tibia and stress the ACL at small knee flexion.

According to Schmitz et al. (2004), the ER perturbation on the standing leg resulted in significantly greater IR of the tibia on the femur and IR or the femur on the pelvis. Also, the authors reported that the ER perturbation induced significantly increasing knee valgus while the IR perturbation induces a knee varus. However, one leg standing with perturbation is not enough to understand the mechanisms of indirect contact ACL injuries. In addition, the perturbed one leg standing and landing situations are completely different, from a biomechanics perspective. The external force, even when it comes from the same direction, may generate different neuromuscular responses in each condition.

There is only one study, to my knowledge, that has been conducted in which the performer was perturbed during landing phase. Using a customized In-Flight Perturbation Generator, Arnett (2007) applied the perturbations to the COM of the participants (linear perturbation) or slightly above the COM (rotational perturbation). He reported that both linear and rotational perturbation conditions showed significantly less GRF_V compared to the non-perturbed condition. In addition, condition for linear perturbation showed significantly greater peak hip and knee extensor moment and ankle plantarflexor moment compared to the non-perturbed condition.

Summary

Although most researchers have agreed that there is a gender disparity of ACL injury rates and common movements of ACL injuries (a plant-and-cut movement or landing movements), there is no common agreement as to which actual knee movement (knee valgus or knee flexion) or other joints are closely related with the ACL tear or rupture during the injury. Furthermore, most research tasks of ACL injury simply mimicked the injury situation as drop landing or cutting movement. Consequently, there is a need to examine female's lower extremity biomechanical response from the unexpected perturbation during landing in order to better understand this complex phenomenon.

CHAPTER 3

METHODS

Participants

Participants were recruited from volleyball, soccer, and basketball clubs/organizations (e.g., YMCA, YWCO) in the surrounding community and UGA: intramural teams, sports club teams and Department of Kinesiology classes. Seventeen individuals participated in the study. This sample size was selected based on a sample-size estimate (statistical power = 0.8, p =0.05) of one dependent variable (maximum knee flexion angle) using data from a previous in-flight perturbation study (Arnett, 2007).

All participants satisfied the following inclusionary criteria: was a healthy female; age was between 18 – 25 years; had prior and/or current competitive or intramural experience in volleyball, basketball, and/or soccer during high school or college or at higher-level competition for at least two years and played recreationally within the last year; or played competitively within the last 5 years and remained physically active since time of last competitive participation. A potential participant was excluded from this study if a participant had any of the following exclusionary criteria: current or chronic injuries potentially affecting the participant's performance, or safety; previous lower extremity, back, or head injuries requiring major medical attention or surgery; problems with balance not remedied; any illness that would negatively affect performance or safety during drop landing; self-reported any symptoms that would potentially affect the participant's performance or safety, including discomfort, pain, light-headedness, dizziness, fainting, and/or nausea,

Instrumentation

- *High-speed digital video motion measurement system*

  An MX-40 Vicon™ camera system (Vicon, Ltd., Oxford, UK), comprised of seven high-speed (240 Hz), digital-video cameras (visible-red light sensitive C-MOS photodiodes, 4.1 megapixel; exposure time = 1/1000 s) and Workstation™ software, were used to capture the locations of the reflective markers on the participant's lower extremity and trunk.

- *Reflective markers*

  40 retro-reflective markers (14 mm each) were placed on the participant's lower extremity. Markers were placed following modified Lu's model for lower extremity (Lu & O'Connor, 1999; 1998) as shown in Figure 2 and Appendix A.

- *Force platforms*

  Customized drop landing platform (185cm x 251cm) with two Bertec® force platforms (Model: 4060-NC) were used to collect (1,200 Hz) ground reaction forces (GRF) from three directions: anterior-posterior (GRF_{A-P}), medio-lateral (GRF_{M-L}), and vertical (GRF_V).

- *In-flight perturbation generator and accoutrements*

  The 'perturber', that is, the in-flight perturbation generator (Figure 3), was used to generate the perturbation during the flight phase of drop landings. The perturber consisted of a trigger sensor, a perturbation cable, and the main device that created an impulse to the participant. The trigger sensor, (length: 61 cm; diameter: 2.5cm) was located on the superior surface at the middle of the drop bar (length 102cm; diameter 2.8cm) that the participant hung from. To

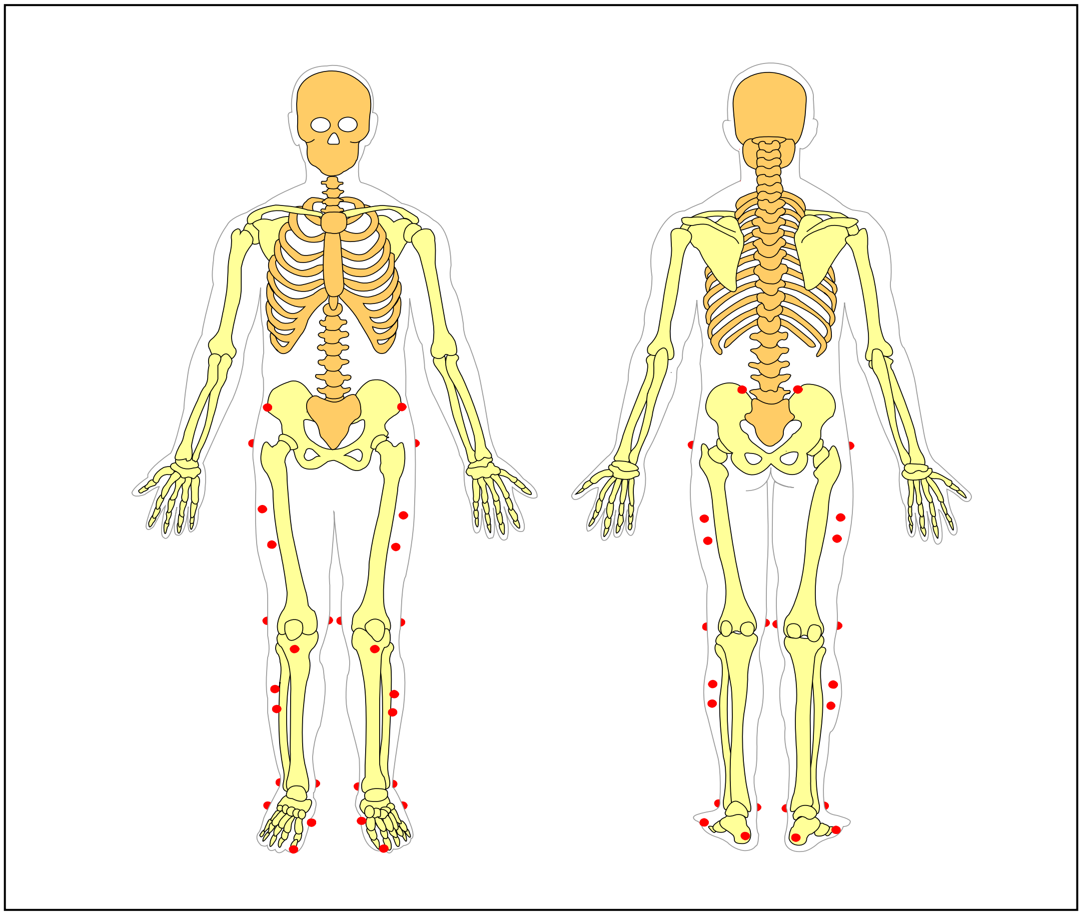

Figure 2 Anterior (Left) and posterior (Right) views of the reflective markers of the lower extremity.

