THE EFFECT OF IN-FLIGHT PERTURBATIONS ON LANDING BIOMECHANICS

by

SCOTT W. ARNETT

(Under the Direction of KATHY J. SIMPSON)

ABSTRACT

The exact mechanisms occurring at the time of an ACL injury are still not known. The focus of recent ACL injury research has been extended to include the interaction of biomechanical and environmental factors. One such interaction is an in-flight perturbation. Although these events have been implicated in ACL injuries, the effects of these events on drop landing biomechanics were previously not known. Therefore, the purpose of the study was to determine the effect of a linear and rotational in-flight perturbation on landing biomechanics.

Twenty five college-aged female soccer and basketball athletes performed drop landings with and without in-flight perturbations. Three dimensional ground reaction forces and lower extremity joint kinematics and kinetics were analyzed. Paired t-tests were used for statistical analyses ($\alpha = 0.05$).

Compared to the non-perturbed condition (CON), peak vertical ground reaction force (VGRF) was decreased during the perturbed condition (PERT). There were no significant lower extremity joint kinematic magnitude differences between conditions for the linear or rotational perturbations. Peak hip and knee extensor moments and peak plantarflexor moments were significantly greater during the PERT compared to the CON condition for the linear perturbation but not the rotational perturbation.

The difference in peak VGRF was in the opposite direction of the original prediction, which could be interpreted as anticipation by the performer leading to an altered landing strategy. During the PERT compared to the CON condition, increased force could be placed on the ACL due to the shear force created by the peak knee extensor moment. In addition, landing strategies used by individual participants varied, specifically at the knee, which may predispose certain individuals to an ACL injury.

Of particular interest were the individual participant variations in joint kinematic and kinetics during the PERT compared to CON landings. The findings of this project support that individual landing strategies exist, and these strategies could predispose certain individuals to an ACL injury. While both perturbations did not lead to similar alterations in landing strategies, it remains that an in-flight perturbation does appear to influence landing biomechanics.

INDEX WORDS: ACL, drop landing, perturbation, joint kinetics

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SCOTT W. ARNETT

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by

SCOTT W. ARNETT

Major Professor:	Kathy J. Simpson
Committee:	Ted Baumgartner Michael Ferrara

Electronic Version Approved:

Maureen Grasso
Dean of the Graduate School
The University of Georgia
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CHAPTER I

INTRODUCTION

Background

The incidence and cost associated with ACL injuries indirectly affects society. In the United States, as of 2002, ACL injuries occurred to 1 in every 3000 persons, resulting in cost for surgical reconstruction and/or intensive physical therapy totaling approximately $17,000 per injury, and nationally, an estimated $1.5 billion annually (Childs, 2002). Accordingly, individuals incurring an ACL injury are ten times more likely to develop knee joint osteoarthritis (Fleming, 2003). Furthermore, ACL injuries experienced by collegiate athletes may result in decreased academic performance, loss of athletic scholarships, and long-term disability (Freedman et al., 1998). Unfortunately, females are 2 to 10 times more likely to incur an ACL injury compared to males participating at the same level (AAOS, 2003).

Interestingly, 70% of all ACL injuries are a result of non-contact mechanisms (Boden et al., 2000), which are defined as injuries occurring when no contact occurs between athletes at the instant of injury (Olsen et al., 2004). Furthermore, 41% of all non-contact ACL injuries occur during landings (Kirkendall & Garrett, 2000). However, a non-contact ACL injury does not account for any contact immediately preceding the injury.

In sports that involve jumping and landing (e.g., basketball rebounding, soccer heading, netball goal), athletes often collide in the air, sometimes resulting in at least one of the athletes landing awkwardly. An awkward landing could result in decreased stability or non-optimal alignment of the lower extremities and trunk during landing, placing the ACL at risk for injury.

During flight, if there is direct contact between players (or a stationary object), it results in an external force applied to the athlete of interest that influences or 'perturbs' the athlete's subsequent movements, potentially leading to an unstable landing. Therefore, for this study, from a mechanical perspective, an 'in-flight perturbation' is an external force applied to the athlete for a short duration of time during the flight phase, leading to a change in momentum. Depending on the location of the athlete's body that a perturbation force is applied, the perturbation will cause either only an increase in linear momentum ('linear perturbation') or an increase in linear and rotational momentum ('rotational perturbation') about the medio-lateral axis of the body. It is likely that a perturbation producing rotational momentum will additionally increase the difficulty of landing in a stable position.

As predicted by rigid-body Newtonian mechanics, only external forces, such as those producing in-flight perturbations can affect the projectile motion as well as the rotational motion of the body. Consequently, during the flight phase of a movement, the athlete cannot alter the projectile path (law of inertia) nor alter the total vertical linear or rotational mechanical energy (conservation of energy) of the body in response to a perturbation. Thus, an individual can only manipulate the effects of an in-flight perturbation during landing, particularly as there is insufficient time to generate internal muscle torques to reposition the relative locations of the body segments to prepare for landing.

For these studies, to simulate the flight and landing phases of basketball rebounding and soccer heading, athletes underwent in-flight perturbations during drop landings. For each of the two studies, one of the following two types of perturbations were investigated: 1) linear only perturbation (LIN-PERT) and 2) a rotational perturbation (ROT-PERT). The biomechanics used during the landing phase after an in-flight perturbation, therefore, are the focus of these studies in order to understand the implications of these events on landing performance. Consequently, in-flight perturbations will require altered landing mechanics as compared to typical landings.

Purpose of the Studies

1. Study #1: Determine if a LIN PERT applied during flight results in altered landing biomechanics compared to the biomechanics produced during a non-perturbed condition (CON).
2. Study #2: Determine if a ROT PERT applied during flight results in altered landing biomechanics compared to the biomechanics produced during a non-perturbed condition (CON).

Premises of the Studies

What happens to an athlete during landing after a LIN-PERT or ROT-PERT has occurred? During a basketball rebound, for example, another athlete may apply a LIN-PERT to the athlete of interest by pushing the athlete forward for a short duration of time. As stated earlier, a perturbation is a force applied for a short interval of time during a flight phase. The magnitude of an in-flight perturbation is the impulse applied to the athlete. In this simple basketball example, impulse = the average force that the opponent applied to the athlete multiplied by the amount of time that the force was applied.

According to the impulse-momentum principle, the magnitude of the linear impulse applied to the athlete determines the relative increase in linear momentum as shown in the equation below:

$$I = \Delta M \tag{1}$$

where I represents the impulse acting on the individual during flight and ΔM represents the change in the momentum of the athlete. See Appendix A for further explanations regarding the impulse applied during a perturbation. Thus, the greater the magnitude of force applied to the athlete at any given time during the perturbation, or the longer the total time during which the perturbation force is applied, the greater the change in the athlete's antero-posterior momentum. Once the perturbation ceases, the athlete's new antero-posterior momentum becomes constant until the athlete contacts the ground. Therefore, with unanticipated and increased anterior-directed momentum, the athlete will contact the ground with greater anterior velocity and at a different angle of approach relative to the ground.

During landing following a LIN PERT, adequate stability of the athlete is required to land safely. Inadequate stability during landing could lead to abnormal loading of the ACL. Therefore, to ensure stability in the antero-posterior direction, the anterior momentum caused by the perturbation will require the athlete to create an opposing impulse in order to reduce the anterior momentum of the body's COM to zero. The opposing impulse was created by posterior-directed ground reaction forces (GRF) in reaction to the athlete's anterior inertia and to the force applied by the athlete to the ground.

Upon contact with the ground, the anterior-directed GRF generated also serves to create static friction so the feet remain fixed to the ground. With the feet fixed to the ground, the ankle joint flexion/extension axis becomes a fixed axis of rotation for sagittal plane rotation of the rest of the body. Hence, the linear momentum transfers into angular momentum, causing the body to rotate forward. To stop rotating, the athlete must create an opposing rotational impulse or else the athlete will continue rotating until the center of mass of the body (COM) moves outside of the base of support and loss of balance occurs.

From the above, it is evident that the athlete must produce successful biomechanical adaptations to land in a stable position following a perturbation. As shown in Figure 1.1, the biomechanical strategy utilized during landing after a perturbation depends on the antero-posterior location of the body's COM (COM_x) relative to the antero-posterior location of the axis for flexion/extension about the ankle joint (ANK_x). The COM_x represents the anterior/posterior location of the entire body's mass and approximately that of the body's weight. During landing, in order for the athlete to be able to rotate without losing balance, stable landings likely result when the body weight shifts to be in line with the ANK_x or slightly behind the ANK_x. Thus, one predicts that stable landings will occur when the COM_x position relative to the ANK_x, defined as COM_RP (COM_RP = $COM_x - ANK_x$) is equal to zero or a small negative number, and unstable landings will occur if COM_RP is a positive number. If COM_RP is > 0 at initial contact, one surmises that the COM_x will move anterior beyond the toe during the landing phase. The weight of the body will create a moment (moment = force x moment arm) any time the line of gravity does not pass through the ANK_x.

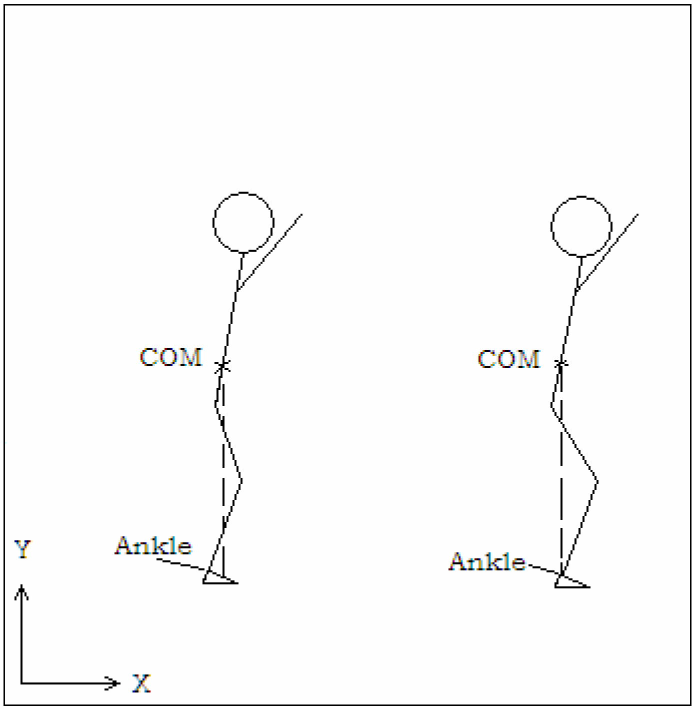

Fig. 1.1 Position of the COM relative to the ankle during landing

As the COM_RP is the magnitude of the moment arm vector for the person's weight, greater COM_RP magnitudes create greater body weight moments that rotate the person forward with greater dorsiflexion acceleration. Consequently, if the body weight moment created is greater than the opposing plantar flexor muscle moment created to rotate the person upright, the participant falls forward. In addition, if the body weight is shifted behind the ANK_x, i.e., COM_RP is a large negative number, the participant will likely rotate backwards (plantarflexion direction) and need to take a step backwards.

In addition, the COM_RP will likely alter joint biomechanics utilized to achieve a successful landing after a perturbation. One predicts that, compared to non-perturbation landings, increased extensor net muscle moments about the ankle, knee, and hip joints will exist to counteract the moment created by body weight when COM_RP is positive and the body's rotational momentum increases due to the perturbation.

Depending on the knee joint position, all else equal, increased knee extensor net muscle moments during perturbation compared to non-perturbation landings could potentially suggest greater quadriceps force and, consequently, increased anterior shear loading on the ACL.

When a perturbation is rotational, it will increase both the linear and rotational momentum of the body. The magnitude of linear momentum will be less than that created in the LIN PERT condition. However, a ROT PERT will increase the rotational momentum of the body during flight, such that the body will already possess angular momentum at contact with the ground.

During landing after a ROT PERT, the posterior GRF's will counteract the linear momentum of the COM of the body similarly as it does when landing after a LIN PERT.

However, in addition to the rotational momentum created once the feet are fixed to the ground, the body will have rotational momentum present before contact. Therefore, it is predicted that increased trunk flexion will shift the line of gravity further forward and closer to the anterior edge of the base of support, compared to a CON landing. As rotational momentum of the body increases after a ROT PERT, it will require altered biomechanics during landing compared to a CON landing. The increase in rotational momentum will likely require increased peak extensor net muscle moments.

Altered landing biomechanics after a LIN PERT or ROT PERT may lead to increases in knee valgus angle and/or net muscle moments. During sport movements, excessive valgus motion and valgus net muscle moments have been implicated in theories of ACL injury mechanisms and are associated with ACL injury in females (Hewett et al., 2005; McLean et al., 2005). As the ACL is a secondary restraint to valgus motion (Inoue et al., 1987; Whiting & Zernicke, 1998), high peak knee valgus angles may produce high strain on the ACL and high peak valgus net muscle moments may produce high tensile stress in the ACL. However, the effects of in-flight perturbations on these variables are not easily predictable, and previous research of in-flight perturbations appears to be non-existent.

Therefore, such perturbations could substantially alter landing kinematics (quantities related to description of movement, e.g., time and space) and kinetics (quantities related to the causes of the changes in motion, e.g., forces cause the body to accelerate) that may be related to ACL loading and deformation. Understanding these alterations in mechanics, therefore, may increase the understanding of mechanisms associated with ACL injury movements involving contact with another person/object during the flight phase of sport movements.

Hypotheses

Study #1, LIN PERT compared to CON will exhibit:

1. Increased angular displacement at the knee and hip joints between initial contact and the end of the landing phase.
2. Increased peak knee and hip flexion angles.
3. Increased peak vertical and posterior GRF.
4. Increased posterior GRF impulse.
5. Increased peak ankle, knee, and hip extensor net muscle moments.
6. Peak knee valgus angles and peak knee valgus net muscle moment values that do not vary from the CON values by more than .5 SD.

Study #2, ROT PERT compared to CON will exhibit:

1. Increased displacement at the knee and hip joints between initial contact and the end of the landing phase.
2. Increased peak knee and hip flexion angles.
3. Increased peak vertical and posterior GRF.
4. Increased posterior GRF impulse.
5. Increased peak ankle, knee, and hip extensor net muscle moments
6. Peak knee valgus angles and peak knee valgus net muscle moment values that do not vary from the CON values by more than .5 SD.

Significance of Study

Non-contact ACL injuries are reported to have a multi-faceted etiology (Hewett et al., 2006) and are not specific to an age group or sport. However, compared to males, females are at an increased risk of injuring their ACL (AAOS, 2003). As female sport participation continues to increase since the passage of Title IX (Thein & Thein, 1996; USDOJ, 2006), reducing the rate of ACL injury for females is especially imperative. The primary focus of recent research has been to compare anatomical and biomechanical factors between genders. The goal being to elucidate factors contributing to the increased incidence of non-contact ACL injuries in females (Decker et al., 2003; Ford et al., 2003; Hass et al., 2005; Hewett et al., 2004; Hewett et al., 2005; Kernozek et al., 2005; Lephart et al., 2002; Salci et al., 2004; Swartz et al., 2005). However, an understanding of ACL injury etiology is still incomplete at this time.

It has been reported that a perturbation is likely one of multiple factors involved in ACL injuries (Hewett et al., 2006). However, at this time, the quantitative effects of in-flight perturbations on landing biomechanics are unknown. In-flight perturbations result in a movement combining three of the four most common mechanical subgoals of a movement associated with non-contact ACL injuries: landing, rapid deceleration of the body, and reacting to a perturbation (Fleming, 2003). For the movement in this study, the athlete performed a vertical drop landing, reacted to an unanticipated perturbation, and rapidly decelerated the body's COM in the antero-posterior and vertical directions during the landing phase.

Limitations

A limitation of both of these studies was that the participants were college-aged club or recreational athletes, limiting the generalizability of the findings to other similar college-aged individuals of similar sport skill levels.

Assumptions

The first assumption of the study was that for the inverse dynamics model, the body acts as multiple rigid objects connected by frictionless pin joints with no force absorption within segments and between joints. The second assumption was that the participant anticipated the laboratory perturbation in a similar way as an athlete anticipates an in-flight hit coming from an opposing athlete during sport participation.

CHAPTER II

REVIEW OF LITERATURE

An understanding of the exact mechanisms leading to an anterior cruciate ligament (ACL) injury during landing is incomplete at this time. It is likely that an ACL injury is multifaceted with specific mechanisms predisposing an individual to injury. A more comprehensive understanding of ACL anatomy and physiology, landing biomechanics, and environmental influences is important for identifying mechanisms predisposing an individual to ACL injury. Therefore, this chapter is presented in three major sections: 1) The ACL, 2) biomechanics of drop landings, and 3) neuromechanical responses to perturbations.

ACL: Anatomy

The ACL (Figure 2.1) is one of four major ligaments of the knee and provides for the functional stability of the knee joint. The ACL inserts on the posterior portion of the medial side of the lateral femoral condyle and runs anteriorly and distally through the femoral intercondylar notch ending where it inserts on the anterior portion of the middle of the tibial plateau (Duthon et al., 2006).

The anatomy of the ACL is complex and consists of many integral parts. The ACL is often considered to have two separate 'bundles' (Duthon et al., 2006; Girgis et al., 1975), the anteromedial and posterolateral bundles, but also has been described as having a third bundle (Amis & Dawkins, 1991), the intermediate bundle. These bundles are named based upon their

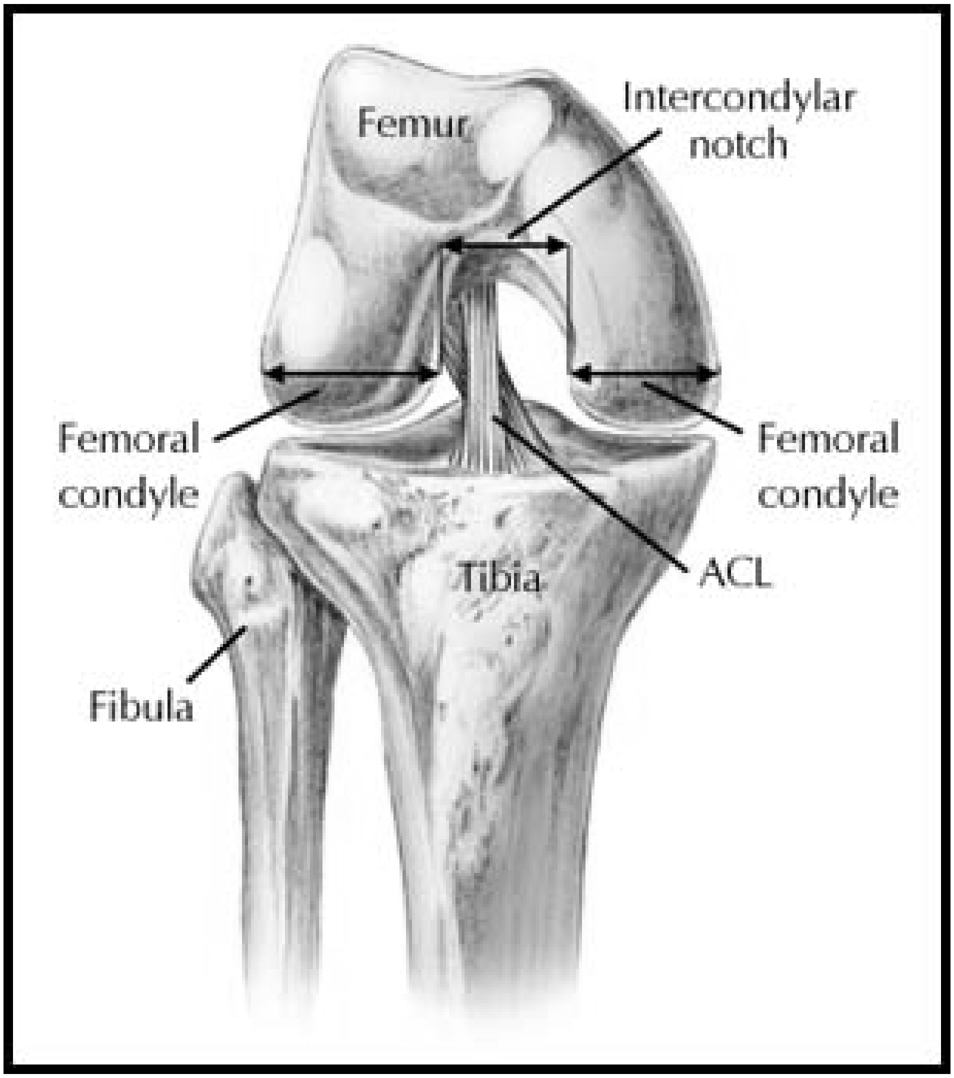

Figure 2.1 The anterior cruciate ligament.