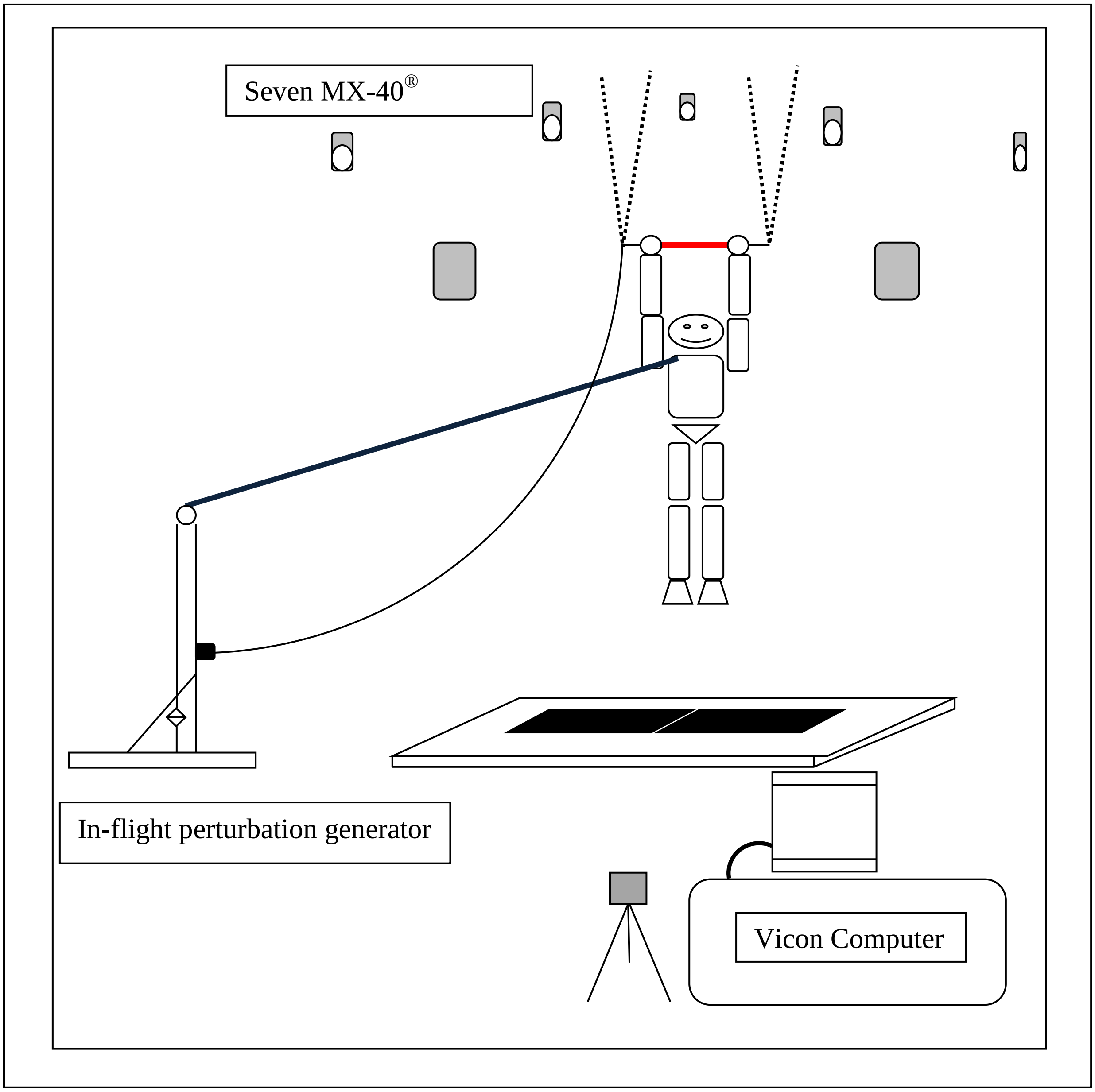

Figure 3. Diagram of the experimental data collection set up. The perturber components are explained in the text. The taut, darker line attached between the participant and the in-flight-perturbation generator is the cable that pulled on the participant in order to apply the perturbation impulse. The lighter, non-taut line shown is the cable that transmitted an electrical signal from the trigger sensor mounted in the top of the drop bar to the perturber to initiate the processes that created the perturbation force.

initiate the perturbation, when the participant began to let go of the drop bar and the grip force dropped below 13.2 N (3 lb), the trigger sensor sent a signal to the main device. The load cell (MLP-200, Transducer Techniques, Temecula, CA) was attached at one end to the harness at the location of one of the participant's acromio-clavicular joint, and the other end to the rubber-coated perturbation cable (length = 4.2 m) that attached to the main device. As the load cell measured tensile strain, the electrical signals from the load cell were later used to confirm the magnitude of the perturbation impulse.

The main device was comprised of, two air regulators (R21-04-000, Wilkerson Co., Pneumatic Division, Richland, MI) air pistons (Speedaire 6W106, Dayton Electric MFG Co., Miles, IL; RLF03A-DAP-NA00, Norgren Inc., Littleton, CO), an air compressor, and weight plates attached to the perturbation cable via a pulley system. To generate the perturbation, pressurized air from the compressor caused the two air pistons, to pull out the pins that supported the weights above the ground. Subsequently, when the weights fell down, the perturbation cable was pulled, thereby transmitting a perturbation force (1.15 times body mass) to the participant.in a lateral direction in a horizontal plane during the flight phase.

- *Anthropometric equipment*

Each body segment's length and circumference were measured using standard anthropometric equipment (digital weight scale, sliding calipers, tape measure, and stadiometer).

<u>Protocol</u>

The protocol consists of three parts: preparation, pre-test, and drop testing procedures.

<u>Preparation procedures</u>

Upon arrival at the Biomechanics Laboratory, the participants were given written and verbal information about the study and testing procedures. Next, the potential participant completed the consent form (Appendix B) prior to completing the Health Status and Physical Activity Questionnaires (Appendix C). The answers were reviewed with the potential participant by the primary investigator to ensure that the participant meets the health- and sport-related inclusionary criteria and not the exclusionary criteria. If the participant does meet the requirements, testing was continued.

The participant worn sport tights or shorts and sleeveless sport top or swim-suit top to minimize marker movement and to ensure the locations of the reflective markers could be captured by the cameras. Also, the participant wore their own court shoes. Next, anthropometric measures was obtained following Zatsiorsky's methods (2002), then the participant was fitted with the body harness. Next, the reflective markers were placed on the participant's skin, clothing, and/or the harness. The participant performed a five-minute warm-up on a stationary bicycle at a self-selected pace.

Pre-test procedures

Prior to starting data collection, a standing static trial was captured. Next, after the participant was taught how to perform the drop landing task, the participant was allowed maximum three practice the drop landing task without perturbations.

Drop test procedures

The participant hang from a bar with both hands (height from the midpoint of the lateral malleolus to the ground = 0.55 m). After the participant received a verbal cue from the researcher, the participant released her hands from the drop bar and land with one foot on each force platform. The participant kept the arms at the initial shoulder position during the entire movement. A trial was determined acceptable if each foot lands only on the correct force platform, and the participant remained stable upon landing for 2 seconds without wobbling.

The participant performed two blocks of testing. The first block of testing, baseline (BASE) was consisted of three acceptable trials of natural drop landing with no expectation of perturbation. Prior to the second block of testing, the perturbation cable was attached to the acromioclavicular joint on the same side of the body as the preferred landing leg. The second block of testing was consisted of three acceptable trials for each of two perturbation conditions: perturbation (PERT) and non-perturbation (NON-PERT), performed in a random order. No more than two trials of the same perturbation condition were performed in subsequent order to reduce feedforward control, anticipatory effects, and development of adaptive strategies (Pavol & Pai, 2002). In addition, the participant had approximately 15 to 20 s of rest between trials to minimize neuromuscular fatigue. Any unacceptable trial was repeated after completing all other trials.

Data Reduction

GRF data reduction

The landing phase was of interest that began at the instant of initial ground contact (IC) and terminated at the first instant when maximum posterior GRF occurred. IC was identified as the instant when the GRF_V magnitude as more than the magnitudes of the five prior data points. GRF data reduction was performed using Visual 3D® software (C-Motion Inc., Germantown, MD). The GRF raw data was filtered using a 15 Hz low-pass, fourth-order Butterworth filter. Peak GRF magnitudes were generated for the maximum peak GRF_V, maximum anterior and posterior GRF_{A-P} values and the maximum medial medial GRF_{M-L}. In addition, time to peak GRF_V (ms) was generated.

Joint kinematic and kinetics data reduction

The three-dimensional coordinates of the reflective markers were generated via a proprietary algorithm (Workstation™, Vicon Inc., Englewood,CA) and smoothed using 15 Hz low-pass fourth order Butterworth filter. All other calculations were performed using Visual 3D® software (C-motion, Inc., Germantown, MD). Joint coordinate systems (Cardan's method) for the ankle, knee, hip, joints of the both legs were used to determine clinical joint angles.

Three-dimensional joint kinetic of the lower extremity of the both legs was calculated for the landing phase using inverse dynamics. Segmental center of mass (COM) locations were determined by Zatsiorsky's method (2002). Body mass and height were used to scale the joint moments.

Statistical Analysis

One sample t-tests were used to test differences among the perturbation conditions ($p < .05$) using mean difference scores between landing conditions. The difference score for each variable was calculated as PERT – BASE. Confidence intervals (CI) at 95% level also were generated.

CHAPTER 4

RESULTS

The characteristics for the 17 participants are shown below (Table 1). All participants performed two blocks of testing (first block: baseline; second block: perturbation). The baseline (BASE) consisted of three acceptable trials of natural drop landing with no expectation of perturbation. The second block of testing consisted of three acceptable trials for each of two perturbation conditions performed in a random order: perturbation (PERT) and non-perturbation (NON-PERT).

Table 1. Participant characteristics

	Mean ± SD	Range
Age (yr)	21.1 ± 1.3	18 - 25
Body Mass (kg)	62 ± 9.9	48.5 - 83.1
Body Height (cm)	166 ± 6	1.55 - 1.76

Frontal plane joint kinematics

The descriptive data of the frontal plane kinematics for the PERT and BASE conditions are located in Figure 4. Table 2 shows the difference-score descriptive, CI and statistical outcomes. Difference-score statistics demonstrated that at IC, the ankle joint was in a 1.5° less inverted position and showed 1.5° less eversion displacement during PERT compared to BASE. At the knee joint, PERT trials resulted in a 1.6° and 2.4° increased IC abduction angle and maximum abduction angle, respectively. However, the CI was not significantly different for the ab/adduction displacement. For the hip joint, at IC, the hip joint was in a more abducted position during PERT compared to BASE landings. A significant difference ($p < 0.000$) existed between the mean difference scores of hip ab/adduction displacement. During PERT landings, participants exhibited hip abduction motion (+) after IC, while during BASE landings, individuals displayed hip adduction (-) motion.

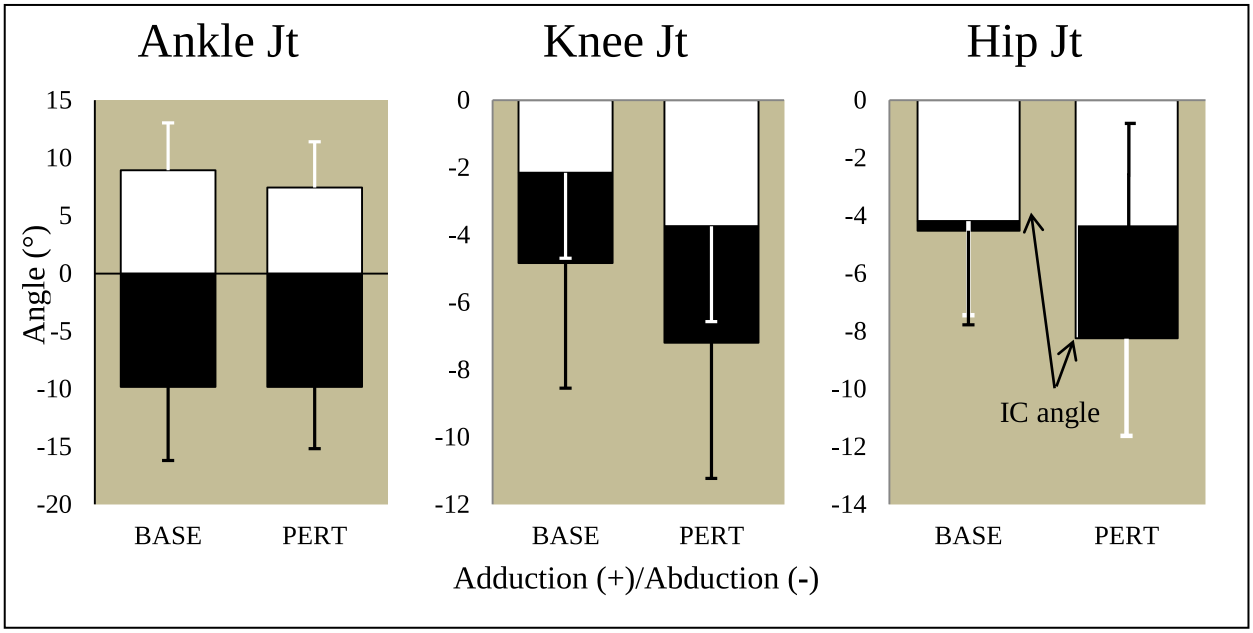

Figure 4. Group means and SD of frontal plane kinematic variables (□ = initial joint angle; ■ = maximum joint angle) for baseline (BASE) and perturbation (PERT) conditions.

Table 2. Means and SD, and the 95 % confidence intervals of difference scores for initial contact (IC) and maximum (MAX) joint angles and joint displacement (DISP) variables. A positive or negative difference score indicates that the value of the perturbation condition was greater or lesser, respectively, than the baseline value.

		Rotation Axis					
Joint	Variables	Add/Abduction (Inv/Everson)			Flex/Extension		
		Mean ± SD	95% CI	*p* value	Mean ± SD	95% CI	*p* value
Hip	IC	**4.1 ± 2.3**	**2.9 - 5.2**	**0.000**	-0.9 ± 2.0	-1.9 - 0.1	0.080
	Max.	-0.2 ± 2.4	-1.4 - 1.1	0.778	**-1.8 ± 3.2**	**-3.4 - -0.1**	**0.037**
	DISP	**4.2 ± 2.1**	**3.2 - 5.3**	**0.000**	-0.9 ± 3.2	-2.5 - 0.7	0.271
Knee	IC	**1.6 ± 1.0**	**1.1 - 2.1**	**0.000**	**-2.1 ± 3.0**	**-3.7 - -0.6**	**0.010**
	Max.	**2.4 ± 2.8**	**0.9 - 3.8**	**0.003**	**-3.1 ± 5.1**	**-5.7 - -0.4**	**0.030**
	DISP	0.8 ± 2.5	-0.5 - 2.1	0.206	-0.9 ± 5.1	-3.6 - 1.7	0.479
Ankle	IC	**-1.5 ± 1.7**	**-2.3 - -0.6**	**0.003**	**3.0 ± 3.3**	**1.3 - 4.3**	**0.002**
	Max.	0.0 ± 2.4	-1.2 - 1.3	0.984	-0.9 ± 2.6	-2.2 - 0.5	0.191
	DISP	**-1.5 ± 2.3**	**-2.7 - 0.3**	**0.018**	2.2 ± 4.8	-0.3 - 4.7	0.077

Note. **Bold** *p* value = a significant t-test comparison ($p < .05$).

Sagittal plane joint kinematics

Means and standard deviations for each landing condition are presented in Figure 5. Group means, standard deviations, the CI and statistical outcomes of the sagittal joint kinematics for the difference scores are located in Table 2. For the IC joint angle, PERT compared to BASE exhibited a 3° greater plantarflexed ankle and 2.1° less flexed knee joint position. In addition, PERT showed a 3.1° less maximum knee flexion and 1.8° less maximum hip flexion position. However, there were no significant differences for the difference scores for flexion/extension displacements about any joint.

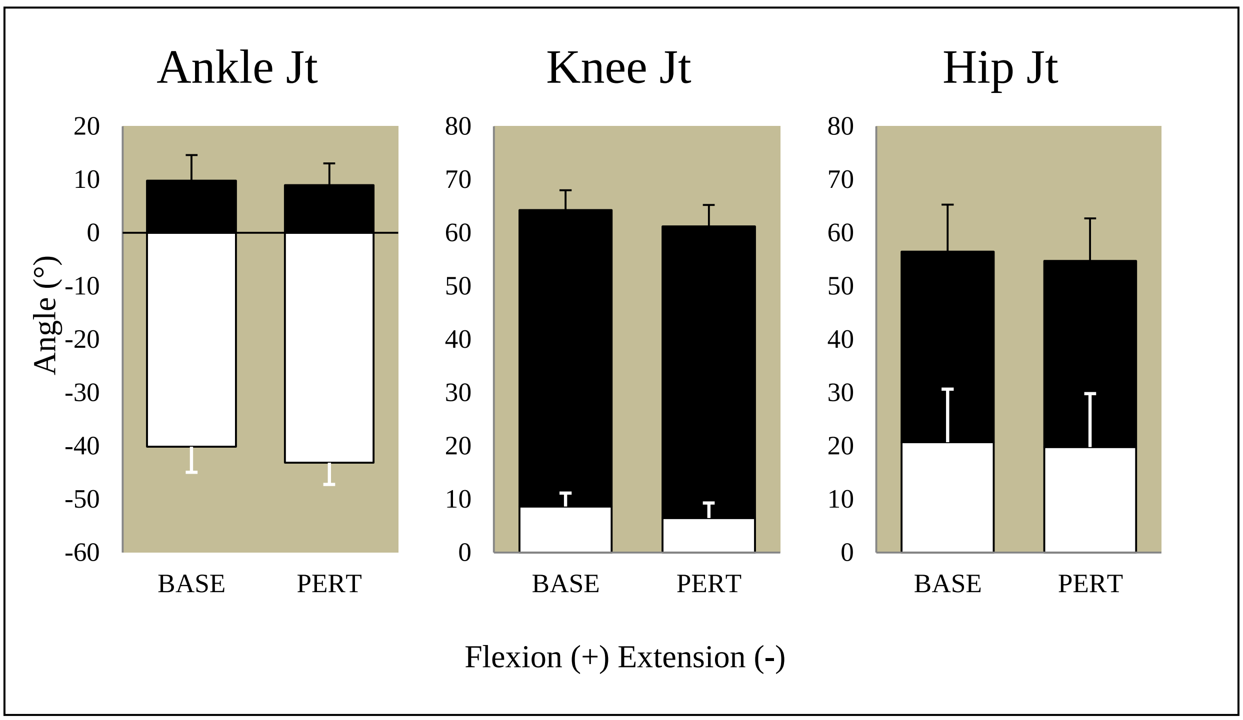

Figure 5. Means and SD for sagittal plane kinematic variables for baseline (BASE) and perturbation conditions (PERT) (□ = initial joint angle; ■ = maximum joint angle).