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tibial insertion (Fu & Stone, 2001). The anteromedial and posterolateral bundles are thought to have separate functional roles (Bach et al., 1997; Fu & Stone, 2001) as the former bundle is in tension as the knee becomes flexed and the latter bundle is in tension as the knee becomes extended (Amis & Dawkins, 1991; Crowninshield et al., 1976). The ACL bundles consist of multiple fascicles enclosed by the paratenon, with each fascicle consisting of multiple subfasciculi surrounded by the epitenon and subfascicular units surrounded by the endotenon (Duthon et al., 2006). Fibers of subfascicular units are composed of two different types of collagen fibrils; 1) Type I accounting for over half of the ACL and resisting tensile forces, and 2) Type II accounting for most of the remaining portion of the ACL and maintaining the ACL's 3-dimensional organization (Strocchi et al., 1992). The remaining portion of the ACL consists of fibroblasts and elastic components (Strocchi et al., 1992), with the latter being one of the systems of the intercellular matrix of the ACL (Duthon et al., 2006).

The intercellular matrix of the ACL is composed of: 1) collagen, 2) glycosaminoglycans, 3) glycol-conjugates, and 4) elastic components. The five different types of collagen (Type I, II, III, IV, VI) present in the matrix have unique functions, locations and concentrations within the ACL. Glycosaminoglycans function to increase the shock absorbing ability of the ACL (Duthon et al., 2006), while the elastic components of the matrix allow the ACL to undergo multidirectional stresses (Strocchi et al., 1992), maximum stress (Strocchi et al., 1992), and strain (Duthon et al., 2006). There are four glycol-conjugates present in the ACL matrix and these components are integral in ACL repair, growth, and normal daily function (Duthon et al., 2006).

ACL: Physiology

Ligaments, in general, are reported to have three main functions (Frank, 2004) allowing them to contribute to the health of the joint. The first of these functions is to provide stability within the joint (Anderson et al., 2000; Frank, 2004). The second function of a ligament is to guide the joint through its full range of motion safely when the joint is loaded in tension (Anderson et al., 2000; Frank, 2004). Finally, ligaments contribute to joint proprioception through neurological feedback mechanisms, leading to muscular activation that protects the joint and provides awareness of joint position (Anderson et al., 2000; Frank, 2004).

Related to its role in maintaining knee joint health, the ACL has specific primary and secondary functions. The primary function of the ACL is to resist anterior translation of the tibia with respect to the femur (Butler et al., 1980; Crowninshield et al., 1976; Fu & Stone, 2001; Furman et al., 1976; Girgis et al., 1975; Whiting & Zernicke, 1998). Conversely, the ACL resists posterior translation of the femur, relative to the tibia, when a posterior force is applied to the distal end of the femur (Whiting & Zernicke, 1998). As a secondary function, the ACL resists internal tibial rotation, specifically when the knee is close to full extension (Beynnon et al., 1997; Crowninshield et al., 1976; Duthon et al., 2006; Furman et al., 1976; Girgis et al., 1975; Whiting & Zernicke, 1998). Another secondary function of the ACL is to resist valgus motion of the knee joint (Inoue et al., 1987; Whiting & Zernicke, 1998).

When discussing the functions of ligaments, it is important to mention ligament properties that provide a supportive role for ligament functions as well as note how these properties may lead to deleterious joint biomechanics. Viscoelastic properties of the

ACL allow it to undergo stress-relaxation (tensile load decreases when ligament is kept at a constant length), creep (elongation of ligament over time under a constant or cyclically repetitive tensile load), and hysteresis (loss of energy with repetitive loading and unloading) (Frank, 2004; Whiting & Zernicke, 1998; Woo et al., 1999). These properties are reported to protect the ligament from injury (Frank, 2004; Woo et al., 1999). As an example, stress-relaxation protects the ACL by decreasing the magnitude of the applied stressor when the ligament is deformed to a constant length (Frank, 2004; Woo et al., 1999). Contrarily, an excess of one of these properties in the ACL, such as creep, may increase joint laxity, which can increase the risk of injury within the knee joint (Frank, 2004; Woo et al., 1999).

The ACL also possesses non-viscoelastic properties, crimp and recruitment, which protect it from trauma and support its functional roles. Viewing the ACL microstructure with polarized light when the fibril is not stretched reveals that the fibrils possess a wavy pattern, hence the term, crimp (Duthon et al., 2006; Frank, 2004). Crimp is proposed to be a protective mechanism (Woo et al., 1999), such that when low tensile loads are applied to the ACL, the waviness, or crimp, of the ligament straightens and provides a safety margin of elongation for the ligament (Duthon et al., 2006). As tensile loads increase, it is proposed that more fibrils are recruited to resist the tensile load, hence the term recruitment (Duthon et al., 2006).

ACL: Mechanical Properties

ACL mechanical properties explain how the ACL behaves when loaded, thereby providing crucial information for clinicians and surgeons, especially those dealing with ACL reconstruction. More importantly in the case of injury, these properties provide details regarding how ACL trauma might occur. An ACL tear or rupture can be induced when the magnitudes of the stress applied to the ACL exceeds its yield limit values. In addition, high rates of loading on the ACL lead to an increase in the yield point strength of the ligament, which implicates strain as a risk factor for ACL injury during these conditions (Whiting & Zernicke, 1998). Furthermore, bundles of the ACL appear to fail due to strain (Butler et al., 1986).

It should be noted that with a given load the ACL's unique structure prevents all bundles from being loaded equally (Woo et al., 1999). In addition , the bundles shift the proportions of stress generated as the knee flexes (Woo et al., 1991). Therefore, measures of mechanical properties are difficult to assess, or inaccurate, due to lack of recruitment of all ACL fibers at the time of measurement (Woo et al., 1991). However, in order to develop a more complete understanding of the mechanical properties of the ACL, the behavior of the ACL during loading must be investigated. Researchers have measured or estimated tensile forces and strain produced in the ACL during axial loading of the ACL, movement through the full ROM of the knee joint, activation of the knee joint musculature, mechanisms associated with ACL injury, and performance of functional movements.

The mechanical requirements of human movement do not typically result in only axial loading of the ACL. However, loads applied along the axis of the ligament recruit more of the ACL (Woo et al., 1999) and when the ACL is loaded axially, compared to other directional loading, knee flexion angle has no influence on the mechanical properties of the ligament (Woo et al., 1987). Using axial loading, Butler and colleagues measured the mechanical properties of individual ACL fascicle units (Butler et al., 1986) and bundles of the ACL (Butler et al., 1992). Using several fascicles from the ACL, Butler et al. (1986) reported the modulus of elasticity (~278-325 MPa), maximum stress (~30-40 MPa), and maximum strain (~14-16 %). However, the authors did not report the ACL bundle from which the fascicles were dissected. To determine if differences existed between bundles of the ACL, Butler et al. (1992) measured the mechanical properties of the anteromedial, anterolateral, and posterolateral bundles of the ACL. Axial loading of the bundles at a strain rate of 100%/s resulted in the following measurements; modulus of elasticity (AMB = 283.1, ALB = 285.9, PLB = 154.9), maximum stress (AMB = 45.7, ALB = 30.6, PLB = 15.4), maximum strain (AMB = 19.1, ALB = 16.1, PLB = 15.2), and strain energy density (AMB = 3.3, ALB = 2.2, PLB = 1.1) (Butler et al., 1992). It was reported that these findings supported that the anterior bundles of the ACL fail at higher maximum stress than the posterior bundle. However, the strain values at which the ACL mechanically fails were determined even though maximum strain was not different between the anterior and posterior bundles (Butler et al., 1992).

Although these analyses provide information regarding the mechanical properties of the ACL, they lack information regarding the effect of loading direction. To develop a complete understanding of ACL mechanics, it is important to determine how ACL orientation during loading influences the forces acting on the ligament. Woo et al. (1991) investigated the effects of ACL orientation, with respect to the femur and tibia, on the mechanical properties of the ACL. ACL mechanical properties were assessed in 22-35 year olds in both an anatomical and tibial orientation. For the anatomical orientation, the femur and tibia were rotated so that the ACL was aligned to the vertical direction but still in its natural anatomical orientation relative to the femur and tibia. For the tibial orientation, the tibia and femur were oriented vertically with the femur translated anteriorly so that the ACL was aligned in the vertical direction and no longer in its natural anatomical alignment relative to the femur and tibia. Stiffness and energy absorbed up to the failure point were not significantly different due to ACL orientation, but in the anatomical orientation, the values for stiffness and energy absorbed were 11% and 40% higher, respectively. The ultimate load of the ACL in the anatomical orientation was significantly greater than that measured in the tibial orientation (2160 ± 157 N and 1602 ± 167 N, respectively). The authors noted that orientation of the ACL played a role in the values of its mechanical properties. However, the orientation of the ACL used in this study may not occur during landing, leading to over- or under-estimation of its mechanical properties during landing.