Ground reaction forces

The ground reaction forces (GRF) values are presented in Figure 6 and means, standard deviations, and confidence intervals for the difference scores are shown in Table 3. Greater peak GRF were exhibited by PERT, compared with BASE. Compared to BASE, PERT demonstrated an increase of 3.8 $N \cdot kg^{-1}$, 0.9 $N \cdot kg^{-}1$ and 0.7 $N \cdot kg^{-}1$ for peak GRF_V, medial GRF and anterior GRF, respectively. There was no significant difference for time to peak GRF. As can be seen bin table 3. There was no consistent effect for the timing of the peak GRF_V among the participants.

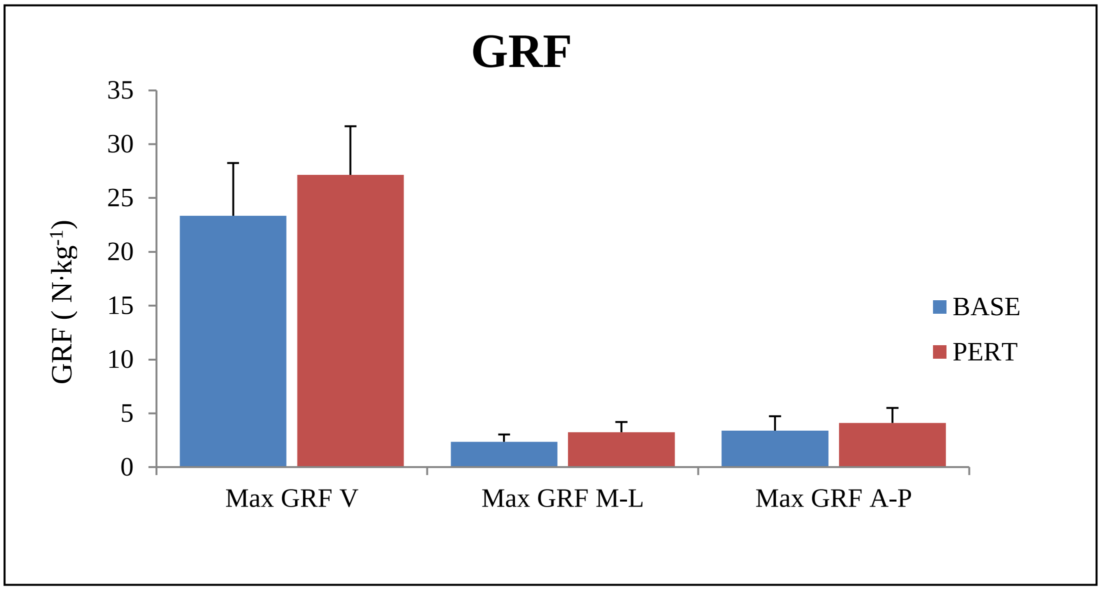

Figure 6. Means and SD of GRF variables. Differences between baseline (BASE) and perturbation (PERT) conditions

Table 3. Mean, SD, and the 95% confidence intervals of group difference scores for GRF variables of maximum (Max.) magnitudes and times to maximum ground reaction forces. A positive or negative difference score indicates that value of the perturbation condition was greater or lesser, respectively, than the value of the baseline.

Variables	Mean ± SD	95% CI	*p* value
GRF_V ($N \cdot kg^{-1}$)	3.80 ± 4.26	1.61 - 5.99	**.002**
GRF_{M-L} ($N \cdot kg^{-1}$)	0.90 ± 0.90	0.42 - 1.37	**.001**
GRF_{A-P} ($N \cdot kg^{-1}$)	0.70 ± 1.20	0.08 - 1.31	**.030**
Time to Max. GRF_V (ms)	0.65 ± 8.80	-3.87 - 5.18	.763

Note. **Bold** *p* value = a significant t-test (p <.05).

Sagittal plane joint kinetics

Figure 7 shows the means and standard deviations for the variables of each landing condition. Group means, standard deviations, and confidence intervals of the difference scores for flexor/extensor joint moments are presented in Table 4. Peak knee extensor and ankle plantarflexor moments for PERT were greater (0.11 $N{\cdot}m{\cdot}(kg{\cdot}m)^{-1}$ and 0.18 $N{\cdot}m{\cdot}(kg{\cdot}m)^{-1}$, respectively) compared to BASE values. The perturbation did not have a consistent effect on hip joint extensor moments among participants, as the 95% CI difference scores ranged from negative (BASE > PERT) to positive (PERT > BASE).

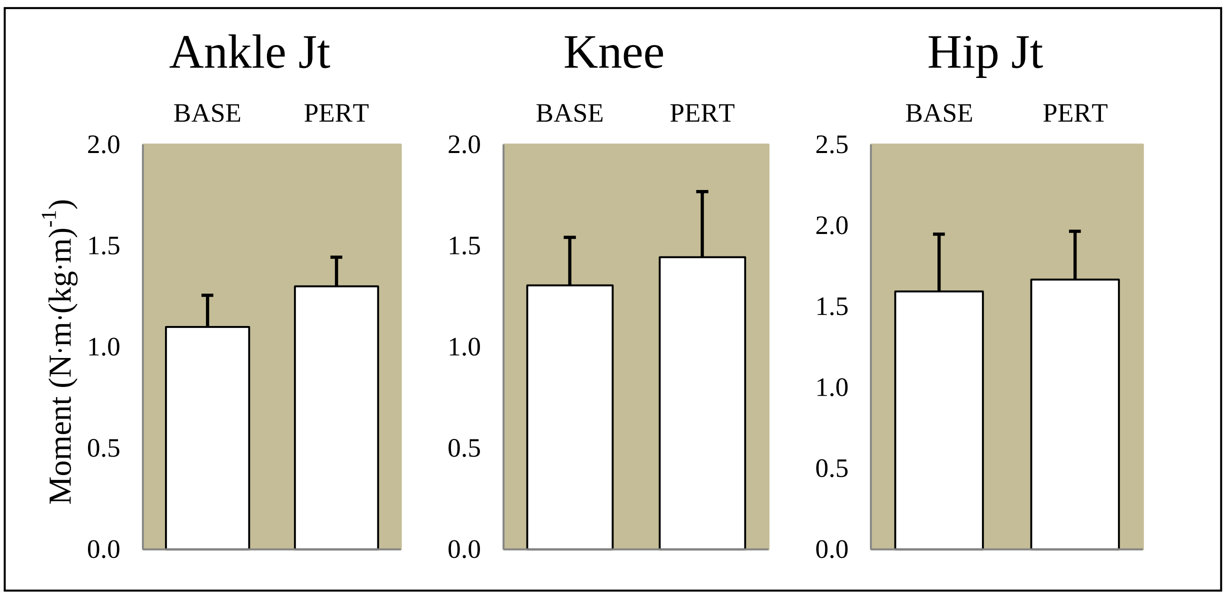

Figure 7. Means and SD for sagittal plane joint moments

Table 4. Mean, SD, and the 95% confidence interval of difference scores for lower extremity flex/extensor joint moments. Variables that displayed significant difference scores (PERT – BASE).

Joint								
Hip			Knee			Ankle		
Mean ± SD	95% CI	*p* value	Mean ± SD	95% CI	*p* value	Mean ± SD	95% CI	*p* value
0.07 ± 0.27	-0.06 - 0.21	0.276	0.14 ± 0.2	0.06 - 0.22	**0.002**	0.19 ± 0.09	0.14 - 0.24	**0.000**

Note. **Bold** *p* value = a significant t-test comparison (*p* <.05).

Frontal plane joint kinetics

Descriptively, as shown in Figure 8 of the ab/adductor joint moments at the hip joint, I qualitatively observed that all participants displayed a pattern from IC to end of the landing phase of adductor-abductor-adductor-abductor joint moments. At the knee joint, although the shape of the moment pattern was similar to the hip joint (i.e., two local maxima and minima), the magnitudes and directions among individuals varied considerably. The only consistently demonstrated moment by all participants at the knee joint was the adductor moment displayed during the initial landing phase. At the ankle joint, all individuals displayed an inversion moment during some portion of the landing phase of both perturbation conditions, but the joint moment pattern otherwise varied widely among participants. Interestingly, although patterns varied among the participants at the knee and ankle joints, for a given individual, the moment patterns for all joints were very similar between BASE and PERT. Therefore, for statistical analyses of the knee and hip joint moments, the first peak adductor moment and the maximum adductor and abductor moment difference scores were chosen. The maximum inversion moment was tested for the ankle joint. Table 5 shows the results for the difference scores of these variables.

Statistically, the difference scores demonstrated that PERT showed 0.01 $N \cdot m \cdot (kg \cdot m)^{-1}$ greater ankle inversion muscle moment compared to BASE. In addition, at the knee joint, the first and maximum adduction joint moments were greater during PERT than BASE (0.04 $N \cdot m \cdot (kg \cdot m)^{-}1$ and 0.1 $N \cdot m \cdot (kg \cdot m)^{-1,}$ respectively). However, for knee abductor moments, the effect of perturbation was inconsistent across participants, as shown by the CI. Moreover, two participants displayed an abductor moment during BASE landings, but not during PERT landings. In addition, only second hip adduction was greater during PERT than BASE (0.08 $N \cdot m \cdot (kg \cdot m)^{-1}$).

Table 5. Mean, SD, and the 95% confidence interval of difference scores for lower extremity abductor/adductor joint moments. Variables that displayed significant difference scores (PERT – BASE).

Joint	Variables	Mean ± SD	95% CI	*p* value
Hip	ADD	0.03 ± 0.14	-0.04 - 0.10	0.346
	ABD	0.02 ± 0.17	-0.06 - 0.11	0.614
	ADD	0.08 ± 0.01	0.00 - 0.16	**0.045**
Knee	ADD	0.04 ± 0.04	0.02 - 0.06	**0.001**
	ABD	-0.05 ± 0.10	-0.11 - 0.00	0.068
	ADD	0.10 ± 0.08	0.02 - 0.06	**0.000**
Ankle	INV	0.01 ± 0.01	0.00 - 0.02	**0.008**

Note. **Bold** *p* value = a significant t-test comparison (p <.05). (ADD:adductor; ABD: abductor; INV: inversion moment)

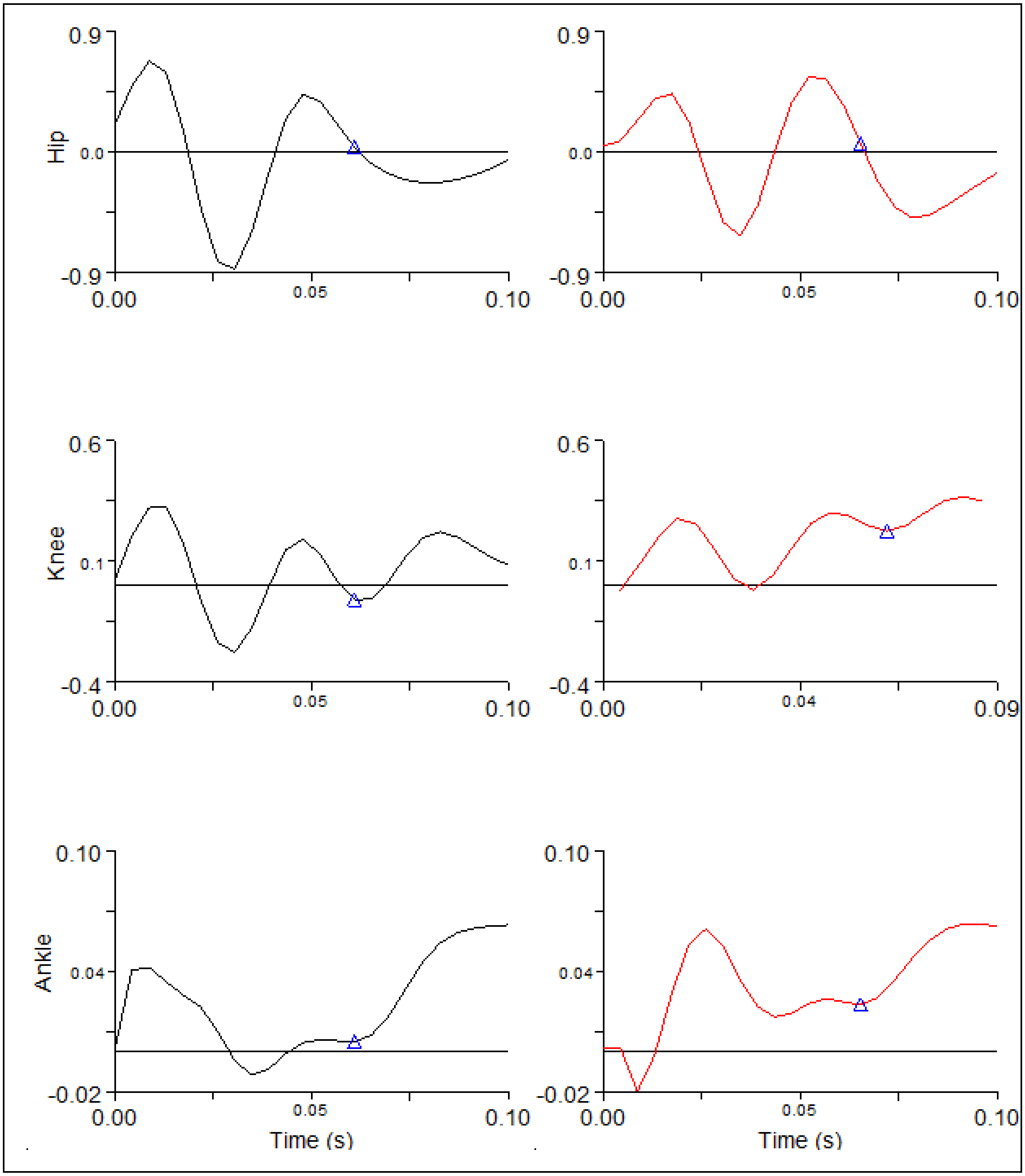

Figure 8. Representative hip and knee abductor/adductor patterns for Baseline (left) and Perturbation (PERT). The triangle represents the instant when the vertical GRF (GRFv) achieved its maximum value.

CHAPTER 5

DISCUSSION

The primary purpose of this study was to determine the lower extremity biomechanical responses exhibited by females during PERT compared to BASE landings. In contrast to a typical drop landing, the laterally-directed perturbation was expected to increase the angular and linear momenta in the frontal plane, and as such, produce different landing angles and increased angular displacements, joint moments and GRF (GRF_M) in the frontal plane. In addition, we also anticipated decreased lower extremity joint flexion displacements and, therefore, increased lower extremity net muscle moments compared to BASE landings. Less lower extremity flexion to attenuate GRF was expected, consequently, to increase the GRF_V and decrease the time to peak GRF_V. The results of this study mostly support the hypotheses, and some of the anticipated justifications.

Frontal plane joint kinematics

For the IC angle of the lower extremity, it was anticipated that PERT, compared to BASE, would induce greater ankle inversion, knee abduction and hip adduction IC angles. Support for our predications for the frontal plane IC angle was mixed. At IC during PERT versus BASE landings, the ankle, unexpectedly, showed a more neutral rather than a more inversed position. However, the knee joint was in a greater abducted position as expected, but the hip joint was significantly greater abducted instead of an adducted angle at IC. One explanation is that the in-flight perturbation pulled the upper trunk laterally without having affected the lower extremity due to inertial lag. Consequently, the more lateral position of the upper trunk would create the increased hip abduction angle, not the upper leg. Furthermore, due to the mass of the trunk shifting laterally during the perturbation, the center of the mass of the body (COM_{BODY}) also would shift laterally, if no other body mass shifted in compensation. In order for the person to land in a stable position, the leg of interest may have abducted slightly to ensure that the leg of interest was under the more laterally-located COM_{BODY} at IC. However, at present, without further analysis of the data, this explanation cannot be confirmed.