Until recently, the mechanical properties of the ACL were primarily investigated on male cadaver knees. Chandrashekar et al. (2006) investigated whether differences in ACL mechanical properties existed between 10 male and female (n = 20) cadaver knees. To load the ACL along its longitudinal axis, bone plugs from the femur and tibia were dissected parallel to the longitudinal axis of the ACL and oriented vertically in the frontal plane and 45° to each other in the sagittal plane (Chandrashekar et al., 2006). The authors reported that the load at failure and stiffness of the ACL was significantly higher in males than females (load at failure = 1818 ± 699 and 1266 ± 527 N, respectively; stiffness = 308 ± 89 and 199 ± 88 N/mm, respectively) (Chandrashekar et al., 2006). After adjusting for age, sex, body mass, height, BMI, and ACL volume, differences between males and females existed for strain at failure (.30 ± .06 and .27 ± .08 %, respectively), stress at failure (26.35 ± 10.08 and 22.58 ± 8.92 MPa, respectively), modulus of elasticity (128 ± 35 and 99 ± 50 MPa, respectively), and strain energy density at failure (3.50 ± 1.69 and 3.17 ± 2.62, respectively) (Chandrashekar et al., 2006). The authors concluded that gender differences in mechanical properties of the ACL could be a major factor influencing the increased incidence of ACL injuries in females.

While axial loading provides information about the mechanical properties of the ACL, it does not provide information about how the ACL behaves during movements associated with the knee joint. Throughout the full range of knee flexion/extension motion the demand placed on each of the ACL bundles changes (Amis & Dawkins, 1991; Crowninshield et al., 1976).

Using a novel strain gage, Bach et al. (1997), surgically implanted liquid metal strain gauges in the ACL to measure strains of the posterolateral and anteromedial bundles. Strain measured in anteromedial as compared to posterolateral bundles was significantly different throughout the full range of motion of the knee (Bach et al., 1997). Furthermore, the authors noted that the different bundles created tension at different intervals of knee flexion, with the posterolateral and anteromedial bundles in tension when the knee was extended and flexed, respectively (Bach et al., 1997). Thus, during sport activities when knee flexion/extension movements are constantly occurring, only part of the ACL provides resistance to loads, which produces higher stresses to the loaded bundles than if all bundles were providing resistance.

During the performance of sport movements, such as a drop landing, activation of lower extremity muscles occurs not only to help control landing biomechanics but also to provide stability to the knee joint and protect the ACL and other passive structures. In contrast, muscle activation can destabilize the joint. The quadriceps angle of pull changes depending on the knee flexion angle. As the knee becomes more extended, a proportion of the quadriceps force creates an anterior shear force on the tibia that the ACL opposes (DeMorat et al., 2004). Therefore, it is important to understand the effect of lower extremity muscle activation on ACL tissue mechanics. Utilizing a simulation model, the force acting on the ACL has been reported to increase as the activation of the quadriceps increases (~100N-500N at 15° of knee flexion, 10-100% of quadriceps activation, respectively) (Pandy & Shelburne, 1997). Therefore, compared to a non-loaded condition, when a constant quadriceps load (100 N) was applied to the tibia, the ACL force increased for the first 50° of knee flexion (Markolf et al., 2004). In addition, Li et al. (1999) also found that a constant 200 N quadriceps load resulted in a significant increase in tensile force acting on the ACL, but only when moving the knee from full extension (27.8 ± 9.3 N) to 15° of knee flexion (44.9 ± 13.8 N). Activation of the gastrocnemius via electrical stimulation results in an increase in ACL strain (3.8%) when the knee is at 15° of knee flexion, compared to when the gastrocnemius is not activated (Fleming et al., 2001). When the gastrocnemius and quadriceps are activated simultaneously, maximum strain in the ACL increases up to approximately 5% (Fleming et al., 2001). However, when the hamstrings apply a load or are activated, the force acting on the ACL and ACL strain decrease at low knee flexion angles (Fleming et al., 2001; Li et al., 1999; Pandy & Shelburne, 1997). The protective effect of the hamstrings muscle group however, is ineffective when the knee is at full extension (Li et al., 1999; Pandy & Shelburne, 1997).

Measurements of tensile force and strain in the ACL during the application of a linear force (anterior tibial force), a rotational moment (valgus/varus, internal/external) or a complex load (e.g., anterior tibial force with valgus/varus moment) provide information into how ACL injury mechanisms influence the tissue mechanics of the ACL. Anterior tibial forces led to increased ACL tensile force, regardless of the amount of knee flexion (Markolf et al., 1995). Compared to baseline, at 30° knee flexion and full extension, respectively, the ACL tensile force displayed quantities nearly 100% and 150% of the applied anterior tibial load (100 N) (Markolf et al., 1995). Anterior tibial forces (up to 130 N) have also been reported to lead to increased strain in the ACL (~ 4%) (Fleming et al., 2001). Internal rotation torques (10 Nm) applied to the tibia resulted in an increased ACL tensile force at all knee flexion angles (Markolf et al., 1995) and with the knee flexed to 20° an increase in internal tibial torque (0-10 Nm) leads to an increase in ACL strain (~3%) (Fleming et al., 2001). Anterior tibial loads combined with a valgus moment (10 Nm) for all knee flexion angles greater than 5° led to increased ACL tensile force, compared to the anterior tibial loading condition alone (Markolf et al., 1995). In addition, the combination of a valgus and internal tibial rotation moment has been reported to lead to an increase in ACL strain, compared to the application of either moment by itself (Shin et al., 2005). The combination of a valgus and internal tibial rotation moment was reported to produce strain values within the range shown to damage the ACL (Butler et al., 1992; Shin et al., 2005). Knee flexion angle played a role in increasing ACL tensile forces when anterior tibial loads were combined with a varus moment (knee flexion angles <30° and >50°) or an internal tibial torque (knee flexion angles <20°) (Markolf et al., 1995).

Curiously, not all of the conditions measured led to an increase in the ACL tensile force. When the combined load applied to the tibia consisted of an anterior linear force and an external tibial torque (10 Nm), the ACL exhibited a decreased tensile force when the knee was flexed greater than 10° (Markolf et al., 1995). However, when the knee was flexed less than 10° the ACL tensile force increased as the knee extended, but the ACL tensile force was not significantly different from the condition with only an anterior load applied to the tibia (Markolf et al., 1995). Therefore, individuals landing with their knees flexed more than 10° and feet externally rotated likely decrease the tensile loading on the ACL during landing. When knee flexion was greater than 40°, a combined load of internal tibial torque and anterior tibial force led to a decrease in ACL tensile force, compared to the anterior tibial load alone. Landing with the knee flexed more than 40° is not an effective landing strategy as it limits the range of motion available at the knee. Although the combined loading utilized in this study occurs during landing, it should be noted that this investigation was not conducted in vivo, therefore, the effects of safety mechanisms (e.g., muscle spindles) during these loading patterns are unknown.

Measurement of ACL strain *in-vivo* allows researchers to assess the tissue mechanics of the ACL during functional movements. Cerulli et al. (2003) compared the strain of the ACL that occurred during single-limb hopping to strain displayed during a Lachman joint laxity test. The hopping movement was selected as one commonly encountered during sports participation and linked to ACL injury (Boden et al., 2000). Average peak ACL strain measured during the hopping test was greater (5.47 ± 0.28%) than that measured during the Lachman test (2.00 ± 0.17%) (Cerulli et al., 2003). Of particular interest in this study is that compared to ACL strain measured at the beginning of the flight phase, ACL strain increased during the flight phase and continued to increase until immediately after the peak vertical ground reaction force occurred (Cerulli et al., 2003). Increases in strain during the flight phase might occur due to pre-activation of the musculature around the knee to prepare for landing (Cerulli et al., 2003).

Heijne et al. (2004) measured ACL strain during the performance of four rehabilitative tasks (step-up, step-down, lunge, and one-legged sit-to-stand). ACL strain was not different between the four exercises and ranged from 1.9-2.8%. However, the authors noted that ACL strain increased as the knee was extended during each exercise (Heijne et al., 2004).

ACL: Non-Contact Injury Mechanisms

Focusing on mechanisms of ACL injury requires understanding the differences between contact and non-contact ACL injuries. Olsen et al. (2004) classified ACL injuries as one of three types based on contact of the player with another player/object at the time of injury. A direct injury is caused by direct contact with some part of the injured lower extremity. During an indirect injury, direct contact occurred with another part of the body besides the injured lower extremity. Non-contact injuries occur when there is no contact between athletes or the athlete and an object. Non-contact ACL injuries constitute 70% of ACL injuries (Boden et al., 2000), implicating the importance of determining mechanisms of non-contact ACL injuries.

To understand what is occurring during injuries to the ACL, researchers must look at the movements occurring at the time of injury. Although this is not always feasible due to lack of video footage and any practical method to obtain other valid biomechanical measures during competition, researchers have questioned the athletes about events that occurred at the time of their injury. However, this technique has limitations, as the validity of the information depends on the participants' ability to report accurately events occurring days or weeks earlier.

Boden et al. (2000) attempted to account for this flaw by examining videotape of ACL injuries in addition to administering questionnaires. Sixty-five males and twenty-five females filled out the questionnaire. Video footage of an additional 27 ACL injuries from collegiate and professional teams were qualitatively analyzed using similar criteria to the questionnaire. However, athletes analyzed from video were not administered a questionnaire. Basketball, football, and soccer were reported as the most frequent sports participated in involving ACL injuries according to results from the questionnaire (Boden et al., 2000). ACL injuries were reported to occur when the knee was close to full extension. Movements with phases that required deceleration with or without a change in direction and landings were the most common movements associated with non-contact ACL injuries according to the questionnaire and the only movements leading to non-contact ACL injury from video tape analysis (Boden et al., 2000). Stepping or landing onto irregular surfaces or in an inverted foot position were commonly reported in the questionnaire as factors involved in landing injuries. Observationally from the videotapes of the ACL injury situations, there was an opposing player in close proximity to the athlete at the time of ACL injury, suggesting to Boden and colleagues (2000), that the injured athlete's movement execution may have been influenced by the opposition's proximity. Knee valgus collapse occurred following quick decelerations and single-leg landings during "most" of the injuries, and backward lean of the trunk was also displayed after a quick deceleration (Boden et al., 2000).

Rapid body deceleration and a fixed foot are components that are often involved during ACL injuries (Boden et al., 2000). Eccentric actions of the knee extensor muscles are needed to prevent knee joint collapse but produce anterior shear forces on the proximal tibia at low knee flexion angles (Boden et al., 2000) that strain the ACL. An investigation of the effect of quadriceps forces on ACL failure implicated quadriceps forces as a factor contributing to ACL rupture (DeMorat et al., 2004). An important conclusion stated by Boden et al. (2000) is that imbalances in knee loading may occur due to unanticipated conditions encountered during sports participation, such as perturbations.