For the frontal plane joint displacements, our hypotheses also were partially upheld. As expected, there was significantly greater displacement about the hip joint due to the perturbation, but not at the ankle and knee joints. Interestingly, the hip joint showed opposite motions during the two landing conditions. During PERT landings, hip adduction motion was exhibited, but abduction motion occurred during BASE. One possible explanation for this finding relates to the lateral trunk flexion. If the trunk laterally flexed due to the perturbation, then the participant would have to adduct the trunk and pelvis to regain an upright position in the frontal plane. A second, explanation is that if the hip joint was in a more abducted position at IC to ensure a stable base of support, then hip abduction during the subsequent landing phase would have to occur to allow the person to then attain a more upright femoral position. (As the segmental data are not available currently, these explanations are not testable at this time.)

At the knee joint, no significant differences existed for potentially two reasons. First, the perturbation condition exhibited greater abducted positions from IC throughout the landings phase, therefore, the displacement didn’t increase. Second, as greater knee adductor moments were generated during PERT landing phase, it is suggested that the adductor muscles and other involved structures controlled the amount of abduction motion during the landing phase. Third, the perturbation did not have a consistent effect among all participants. As can be seen by the 95%

CI of the difference score for knee abduction displacement, some participants decreased or did not display any difference between the landing conditions.

Increased knee abduction angles exhibited throughout the landing and greater hip adduction motion exhibited during the PERT versus the BASE condition suggest that a lateral perturbation PERT could potentially increase the risk of an ACL injury to the leg of interest. A more abducted knee joint at touchdown and throughout the landing phase is considered to be an ACL injury risk factor because abduction may increase the strain and place a greater stress on parts of the ACL (Bendjaballah, et al., 1997; Ford, et al., 2003; Hewett, et al., 2005a; Malinzak, et al., 2001). In addition, Hewett, et al. (Hewett, et al., 2009) confirmed this relationship between frontal joint movements and ACL injury for females analyzing 17 (7 male and 10 female ACL injured players) ACL injury cases that occurred during competition and captured on video. For most of the females who suffered an ACL injury, greater lateral trunk and knee abduction angles were observed compared to the ACL-injured males.

Sagittal plane joint kinematics

Fort the sagittal plane joint kinematics, we expected joint displacements would be less during PERT as opposed to BASE landings, due to the amount of flexion that would occur during the landing, not the angle at IC. Decreased displacements were surmised as a strategy to prevent abnormal frontal plane joint alignments and motions due to the perturbation. However, none of our predictions of the sagittal joint kinematics were supported.

First, the IC results were unanticipated. Compared to BASE, PERT landings exhibited more extended ankle, knee and hip joint positions. We did not expect any change of sagittal plane IC angles between two landing conditions because we assumed that the lateral in-flight perturbation would affect only frontal plane joint kinematics until IC. However, one possible reason for a more extended landed position is that the participant may subconsciously stiffen the joints in anticipation of the perturbation. Future comparisons of the SHAM and BASE trials will allow us to test this notion of feedforward anticipatory effects.

The extended knee landing position, however, has potential behavioral consequences during perturbation situations in which the player anticipates the perturbation. First, an extended knee joint position at IC of landing movements is considered as a risk factor for ACL injuries (Krosshaug, et al., 2007; Olsen, et al., 2004). This may be due to the anterior shear force that acts on the proximal tibia, as it is highest when the knee joint is in an extended landing position, according to Nunley, et al, (2003).

A potential positive consequence of a more extended lower extremity at IC is that it can allow for greater flexion displacement during the rest of the landing phase, which is considered to be an effective part of a shock-attenuation strategy (Decker, et al., 2003; Devita & Skelly, 1992). However, the more extended IC angles appeared not to be associated with altered flexion displacements, as no displacements were significantly different.

Ground reaction forces

Our expectations that PERT, compared to BASE, would elicit greater peak GRFv and GRF_M, were supported. Our results indicated that the PERT condition did create increased peak GRFs in the antero-posterior direction, too. Peak GRF_{M-L} was increased by 9% body weight during PERT compared to BASE landings. For the medio-lateral direction, one potential rationale was our original explanation. Increased lateral momentum of the body, including the leg of interest (i.e., the leg ipsilateral to the side of the perturbation) would cause the need for greater opposing medial GRF impulse to be applied to that leg. Moreover, the abnormal lower extremity alignments in the frontal plane during PERT would generate increased GRF_M when the person pushed laterally against the ground to return the laterally-shifted COM_{BODY} and the lower extremity back to a more neutral position.

In the vertical direction, for peak GRF_V, PERT elicited 39% greater force relative to body weight compared to BASE. One likely explanation is that participants did not use greater lower extremity flexion during PERT landings than typical landings that would attenuate the impact forces, but did increase the joint moments that would result in pushing downward against the ground with greater force (Podraza & White, 2010)

Behaviorally, higher peak GRF_V is considered a biomechanical risk factor for ACL injury, particularly for females (Hewett, et al., 2005a; Kernozek, et al., 2005). Kernozek and Ragan (2008), using inverse dynamics data and ACL-related cadaver data to model ACL loading, reported that GRF_V play an important role in affecting the amount of force and strain placed on the ACL.

Sagittal plane joint kinetics

It was anticipated that increased extensor lower extremity net muscle moments during PERT would be displayed compared to BASE, due to the gained linear and angular momenta from the in-flight perturbation that would need to be counteracted and reduced during the landing phase. The results for the ankle and knee joints supported our suppositions. Compared to BASE, PERT showed greater peak eccentric ankle plantarflexor and knee extensor moments during the landing phase. These moments increased during PERT landings potentially to control the downward momentum without increasing lower extremity flexion.

Our results of the extensor joint moments were similar to Arnett (2007). He investigated the biomechanical effects of a somersault axis in-flight perturbation for drop landings. It was, to my knowledge, the first study associated with an in-flight perturbation during drop landing. He reported that perturbation condition showed greater lower extremity extensor moments compared control group. Participants in his study used lower extremity moments, however, to reduce the somersault momentum.

Although we cannot directly measure ACL forces, thus, elucidate ACL injury mechanisms with confidence, the increased knee extensor net muscle moment demonstrated by the PERT condition could suggest increased ACL loading based on prior studies. If the increase in knee extensor moment was due, in part, to greater quadriceps muscle moment, then greater anterior tibial shear force relative to the femur due to extended landing position at IC is implied that may increase ACL loading and strain (Nunley, et al., 2003; Yu & Garrett, 2007). Conversely, if the increased knee moment was due, in part, to reduced knee flexor moment (implying less hamstrings force), then the ACL loading and strain may increase due to a reduced contribution of the hamstrings to resist anterior tibial-posterior femoral shear forces.

<u>Frontal plane joint kinetics</u>

Before interpreting the quantitative outcomes, some descriptive observations are worth noting. As the shape of the moment patterns at all three joints were similar between PERT and BASE for a given person, this suggests that using difference scores to determine how perturbations affect peak moment values was an appropriate analysis strategy.

We had anticipated that individual participant variability would be displayed for hip and knee ab/adductor joint moment patterns (Bates, 1996; James, et al., 2003), but not for the ankle in/eversion patterns. Therefore, we were surprised to find that among the lower extremity joints, the hip joint displayed the most consistent patterns among participants. All participants displayed hip adductor, abductor, adductor, then abductor moments from IC to the end of the landing phase. The individual participant variation observed in this study supports Hewett (2005a) finding that not all female athletes land similarly. They also observed that females that later suffer an ACL injury may display different landing biomechanics compared to their non-injured female counterparts.

Also of interest was that the shape of the knee joint pattern often was similar to that of the hip joint, for a given person. Moreover, the times to the first peak adductor moment at the hip and knee joints were closely linked.

Quantitatively, It was surmised that the adductor joint moment acting about the anterior-posterior axis of the knee joint would increase during PERT compared to BASE landings. The basic assumption for this prediction is that the abnormal knee abduction alignment at IC due to the in-flight perturbation would require greater knee adductor net muscle moments during landing phase. Our results confirmed our hypothesis. Greater peak knee adductor moments (first and maximum) and hip adductor moment (maximum) were displayed for PERT compared to BASE, but not maximum abductor moments.

One explanation for greater peak knee adductor joint moments also is supported by our frontal and sagittal plane joint kinematic outcomes. By limiting hip and knee sagittal plane motion, greater knee valgus (knee abduction) motion and knee adductor joint moments occur (Pollard, et al., 2010).

Behaviorally, increased knee adductor joint moments may be an adverse consequence of a lateral perturbation. Conversely, increasing adductor moments at the knee joint when the body has more lateral, linear as well as angular momenta due to this perturbation perhaps is beneficial to stabilize the knee joint in the frontal plane. Moreover, by creating greater knee adductor joint moments during PERT landings, greater knee abduction motion was prevented.