Other mechanisms predisposing an individual to ACL injury have been suggested, including the role of the hamstrings muscle group (More et al., 1993) and hip flexion (Ball et al., 1999). Using a cadaver knee model, More et al. (1993) reported that anterior tibial translation and internal tibial rotation were reduced when the hamstrings applied a load to the knee model. Thus, proper functioning of the hamstrings muscle group is vital for ACL health, which may not be possible if the hamstring muscle group is excessively flexible (Boden et al., 2000). In conjunction with the role of the hamstrings muscle group is the role of hip flexion in promoting knee joint stability. Ball et al. (1999) reported that the amount of hip flexion played a crucial role in the ability of the hamstrings to counteract the force produced by the quadriceps. Increased hip flexion resulted in increases in the ratio of hamstring to quadriceps isometric force (Ball et al., 1999).

Drop Landings: Biomechanics

This section is divided into three major sections. The first section provides a description of the drop landing movement. The second and third sections discuss the literature focused on how manipulation of environmental and movement-technique factors effect landing strategies, respectively.

Biomechanical analyses of drop landings have been conducted for approximately three decades. However, the concept of a drop landing and the biomechanics of drop landings must be explained before interpretations can be made about ACL related research using drop landings. Drop landings provide a movement closely resembling falling towards the ground, similar to the movement performed during the latter portion of a basketball rebound, volleyball block, or jumping to head a soccer ball. Drop landings are a laboratory created movement, allowing researchers to investigate the effects of environmental, contextual, and/or physical factors on human landing.

Drop landings consist of two major phases, 1) the flight phase and 2) the landing phase. In a laboratory, drop landings are initiated when an individual steps from a standardized height platform or releases from a bar suspended from the ceiling. The subsequent movement consists of the individual falling towards the ground until contact occurs between the foot and ground (i.e., initial contact). Following contact with the ground, the lower extremity joints collapse under control to help decrease the momentum of the body's center of mass to zero, which is considered a successful landing (Devita & Skelly, 1992). Lees (1981) provided an explanation supporting the advantage of joint collapse during landing in that if the human body is rigid upon contact, no joint collapse, its momentum quickly decreases leading to high external force production. However, if the joints collapse, the body acts more like a spring increasing the time interval over which momentum decreases, leading to decreases in force production (Lees, 1981).

According to these proposed strategies, it would seem that as environmental factors were manipulated, such as increases in drop landing height, the individual would manipulate her biomechanics to increase this spring-like strategy. An increase in drop landing height will likely result in an altered landing strategy due to accommodating increased velocity at contact, compared to a drop landing from a lower height. By having participants perform drop landings from heights lower and higher than their respective max vertical jump, James et al. (2003) classified landing strategies based upon an individual's response to these manipulations. The three strategies defined were: 1) positive/negative biomechanical, positive – increases in drop landing height lead to increases in vertical ground reaction force but forces increase at a rate less than the increase in height, negative – increases in drop landing height lead to decreases in vertical ground reaction force; 2) neuromuscular, increases in drop landing height lead to no increase in vertical ground reaction force; and 3) Newtonian, a 1-to-1 linear relationship between drop landing height and the vertical ground reaction force. The authors concluded that Newtonian strategies were not present in this sample, however, some participants utilized neuromuscular (fully accommodating) response strategies for specific variables. For most variables, participants utilized negative and positive biomechanical strategies. Further, the model was able to differentiate between groups using either a positive or negative biomechanical landing strategy, which the authors noted could be used as a tool to define populations at risk for injury (James et al., 2003).

Changes in lower extremity kinematics and the temporal patterns of these motions are one way in which landing strategies are adjusted as drop landing height increases. McNitt-Gray (1991) reported that during landing the COM of the body reached a lower vertical position and knee and hip flexion angles increased as drop landing height increased (.32, .72, and 1.28m). Similar results have been reported with participants demonstrating increases in the range of motion at the ankle, knee, and hip joints during landing when dropping from .32, .62, and 1.03m (Zhang et al., 2000). Furthermore, ankle, knee, and hip angular velocities increased with an increase in drop landing height (McNitt-Gray, 1991). However, McNitt-Gray (1991) reported that the body's COM position and ankle, knee, and hip joint position (ankle, knee, and hip) at initial contact with the ground was similar regardless of drop landing height.

Ground reaction forces play a vital role in decreasing an individual's momentum during landing. Thus, increases in the magnitude of the ground reaction forces will likely occur to help decrease the momentum gained from an increase in drop landing height. Peak vertical and antero-posterior ground reaction forces (VGRF and A-P GRF, respectively), and peak rate of application of VGRF have been reported to increase as drop landing height increases (McNitt-Gray, 1991). Similar results have been reported for peak VGRF when dropping from heights different from those utilized by McNitt-Gray (1991) (Dufek & Bates, 1990; Zhang et al., 2000). Conversely, absolute time to peak VGRF and A-P GRF decreased as drop landing height increased (McNitt-Gray, 1991).

Lower extremity joint kinetics play a large role in energy dissipation during landing. In order to decrease the momentum of the body to zero, the lower extremity musculature must produce eccentric torques to help decrease the velocity of the COM of the body. Peak joint extensor moments at the ankle, knee, and hip have been reported to increase as drop landing height increases (McNitt-Gray, 1993; Zhang et al., 2000).

McNitt-Gray (1993) noted that increases in drop landing height led to a more prominent role of the lower extremity muscle groups in controlling lower extremity kinematics and an increased relative demand of the knee extensor musculature, compared to the ankle and hip. This latter factor raises concern as the ACL is likely to receive some of the increased demand of the knee extensor musculature, especially when the knee is almost fully extended (DeMorat et al., 2004). An increase in drop landing height also led to increased work done at the ankle, knee, and hip joints, with the majority of the work done during the first half of landing (McNitt-Gray, 1993).

McNitt-Gray (McNitt-Gray, 1991; McNitt-Gray, 1993) noted several interesting conclusions regarding her analyses of landing strategies and drop landing height. The author noted that an extended position of the joints at contact contributed to an increased range of motion during landing. This increased range of motion could allow the performer to have more flexibility over the landing strategy utilized, which, in turn, could provide a safety margin during landing. However, an extended knee position during landing is proposed as a mechanism for ACL injury (Boden et al., 2000). It was also noted that the joints or segments closest to the point of application of the force reached zero velocity first, compared to joints and segments further away from the point of application of the force (McNitt-Gray, 1991). Lastly, there appeared to be a common temporal strategy used across drop landing heights with adjustments made to the magnitude of specific biomechanical variables to accommodate increases in contact velocity (McNitt-Gray, 1993). It should be noted that the methodology utilized in these two studies warrants interpretive caution, as only males were studied and a single trial was analyzed based upon the participant choosing his ‘preferred’ trial.

Investigating the manipulation of movement-technique factors provides useful information regarding the effectiveness of certain strategies on landing biomechanics. Of particular interest is an understanding of the effects of lower limb stiffness on the biomechanics of drop landings. Lees (1981) predicted that increases in lower extremity stiffness result in increases in ground reaction forces, compared to behaving more like a spring by flexing the lower extremity joints.

DeVita and Skelly (1992) investigated the kinetics and energetics at the ankle, knee, and hip both prior to initial contact and during soft and hard landings. The authors found that the ankle plantarflexors were the primary energy absorber during landing, followed by the knee extensors with the hip extensors providing the least amount of energy absorption of the three lower extremity joints (Devita & Skelly, 1992).

To determine patterns of change in lower extremity energy absorption during drop landings Zhang et al. (2000) had participants perform three landing techniques (soft, normal, stiff). Vertical ground reaction forces increased from soft to normal and normal to stiff and range of motion of the knee and hip decreased from soft to normal and normal to stiff (Zhang et al., 2000). Range of motion of the ankle decreased between normal and stiff and soft and stiff conditions but was not different between soft and normal conditions (Zhang et al., 2000). Work done at the hip decreased as landing strategy moved from soft to normal to stiff, while the work done at the knee showed differences only when dropping from 32 and 62 cm and the work at the ankle showed differences for some techniques at all heights but not all techniques like that at the hip (Zhang et al., 2000). The work at the knee appeared higher than the work performed at the ankle and hip when compared across all height by technique combinations and the knee and hip performed the majority of total lower extremity work, although statistical significance was not determined for either of these findings (Zhang et al., 2000).

The influence of ankle position at initial contact on energy absorption and subsequent performance was investigated during the execution of drop jumping (Kovacs et al., 1999). Participants were required to perform drop jumps utilizing both a forefoot and heel to toe landing technique. Heel to toe landing resulted in a higher first peak VGRF and a lower second peak VGRF than the forefoot landing technique, although heel to toe landing resulted in the highest peak VGRF. The two techniques resulted in power differences during landing with the forefoot technique utilizing mainly the knee and ankle during the entire landing phase and the heel to toe technique utilizing the hip and knee during the beginning of landing and the knee and ankle during the latter portion of landing (Kovacs et al., 1999). The author noted that foot position at contact redistributes the contributions of lower extremity joint power production and likely results in greater energy production in forefoot landing due to strain energy stored in the plantarflexor muscle group (Kovacs et al., 1999).

The interest in lower limb stiffness was later investigated to determine if ankle-focused landing strategies influenced landing biomechanics (Self & Paine, 2001). In this study, four landing strategies were utilized to determine the role of ankle plantarflexion at initial contact, combined with a flexed or stiff knee, on ankle kinematics and kinetics during landing. Self and Paine (2001) reported that a stiff-knee combined with a near flat-footed landing strategy resulted in the highest peak VGRF and tibial acceleration. An interesting fact was that a stiff knee combined with increased plantarflexion landing strategy resulted in similar peak VGRF and tibial acceleration values as the landing strategy using natural knee flexion combined with natural plantarflexion (Self & Paine, 2001). Although the author reported that bending the knee places the Achilles tendon at a disadvantage for energy dissipation, low knee flexion is a mechanism for ACL injury and increases ACL strain due to the pull of the gastrocnemius muscle, the primary muscle involved in plantarflexion (Boden et al., 2000; Fleming et al., 2001).

Two methodological concerns have arisen regarding drop landings, with the first regarding lower extremity bilateral symmetry. All but one of the investigations discussed up to this point have utilized testing protocols where participants perform a double-leg drop landing, a drop landing where both feet contact the ground simultaneously. However, several of the studies utilized a single-leg analysis (Devita & Skelly, 1992; Dufek & Bates, 1990; Lees, 1981), one foot on the force platform and the contralateral foot on the ground next to the platform, and two of the studies had participants land with both feet on the platform (McNitt-Gray, 1991; McNitt-Gray, 1993). Schot et al. (1994) investigated whether this approach was tactful and determined if bilateral symmetry between limbs existed during the performance of drop landings. The authors stated that asymmetries between limbs for ground reaction forces were smaller than joint moment asymmetries, but ground reaction force asymmetries were 3 times more likely to occur than joint moment asymmetries (Schot et al., 1994). While this research warrants the assessment of bilateral symmetry while collecting data, this assessment is limited to a laboratory equipped with two force platforms.

The second methodological concern has to do with studies utilizing a platform to initiate performance of drop landings. It is suggested that protocols involving a platform in the task protocol likely results in erroneous drop landing heights (Kibele, 1999). The author reported that platform heights and the actual height a participant falls will vary based on the manipulation of the body's COM. This error is likely to result when using a platform for drop landing initiation, however, utilizing a bar suspended from the ceiling has been reported to result in increased reliability of drop landing heights (Kernozek et al., 2005).

Drop Landings: ACL-Related Literature

A majority of the biomechanical research on drop landings during the past decade concentrated on mechanisms associated with ACL injury with the primary focus on determining gender differences. Concentration on biomechanical variables, such as, ground reaction forces, joint kinematics, and joint kinetics has led to this line of research in hopes of better understanding mechanisms predisposing females to an increased incidence of ACL injuries.

In vivo investigations of the forces produced on the ACL during landing provides invaluable information about what the ACL undergoes during landing. However, this information proves elusive as implanting a force transducer in an intact knee is invasive, difficult to perform, and the accuracy of this technique may not be worth the risk of infection and medical costs.

However, researchers have taken a step toward better understanding ACL mechanics by modeling the lower extremity during landing (Pflum et al., 2004). Utilizing data from a single participant (28yr, 180cm, 82kg) performing drop landings, Pflum et al. created a forward dynamic model to determine the forces acting on the ACL during landing. The model predicted that the maximum force applied to the ACL during landing was 253 N and occurred 40 ms after initial contact. The authors disproved that the anterior tibial loading due to the shear component of the patellar tendon force could solely induce an ACL injury. Instead, they suggested that the maximum force applied to the ACL was due to the interaction of three forces, including the patellar tendon force, tibiofemoral compressive force, and, indirectly, the resultant ground reaction force.

However, these findings should be viewed with caution, as the model was reported to have several limitations (Pflum et al., 2004) including use of a knee modeled as a 1 degree of freedom hinge joint. The effect of the resultant ground reaction force contributing to the maximum force applied to the ACL also warrants caution as this methodological technique was previously reported to be inaccurate (Winter, 2005). Furthermore, the role of the quadriceps in inducing ACL injury via the pull of the patellar tendon is debatable (DeMorat et al., 2004). However, the proposed idea regarding tibial slope contributing to forces transmitted to the ACL is interesting and warrants further investigation (Pflum et al., 2004).

Simulation experiments using cadaveric knees allowed researchers to investigate the amount of strain the ACL undergoes during a movement associated with ACL injury, drop landings (Withrow et al., 2005; Withrow et al., 2005). Withrow et al. (2005a, 2005b) conducted two separate experiments to investigate the effect of hamstring tension and valgus loading on ACL strain during simulated drop landings. In the first study, peak ACL strain was lower during simulated landings when the hamstrings were pre-tensioned before impact, compared to no hamstring pre-tension prior to impact. The results of the second study were that a combination of a flexion and valgus moment at the knee during simulated landings resulted in increased ACL strain compared to a flexion only moment.

The effect of maturation on the biomechanics of drop landings has previously been investigated (Hass et al., 2005; Hewett et al., 2004; Swartz et al., 2005). Differences in ground reaction forces due to maturational level were demonstrated with pre-pubescent females exhibiting increased first peak VGRF and time to second peak VGRF, compared to post-pubescent females (Hass et al., 2005). Swartz et al. (2005) also reported maturational differences with prepubescent males and females exhibiting increased peak VGRF and decreased time to peak VGRF, compared to post-pubescent males and females.

Knee flexion at contact was different based on the maturational level of females with pre-pubescent females exhibiting greater values than post-pubescent females (20° and 15°, respectively) (Hass et al., 2005), but both groups of females utilized a knee joint angle linked as a mechanism of ACL injury (Boden et al., 2000). While Swartz et al. (2005) reported that maturational level did not influence knee flexion at contact, knee flexion angles at initial contact, for both pre- and post-pubescent females and males, were less than those reported by Hass et al. (2005). However, the pre-pubescent males and females in the investigation of Swartz et al. (2005) had less hip flexion at contact and less knee and hip flexion at peak VGRF, compared to post-pubescent males and females. The pre-pubescent group also demonstrated increased knee valgus angles at initial contact and at peak VGRF (Swartz et al., 2005). The results of Hewett et al. (2004) contradict the findings of Swartz et al. (2005) in that the post-pubescent females exhibited larger valgus angles at initial contact and max valgus angles, compared to pre- and early-pubescent females and post-pubescent males. Although the findings regarding valgus angle are inconsistent across these studies, it remains that a valgus loading is associated with increased strain in the ACL (Shin et al., 2005; Withrow et al., 2005).

Hass et al. (2005) were the only group of those listed that investigated maturational effects on extensor and abduction/adduction knee moments. Pre-pubescent females exhibited a peak extensor moment higher than that produced by the post-pubescent females. There was no maturational influence on the peak abduction/adduction moment produced at the knee during landing (Hass et al., 2005).

Gender comparisons constitute the remaining parts of this discussion, examining differences between males and females for ground reaction forces, lower extremity kinematics, joint moments, joint power and work, and muscle activation. It should be noted that one of the studies in this discussion was a prospective study of 205 females where the investigators utilized baseline measures and injury incidence during competitive soccer, basketball, and volleyball seasons to predict ACL injury (Hewett et al., 2005).

Gender differences in VGRF are inconsistent across the studies investigating factors that may predispose females to ACL injury. Differences were found between genders, with females exhibiting increased VGRF (Hewett et al., 2005; Kernozek et al., 2005; Salci et al., 2004) and decreased A-P GRF (Kernozek et al., 2005). However, other investigators reported that no statistical differences in VGRF were demonstrated between genders (Decker et al., 2003; Ford et al., 2003; Lephart et al., 2002). The study conducted by Lephart et al. (2002) utilized a single-leg landing protocol from a 20 cm platform, while the other studies resulting in no effect between genders for VGRF utilized task protocols similar to those used in the studies where gender differences were reported (double-leg landing, 31-60cm platform or suspended bar).

Lower extremity joint angles at initial contact and during landing have also been inconsistent across studies. Gender differences did not exist for knee flexion angle at initial contact (Kernozek et al., 2005) or during landing (Kernozek et al., 2005; Urabe et al., 2005), however, other researchers have reported gender differences in knee flexion angle at initial contact (Decker et al., 2003; Hewett et al., 2005) and during landing (Lephart et al., 2002; Salci et al., 2004). Ford et al. (2003) reported that females demonstrated increased peak valgus angles, total valgus motion, and valgus in the dominant limb. The former result was supported by the work of Hewett et al. (2005) and Kernozek et al. (2005) and the increased valgus range of motion was also supported by Kernozek et al. (2005). Another important finding was decreased hip flexion in females, compared to males (Salci et al., 2004), which may be a strategy utilized by females predisposing them to ACL injury (Ball et al., 1999).

Knee extensor moments produced during landing did not support a gender effect. No differences between males and females were reported by Decker et al. (2005), and Salci et al. (2004) reported a difference for one of the four conditions they tested. The difference might be explained due to the inclusion of joint friction estimations when calculating joint moments (Salci et al., 2004). In the study investigating females incurring an ACL injury during the season, it was reported injured females exhibited increased peak knee abduction moments (Hewett et al., 2005), which is supported by Kernozek et al. (2005) where it was reported that females demonstrated a decreased adduction moment, compared to males. While it was not an abduction moment, the decreased adduction moment can be interpreted as leaning towards an abduction moment. Also of interest from the work of Hewett et al. (2005) was that the difference in abduction moments between limbs was 6.4 times greater in the injured females and knee abduction moments and angles were significant predictors of ACL injury.

Joint power and work was investigated by only one of the authors listed in this discussion (Decker et al., 2003). Females produced more peak knee and ankle power than males and the knee and ankle absorbed more energy than the hip in females. Work produced by the hip () was lower in females than males, while ankle and knee work was higher in the females, compared to males.

Neuromechanical Responses to Perturbations

Perturbations and unanticipated events occur during sports participation and are common in basketball, soccer, and netball. A perturbation is defined as a secondary influence on a system causing it to deviate slightly; a small modification in a physical system (Stein, 1970). Accordingly, interaction between athletes while falling towards the ground is termed an in-flight perturbation. Unanticipated events are defined as events occurring during sports participation for which athletes cannot prepare neuromuscular strategies, hence, creating biomechanical responses to adequately prepare for and protect tissues from these events are not always possible. In-flight perturbations and unanticipated events are usually the same in one, in that the effects of in-flight perturbations are typically unanticipated events. At this time, research examining the effects of unanticipated perturbations on non-drop landing movements (Pavol & Pai, 2002) and unanticipated drop landings (Fu & Hui-Chan, 2002) exists. At this time however, investigations of the effects of in-flight perturbations on drop landing performance do not exist.