CHAPTER 6

SUMMARY, CONCLUSIONS, RECOMMENDATIONS

SUMMARY

The overall purpose of the study was to determine the effects of lateral flight-phase perturbations on the landing phase biomechanics of female athletes who performed drop landings. The in-flight perturbation created linear momentum of the body in the lateral direction as well as a lateral flexion torque to be applied to the trunk. The specific purpose of study was to determine if applying an unexpected perturbation during the flight phase resulted in altered lower extremity biomechanics compared to the biomechanics displayed during a typical landing.

Seventeen female collegiate recreational athletes voluntarily participated in this study. Participants performed two blocks of testing. The first block (baseline = BASE) of the testing consisted of three acceptable drop landing trials with no perturbation nor any expectation of a perturbation. The second block of testing was also consisted of three acceptable trials for each of two perturbation conditions, perturbation (BASE) and nonperturbation (SHAM), performed in a random order. The perturbation was a lateral horizontal pulling force (1.15 x body mass) applied to the acromioclavicular joint on the side ipsilateral to the dominant leg.

During drop landing trials, the spatial locations of the reflective markers were captured and ground reaction force signals obtained using a 3D motion capture and analysis system. Due to variability among individual participants for joint kinematics and kinetics in the frontal plane, difference scores (PERT – BASE value) for each person for all variables were generated, One-sample t-tests of the 95% confidence intervals for the difference scores were compared to the null hypothesis CI to determine whether differences between the PERT and BASE values were significant.

The results of this study demonstrated that the peak magnitudes of the GRF_V and GRF_M were increased by the in-flight perturbation as predicted. Furthermore, participants landed in a more extended position while knee was abducted and hip adducted at IC. The IC and peak joint angles were more likely affected than joint displacements. Lack of displacement differences was due to differing participant responses to the perturbation and/or perhaps an attempt to minimize excessive ab/adduction motions.

In addition, knee extensor and adductor net muscle moments were significantly increased due to the in-flight perturbation. Increased moments may have positive and negative consequences associated with ACL injury mechanism.

CONCLUSIONS

Based on our results, lateral in-flight perturbations affect both frontal plane and sagittal plane lower-extremity biomechanics. Behaviorally, these outcomes thus suggest how female athletes typically negotiate safe landings after a semi-anticipated perturbation, such as a push or a bump sideways by an opponent while in the air. The ability to generate sufficient, eccentric extensor and adductor moments about the lower extremity joints is of primary importance.

Simultaneously, potential mechanisms for a soft-tissue injury during landing after a lateral perturbation have begun to be identified via our findings. A more extended, abducted knee, and adducted hip at initial ground contact, greater vertical impact and medio-lateral GRF and increased eccentric knee joint moments also suggest increased injury risk.

Our results, to the best of our knowledge, are unique. At present, only one study has been conducted in which the performer was perturbed during the flight phase (Arnett, 2007). Arnett perturbed individuals anteriorly, whereas we perturbed people in a lateral direction. The landing strategies required to maintain stability and land safely after a lateral compared to an anterior-directed perturbation, however, are different and place greater demands on muscles and other structures that help stabilize the joints in the frontal plane.

RECOMMENDATIONS

There are five recommendations:

1) After a lateral, in-flight perturbation, during the landing phase, one leg may bear more of the vertical and medio-lateral GRF loads than the other leg. Therefore, future studies should be conducted to investigate limb differences in order to see whether one limb is at higher risk of injury than the other.

2) A greater understanding of the role of the trunk in affecting the responses of the lower extremity after unexpected in-flight perturbations from any direction also are needed.

3) Estimating the strains and stresses of the ACL for studies similar to this one will confirm whether lateral perturbations truly do increase the risk of ACL injury or not.

4) While using a similar perturbation protocol as that used in this study, also including a psychological distraction, such as requiring the participant to make an instant decision as to the direction to move the body during the landing phase, will provide more detailed knowledge about participant landing strategies in a an even more realistic context.

5) Based on the observations of varying individual participant landing biomechanics, analyzing the data after separating the participants into sub-

groups, such as knee abductors versus knee adductor participants, will be helpful in elucidating the biomechanical consequences of different landing strategies.

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APPENDIX A

DESCRIPTIONS AND LOCATIONS OF THE MARKER-SET

Body segment	Label	Descriptions
Pelvis	LASI	Left ASIS
	RASI	Right ASIS
	LPSI	Left PSIS
	RPSI	Right PSIS
Left femur	LTRO	Left greater trocanter
	LTHI1	Left thigh tracking marker1
	LTHI2	Left thigh tracking marker2
	LTHI3	Left thigh tracking marker3
	LTHI4	Left thigh tracking marker4
	LLFC	Left lateral femoral epicondyle
	LMFC	Left medial femoral epicondyle
Right femur	RTRO	Right greater trocanter
	RTHI1	Right thigh tracking marker1
	RTHI2	Right thigh tracking marker2
	RTHI3	Right thigh tracking marker3
	RTHI4	Right thigh tracking marker4
	RLFC	Right lateral femoral epicondyle
	RMFC	Right medial femoral epicondyle
Left tibia	LTT	Left tibial tuberosity
	LSK1	Left shank tracking marker1
	LSK2	Left shank tracking marker2

	LSK3	Left shank tracking marker3
	LSK4	Left shank tracking marker4
	LLMA	Left lateral malleolus
	LMMA	Left medial malleolus
Right tibia	RTT	Right tibial tuberosity
	RSK1	Right shank tracking marker1
	RSK2	Right shank tracking marker2
	RSK3	Right shank tracking marker3
	RSK4	Right shank tracking marker4
	RLMA	Right lateral malleolus
	RMMA	Right medial malleolus
Left foot	LHEE	Left heel
	LFMT	Left fifth metatarsal
	LNTC	Left Navicular tubercle
	LTOE	Left middle foot of 3rd distal metatarsal
Right foot	RHEE	Right heel
	RFMT	Right fifth metatarsal
	RNTC	Right Navicular tubercle
	RTOE	Right middle foot of 3rd distal metatarsal

APPENDIX B

CONSENT FORM

INFORMED CONSENT FOR RESEARCH PARTICIPATION

I, ________________________ agree to participate in the research study entitled, ***"The effects of rotational in-flight perturbation on lower extremity biomechanics during drop landing"***, *w*hich is being conducted by Dr. Kathy Simpson (706/542-4385) and Mr. Jae Pom Yom (542-4132), Department of Kinesiology at the University of Georgia, Athens, GA. I understand that my participation is entirely voluntary. I can refuse to take part in this study, and I can stop taking part at any time without giving any reason, and without penalty or loss of benefits to which I am otherwise entitled. If I withdraw my consent at any time, I can have the results of the participation, to the extent that it can be identified as, mine, returned to me, removed from the research records, and destroyed.

We are interested in learning how athletes perform landing movements when performing a movement that requires them to jump into the air. We are particularly interested in the movements and physics (biomechanics) that occur to the athlete when contact with another person in the air occurs (e.g., athlete gets bumped by another player during a basketball rebound). Therefore, the **objective of this study** is to understand the biomechanics exhibited when a 'perturbation' (i.e., a small pull) is applied to people during the flight phase of a drop landing compared to a typical drop landing. The drop landing is a movement in which a person hangs from a bar suspended above the ground, then drop down onto the ground and lands.

I may benefit by developing a greater understanding of the factors that affect landing performance and injury and how these factors may influence my own landings. The benefit to society is that we will learn how people land safely during contact situations in sport. This initial step will lead us towards better understanding how knee injuries, including ACL injury, can occur under similar circumstances. Consequently, we can devise more effective injury prevention programs.

My part in this study will last for approximately 90 min.

The procedures are as follows: after I, the participant, sign this consent form, I will be given a confidential questionnaire regarding my current health status and history of injuries or other medical conditions and history of sport participation and current physical activities I am engaged in. After the researchers review my answers, if I am ineligible to continue participating in this study, I can have all my forms returned to me or have all my records removed and destroyed by the researchers.

If the researchers ascertain that I am healthy and have no symptoms that would compromise my performances and/or safety, I will continue participation. I will be asked to demonstrate my walking, standing on one leg, and kicking a ball. Certain measurements of my body dimensions, e.g., height, weight, leg length, etc. will be made. I will have a safety harness placed on my trunk, which will be used to later for the drop landing performances. Next, similar to techniques used in animation, reflective markers will be attached to my body to track the movements of the reflective markers via special video cameras during my drop landing performances. I will undergo a warm-up consisting of 5 min stationary bicycling. The perturbation cable that will apply the pulling force and a safety cable to prevent me from falling during the drop landings will be attached to the harness. For the drop landing task, I will hang from a bar (22" from the ground to my ankle), then drop and land with my feet on force platforms (devices that measures forces I apply to the ground during landing). I will practice the task several times. For data collection, I will perform the drop landing approximately 15- 20 times while also being videotaped with a regular video camera. After the first several drop landings, for subsequent drop landings, the perturbation cable may or may not apply a pulling force to me when I am in the air.