Research of unanticipated perturbations during human movement provides important insight into the neuromuscular and biomechanical responses to such events. Utilizing two low-friction platforms, Pavol and Pai (2002) investigated the effects of an unexpected slip when rising from a seated position to determine if unanticipated events elicited a change in the anticipatory control used by individuals. Several interesting results were reported including changes in anticipatory control based on the condition last experienced, such as, after induced slips participants began to shift the COM further anteriorly at seat-off (no longer touching the seat) to prepare for a potential perturbation (Pavol & Pai, 2002). Repeated exposure to a condition led to a fixed response within two trials, indicating that anticipatory control strategies were adjusted within two exposures to a condition and remained the same during subsequent exposures to the same condition. In addition, the authors noted that over all trials participants utilized an anticipatory control strategy so that a loss of balance was minimized, regardless of the condition (Pavol & Pai, 2002).

While the previous research provides insight into biomechanical responses to unanticipated perturbations during a sit-to-stand movement, whether these same responses occur during drop landings is more crucial for understanding ACL injury. Using a safety harness suspended from the ceiling held by an electromagnet, Fu and Hui-Chan (2002) compared ankle muscle activation patterns and ground reaction forces during anticipated and unanticipated drop landings. The authors reported during anticipated drop landings onset time of muscle activation for the medial gastrocnemius and tibialis anterior occurred sooner and later, respectively (Fu & Hui-Chan, 2002). The first peak VGRF was 17% lower during anticipated drop landings, compared to unanticipated drop landings, however, results of the maximum VGRF were not reported. Upon visual inspection of force-time curves from a single participant, it appears that unanticipated drop landings resulted in increased maximum VGRF, compared to anticipated drop landings (Fu & Hui-Chan, 2002). While this information provides insight about unanticipated events involving drop landings, there is still a lot of information required for a more complete understanding of the effects of in-flight perturbations, especially those encountered during sports participation leading to ACL injury (Olsen et al., 2004).

Summary

Due to its unique structure, the entire ACL is not recruited when loaded possibly leading to excessive loading to one of the bundles or specific fibers of the ACL. This non-uniform loading may predispose a bundle or the entire ligament to tearing or rupture depending on the biomechanics associated with a particular movement, such as landing. Landing is reported as one of the more common mechanisms for ACL injury, and is partly due to the high eccentric actions of the quadriceps required to prevent knee joint collapse. Investigations of differences between genders when performing drop landings have been mostly inconsistent with the exception that valgus moments and angles are the best predictors of ACL injury. Research on perturbations and unanticipated events is scarce especially concerning drop landing movements. Therefore, the role of unanticipated perturbations during the flight phase of drop landings may provide insight into mechanisms of ACL injury encountered during sports involving these events.

CHAPTER 3

THE EFFECT OF IN-FLIGHT PERTURBATIONS ON LANDING BIOMECHANICS: I. LINEAR PERTURBATION[1]

[1]S.W. Arnett, Y. Fu, R. Thompson, P. Sigurdsson, & K.J. Simpson. To be submitted to *Journal of Applied Biomechanics*.

Abstract

The exact mechanisms occurring at the time of an ACL injury are still elusive at this time. Although this type of injury is not gender specific, it occurs more prevalently in females, which has led to research on risk factors within the individual. Personal factors, such as anatomical, hormonal, and biomechanical, have been implicated in increasing the risk of incurring an ACL injury during sports participation. More recently, the focus of ACL injury research has been extended to include the interaction of biomechanical and environmental factors. One such interaction is an in-flight perturbation. Although these events have been implicated in ACL injuries, the effects of these events on drop landing biomechanics are not known at this time. Therefore, the purpose of this study was to determine the effect of a linear in-flight perturbation on landing biomechanics.

Thirteen college-aged female soccer and basketball athletes performed drop landings with and without in-flight perturbations. Three dimensional ground reaction forces and lower extremity joint kinematics and kinetics were analyzed. Paired t-tests were used for statistical analyses ($\alpha = 0.05$).

Compared to the non-perturbed condition (CON), peak vertical ground reaction force (VGRF) was decreased during the perturbed condition (PERT). No significant differences were demonstrated for peak posterior ground reaction force (GRF) or lower extremity joint kinematics between conditions. Peak hip and knee extensor moments and peak plantarflexor moments were significantly greater during the PERT compared to the CON condition.

The significant differences found between conditions for peak vertical ground reaction force and moments are insightful and at the same time raise questions. The difference in peak VGRF could be interpreted as anticipation by the performer leading to an altered landing strategy. During the PERT compared to the CON condition, increased force could be placed on the ACL due to the shear force created by the peak knee extensor moment. The individual variation occurring at the knee joint between conditions supports that individuals utilized different strategies when landing and this variation could possibly be linked to increased risk for an ACL injury. While all of the hypotheses were not supported by the data, it remains that an in-flight perturbation does influence landing biomechanics.

Keywords: landing, ACL, perturbation, joint kinetics

Introduction

In the United States, ACL injuries occur to 1 out of every 3000 persons, resulting in approximately 100,000 ACL injuries per year (Childs, 2002). The incidence of these injuries combined with rehabilitation costs (surgical reconstruction and/or intensive physical therapy) results in an annual cost of approximately $1.5 billion (Childs, 2002). Even more alarming is that ACL injuries of U.S. female athletes account for approximately 38% of the nation's total ACL injury rate (Toth & Cordasco, 2001), with approximately $646 million spent for treatment of high school and collegiate female athletes (Childs, 2002). Unfortunately, females are 2 to 10 times more likely to experience an ACL injury compared to males participating at the same level of sport (AAOS, 2003). However, the mechanisms underlying the increased incidence of ACL injuries in females are not fully understood.

Comparing genders to elucidate factors contributing to the increased incidence of non-contact ACL injuries in the female population has been a popular focus of non-contact ACL injury research (Decker et al., 2003; Ford et al., 2003; Hass et al., 2005; Hewett et al., 2004; Hewett et al., 2005; Kernozek et al., 2005; Lephart et al., 2002; Salci et al., 2004; Swartz et al., 2005). The major risk factors associated with an increased incidence of non-contact ACL injuries in females are anatomical, hormonal, and biomechanical/functional (Boden et al., 2000; Childs, 2002; Griffin et al., 2000; Markolf et al., 1995). Anatomical risk factors consist of a congenitally narrow femoral intercondylar notch width (Hutchinson & Ireland, 1995) or a "large" Q angle (Shambaugh et al., 1991). Increased concentration of estrogen and progesterone, such as occurs during the menstrual cycle, contributing to knee joint laxity (Heitz et al., 1999) is a hormonal risk factor. A functional factor related to the performer occurs after an unanticipated event, with an overcorrection in anticipatory control during the subsequent movement leading to a high risk situation (Pavol & Pai, 2002).

However, having one of the above risk factors does not guarantee the occurrence of a non-contact ACL injury. Non-contact ACL injuries are reported to have a multi-faceted etiology (Hewett et al., 2006) and are not entirely specific to an age group, gender, or sport.

Among the etiological factors thought to be related to ACL injury, biomechanical movement techniques associated with non-contact ACL injuries have been identified. These injuries most commonly occur when a person is reacting to a perturbation, landing, pivoting, or decelerating the body (Fleming, 2003). Non-contact ACL injuries are more likely to occur when the knee is close to full extension (Boden et al., 2000) and rapid body deceleration with the foot fixed to the ground are components often involved during non-contact ACL injuries (Boden et al., 2000). Observing that the proximity of an opposing player might lead to a non-contact ACL injury, Boden et al. (2000) proposed that the injured athlete's movement execution may be influenced by the opposition's proximity. Additionally, knee valgus collapse and backward lean of the trunk following quick decelerations and single-leg landings were implicated in the etiology of a majority of non-contact ACL injuries occurring during competition (Boden et al., 2000).

The biomechanics used during a successful landing load the ACL. During landing, the knee extensor muscles must produce an eccentric action in order to prevent knee joint collapse about the flexion/extension axis (Devita & Skelly, 1992). This eccentric action produces an anterior shear force on the proximal tibia when the knee is almost fully extended (Boden et al., 2000), which could possibly lead to an ACL tear (DeMorat et al., 2004). Boden et al. (2000) also believe that imbalances between muscle forces and mechanical loading of the knee may occur due to unanticipated conditions encountered during sports participation, such as perturbations.

It has been reported that a perturbation is likely one of multiple factors involved in ACL injuries (Hewett et al., 2006; Krosshaug et al., 2007). In-flight perturbations create three of the four most common movement events associated with non-contact ACL injuries: landing, reacting to a perturbation, and rapid deceleration of the body (Fleming, 2003). However, at this time, the quantitative effects of in-flight perturbations on landing biomechanics are not known.

In many sports involving jumping and landing (e.g., basketball rebounding, soccer heading, netball goal), athletes collide in the air with another athlete (or a stationary object). This type of collision often results in an external force applied to the player of interest. This force influences or 'perturbs' the player's motion, potentially leading to an unstable landing. The point of application that the perturbation force is applied determines the resulting change in momentum. If the in-flight perturbation is applied at the body's COM it causes an increase in linear momentum ('linear in-flight perturbation'). Therefore, for this study, from a mechanical perspective, a linear in-flight perturbation (PERT) is an external impulse applied at the body's COM, which leads to a change in linear momentum.

As predicted by rigid-body Newtonian mechanics, only external forces, such as those producing in-flight perturbations can affect the projectile motion of the body. Consequently, during the flight phase of a movement, the performer cannot alter the projectile path nor alter the total vertical or horizontal linear energy of the body in response to a perturbation. Thus, an individual can only manipulate the effects of an in-flight perturbation during landing, particularly as there is insufficient time to generate internal muscle torques to reposition the body segments to prepare for landing.

According to the impulse-momentum principle, the magnitude of the linear impulse applied to the performer during an in-flight perturbation determines how much the performer's linear momentum increases. Once the perturbation ceases, the performer's new antero-posterior momentum is constant until the performer contacts the ground. Therefore, with unanticipated and increased anterior momentum, the player will contact the ground with greater anterior velocity and at a different angle of approach relative to the ground.