At any time before or during testing, I know that I am to let any researcher know immediately if I feel discomfort or pain, or experience other symptoms that could also affect my safety, for example: trouble with my balance, dizziness, nausea, excessively hot, achy body, etc. Testing will stop immediately. If the problem is minor and resolvable, if I feel that the problem is resolved, and I feel no further signs or symptoms, I can choose to continue participation, postpone completing the rest of the tasks until later, or withdraw from the study. The researchers also reserve the right to stop testing temporarily or permanently, or ask that I provide medical clearance before I can continue participating.

Upon completion of testing, I will be given a $10 gift certificate from Target (or a comparable store). If I want to receive extra credit in a KINS class that the instructor[1] has approved for participation, I am responsible for knowing/following the instructor's policies and procedures for obtaining extra credit. (For example, my instructor may only allow extra credit if not receiving payment or equivalent goods, hence, I would have to choose between receiving extra credit (amount of extra credit = .02%/hr of participation, rounded off to the nearest ½ hr) or the gift certificate). If I would rather not participate in this research, there will be an equivalent alternative to earn the extra credit which I should discuss with my class instructor. If I withdraw from the study without having performed any drop landing trials, I will not be given a gift certificate or extra credit. In addition, if I am asked to withdraw during the study due to not having provided complete or accurate information on the Health Status and Physical Activity questionnaire as would be reasonably expected, then I may become ineligible for receiving a gift certificate or extra credit.

Minimal risks are foreseen because the task is very similar to movements I have performed without injury many times, such as blocking in volleyball or basketball rebounding; the task is in a safer setting than in an actual practice or competition situation, and the number of times I will do the drop landing task is much fewer than if participated in a practice drill involving jumping/landing in volleyball and/or basketball. The amount of perturbation is low; it is less than what is experienced during basketball. It is equivalent to standing and being pushed gently with enough force to cause me to start walking or sidestepping at a natural pace (slightly greater than 3 mph). However, slight muscle discomfort or soreness in my legs or shoulders may occur for a few days after participation, particularly if I have not been performing physical activity involving landings recently on a regular basis.

The researchers will exercise all reasonable care to protect me from harm as a result of my participation. In the event of an injury as an immediate and direct result of my participation, the researchers sole responsibility is to provide immediate, emergency care, and arranging for transportation to an appropriate facility if additional care is needed. I will not receive any financial assistance for additional medical or other costs. As a participant, I do not relinquish or waive any of my legal rights.

The only people who will know that I am a research participant are members of the research team. No identifying information about me or provided by me during the research will be shared with others, except if necessary to protect my rights or welfare (for example, if I am injured and need emergency care); or if required by law. All of my individually-identifiable data files and information will be confidential, identifiable only by a participant ID code that is known only to the researchers and stored in a secure area. Only the researchers will have access to the data. Electronic data files will be protected via computer and electronic file security methods. As only the locations of the reflective markers are visible from the file generated from the special video cameras, I cannot be identified by someone viewing these files. For the regular video files, these files will only be viewed by the researchers if necessary to confirm my data are correct. I am welcome to view my video and other available files, and to have feedback provided to me about my landing technique upon completion of the tasks. The regular video files will be destroyed as soon as the researchers complete analyzing my data or 3/30/2013, whichever comes first. The rest of my data will be identifiable by my participant ID code until 3/30/2015. At that time, my data will have the participant ID code removed so it will be unidentifiable.

For any further questions about the research, please contact: Co-investigator, Jae Pom Yom at 706.542-4132 or steve76@uga.edu or the primary investigator, Dr. Kathy Simpson at 706.542-4385 or kjsimpsonuga@gmail.com.

I understand the procedures described above. My questions have been answered to my satisfaction, and I agree to participate in this study. I have been given a copy of this form.

Please sign both copies of this form. Keep one and return the other to the investigator.

____   ______/______/______

Signature of Participant   Date

____   ______/______/______

Signature of Researcher(s)   Date

Additional questions or problems regarding your rights as a research participant should be addressed to The Chairperson, Institutional Review Board, University of Georgia, 629 Boyd Graduate Studies Research Center, Athens, Georgia 30602-7411; Telephone (706) 542-3199; E-Mail Address ***IRB@uga.edu***

APPENDIX C

HEALTH STATUS AND PHYSICAL ACTIVITY QUESTIONNAIRES

Health Status and Physical Activity Questionnaire

For researcher's use only
PP# __________
Date _________

The purpose of this questionnaire is to help us assess your past medical history and current health status to ensure eligibility of participation, and that you have no current or past conditions that would affect your performance today. Second, we are gathering information about your prior and current participation in selected sports and other physical activities.

Please ask the researcher if you have any questions or need assistance. Your participation is greatly appreciated!
Age: _____ yr Gender: (Place an X in appropriate blank) ____ Female ____ Male

MEDICAL HISTORY AND CURRENT HEALTH STATUS

<u>Medical History</u>

- X out the "Y" (yes) or "N" (no).
- If more room is needed to answer a question, continue answer on back of page.)

1. Have you ever had any injuries to your <u>ankle(s)</u>? Y/N
   If yes, list each injury, when it occurred and whether medical attention was required.
   ____

2. Have you ever had any injuries to your <u>knee(s)</u>? Y/N
   If yes, list each injury, when it occurred and whether medical attention was required.
   ____
3. Have you ever had any injuries to your hip(s)? Y/N
   If yes, list each injury and when it occurred. ______________________________
   ____
4. Have you ever had any injuries to your head? Y/N
   If yes, list each injury and when it occurred. ______________________________
   ____
5. Have you ever had any broken bones? Y/N
   If yes, list each broken bone and when it occurred. ________________________
   ____
6. Have you ever had any back or spine injuries? Y/N
   If yes, list each injury and when it occurred. ______________________________
   ____

7. Have you ever had any major surgeries? Y/N
   If yes, list each surgery and when it occurred. ______________________________
   ____
8. Have you experienced chronic, severe or recurring inner ear problems (e.g., recurrent infections)? Y/N
   If yes, list each problem and when it occurred. ______________________________
   ____
9. Have you experienced chronic or severe dizziness, problems with balance, and/or excessive clumsiness within the last year? Y/N
   If yes, please explain. ____
   ____
10. Do you have any medical-related problems are not listed above? Y/N
    If yes, list each condition. ____
    ____

Current Health Status

1. If you have any of the following symptoms, place a check in the blank provided.

________ pain	________ dizziness	________ trouble with balance
________ muscle soreness	________ coordination difficulties	________ vison-related problem
________ hearing-related problem	________ inability to concentrate	________ tired

2. Are you currently ill? Y/N
   If yes, please explain.

3. How much sleep did you get the night before last? ___ hr last night? ___________ hr
4. Are you currently taking any prescription or over-the-counter medications? Y/N
   If yes to above, are any of these medications being used to control pain, physical discomfort, dizziness, or balance problem? Y/N

PHYSICAL ACTIVITY:

Soccer, volleyball and/or basketball experience

1. Please check any and all levels of participation that you engaged in for competitive soccer, volleyball and/or basketball.
   ___ high school varsity ___college intramural ___college varsity ___recreational
   ___club
2. How long have you participated in competitive soccer/volleyball/basketball?
   ___years ___months
3. How long has it been since you last participated on a regular basis in soccer/volleyball/basketball? If still currently participating on a regular basis, enter "0."

Soccer: ___years ___months
Volleyball: ___years ___months
Basketball: ___years ___months

4. How often do you practice every week/month?

___1 day ___2 days ___3 days ___4 or more days/week
___1 day ___2 days ____3 days /month

5. How long is the average duration of each practice?
___1 hour ___2 hours ___3 or more hours

6. Please fill out the following chart with your current level of physical activity (i.e., how often you workout each week) and the type of physical activity you engage in.

Activity	Level of activity each week			
	0-1 hours/week	1-2 hours/week	2-3 hours/week	4+ hours/week
soccer/basketball/volleyball if you are not on a competitive team at present?				
aerobic-related (running, swimming, cycling, spinning etc.)				
yoga/Pilates				
weight lifting				
Outdoor (e.g., hiking, kayaking, rockclimbing, orienteering, geocaching) other (please list) Martial arts dance gymnastics court sports (e.g., tennis, squash, racquetball) Other				

APPENDIX D

REPRESENTATIVE LOWER EXTREMITY JOINT ANGLE GRPAHS