During landing after a PERT, the performer must become stable to land safely. Inadequate stability during landing could lead to abnormal loading of the ACL. Upon contact with the ground, the posterior ground reaction force (GRF) serves to create static friction so the feet remain fixed to the ground. With the feet fixed to the ground, the ankle joint flexion/extension axis becomes a fixed axis of rotation for sagittal plane rotation of the rest of the body. Hence, the linear momentum is transferred into rotational momentum, causing the body to rotate forward. To stop rotating, the performer must create an opposing rotational impulse or else the performer will continue rotating until the center of mass of the body (COM) moves outside of the base of support and loss of balance occurs. Therefore, to counteract the rotational momentum of the body after ground contact, the performer would have to produce increased posterior GRF and increased lower extremity joint moments, compared to a typical landing.

The biomechanics used to cope with in-flight perturbations during the impact phase of landing, therefore, are of interest in this study to understand the implications of these events on landing performance. Consequently, in-flight perturbations are anticipated to require altered landing mechanics compared to typical landings. In addition, understanding these alterations in mechanics may help us understand mechanisms of ACL injury following contact with another person/object during the flight phase of sport movements. Therefore, the purpose of this study was to determine if an in-flight perturbation applied at the body's COM (LIN PERT) results in altered landing biomechanics, compared to the biomechanics produced during a non-perturbed condition (CON). The researchers hypothesized that the PERT compared to the CON condition would result in increased angular displacement, increased peak angles, and increased peak extensor moments for the hip, knee, and ankle. In addition, it was hypothesized that posterior GRF impulse and peak VGRF and posterior GRF would be increased during the PERT compared to the CON condition. Peak knee valgus angles and moments were not expected to vary by more than .5 SD for the PERT compared to the CON condition.

Methods

Design

The design of this study was quasi-experimental and the independent variable was the perturbation (PERT) applied to the person. For this study, the initial block of drop landing trials performed (prior to any perturbation trials) represents the control condition (CON).

Participants

Participants consisted of 13 college-aged, recreationally active females. Participants were recruited from the Department of Kinesiology's undergraduate major and Basic Physical Activity Program courses via listservs and within the University of Georgia (Athens campus) by posting flyers. Participants must have participated in competitive level (varsity or recreational league) basketball or soccer in one of two ways: 1) participated within the last year or 2) participated within the last five years and performed 30 minutes of moderate aerobic activity (i.e., walking, jogging, cycling) at least 3 days/week for the past six months. Participants were excluded from the study based on the following criteria: current lower extremity injury (e.g., ankle sprain, patellar tendonitis, torn meniscus); lower extremity malalignment (assessed by physical therapist); taking medications affecting performance or safety (e.g., pain medication, blood pressure medication, cough syrup containing alcohol); previous lower extremity, back, or head injury within last year; diagnosed cardiovascular or pulmonary condition (e.g., heart murmur, hypertension, chronic obstructive pulmonary disease); any visual, vestibular, neurological, or other problems affecting balance; or performed lower extremity resistance training within the past 48 hours. Participants were randomly assigned to this or a second, similar study. Using an adjusted effect size for a repeated measures statistical test (Lipsey, 1990) on outcome variables from an initial subset of the participants in this study, a sample of 13 participants was predicted to elicit adequate power (0.80) for most of the outcome variables. Three biomechanical variables (kinematic) needed a sample size in excess of 100 to obtain adequate statistical power.

Drop Landing Procedures

Test Task:

One drop landing task was performed for each trial. To execute the task, the participant began each trial by hanging from a metal bar (Figure 3.1) set for a drop height of 0.60 m from the distal portion of the fibula to the ground. The participant then released the bar when ready, dropped down, and landed naturally with the right foot contacting a force platform and maintained that position for a minimum of 2 seconds. Throughout the task, the participant maintained an arm position where the arms remained overhead during landing.

A proprietary device called "BIFIC" (Figure 3.1) was used to create in-flight perturbations. This device consists of: a) a "start bar" consisting of a 0.61m long load cell coated in a rubber polymer attached to the superior surface of the drop bar; and b) an apparatus utilizing an air piston system that applied a short anterior impulse to the participant via a steel cable attached to a belt worn by the participant. The time the force was applied to the performer was constant and the force magnitude (1.15 x body mass) was adjusted using an air regulator (R21-04-000, Wilkerson Co., Pneumatic Division, Richland, MI). During a PERT trial when the participant released the drop bar, the perturbation process was activated when force applied to the start bar dropped below 13.2 N (3 lbs).

A seven-CMOS 4.1 megapixel, Vicon MX™ camera system and Workstation™ software (v. 5.2.4) (Vicon, Inc., Englewood, CA) were used to capture the spatial locations of reflective markers (sampling frequency = 240 Hz, shutter speed = 1/1000 s). For video acquisition, 47 reflective markers were placed on the participant's skin and

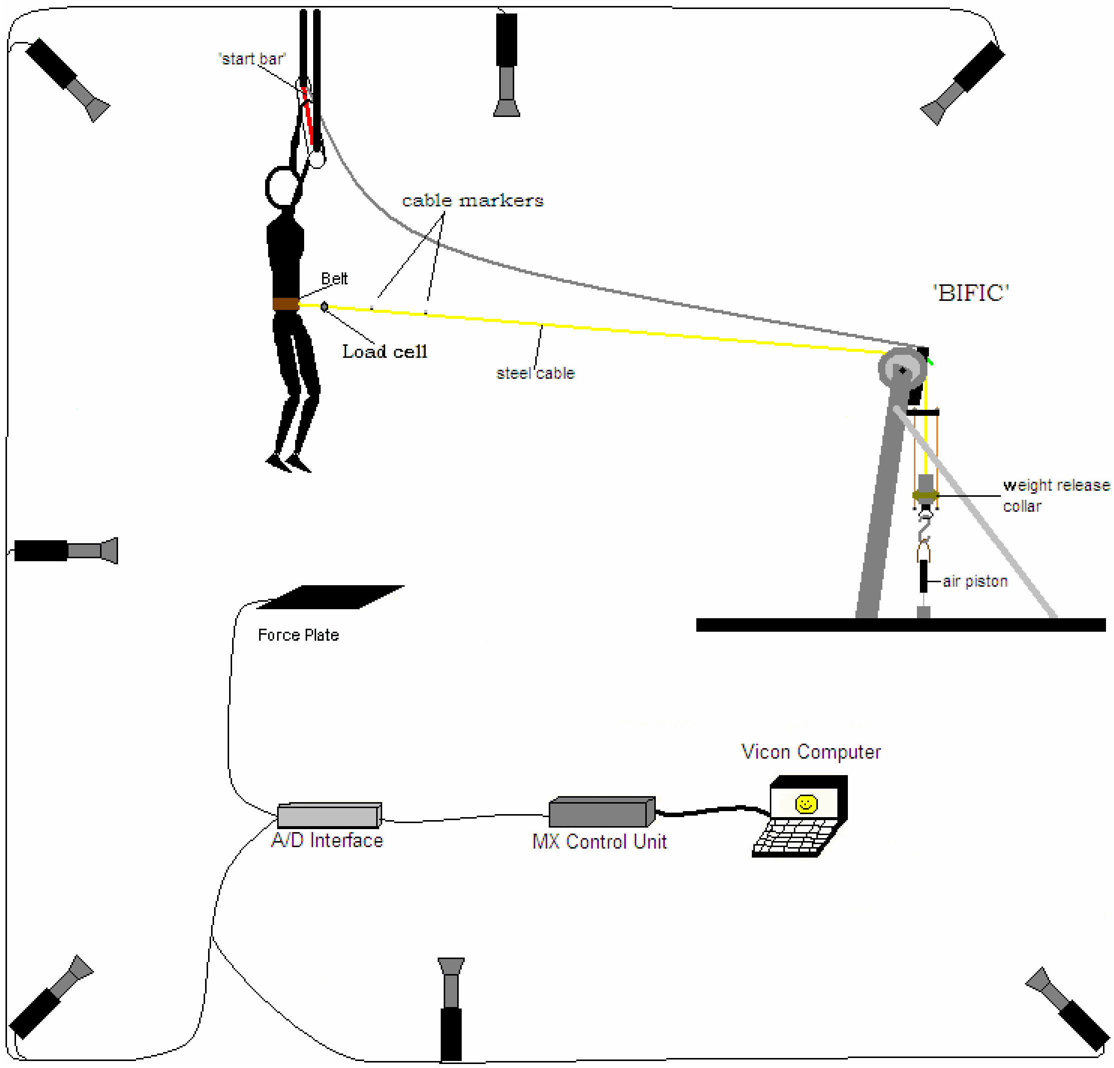

Figure 3.1. Diagram of the experimental set-up used during both studies

The experimental setup is shown in Figure 3.1. Seven cameras, a single force plate, the perturbation device ('BIFIC'), and a load cell were used during this experiment. Below is a detailed description of the instrumentation used in the experiment.

clothing using the designated marker set (Table B1, Appendix B) (Lu & O'Connor, 1999), and two markers were placed on the apparatus cable 0.61 m from the participant. For calibration of the cameras and system, an L-frame (Ergocal – 14mm markers, Vicon, Inc., Englewood, CA) and wand (240 mm wand – 14 mm markers, Vicon, Inc., Englewood, CA) were used for static and dynamic calibration of the testing space, respectively.

A single AMTI force platform (Model OR6-6-1, Advanced Mechanical Technology, Inc., Newton, MA) was used to collect ground reaction force (GRF) signals. Raw GRF signals from the antero-posterior (AP-GRF) and vertical (VGRF) directions were collected (1200 Hz) and amplified (gain = 4000) prior to being saved on a desktop computer.

A load cell (MLP-200, Transducer Techniques, Temecula, CA) was used to collect force signals produced by BIFIC. Raw force signals were collected (1200 Hz) conditioned, and amplified (gain = 1).

A classic reaction board technique (Fig. 3.2) (Hamill & Knutzen, 1995) was used to determine the vertical location of the participant’s COM for a static position similar to the body position that was exhibited during the flight phase. Thus, the belt was properly aligned relative to the performer’s vertical COM location so the perturbation applied to the person during testing was at the correct location. The reaction board, consisting of a rectangular wooden plank (length = approximately 2 m, width = approximately 0.6 m) with conduit legs attached to each end, rested on the force plate on one end and the other end rested on the ground. For more information on calculating the performer’s vertical COM location, see Appendix C.

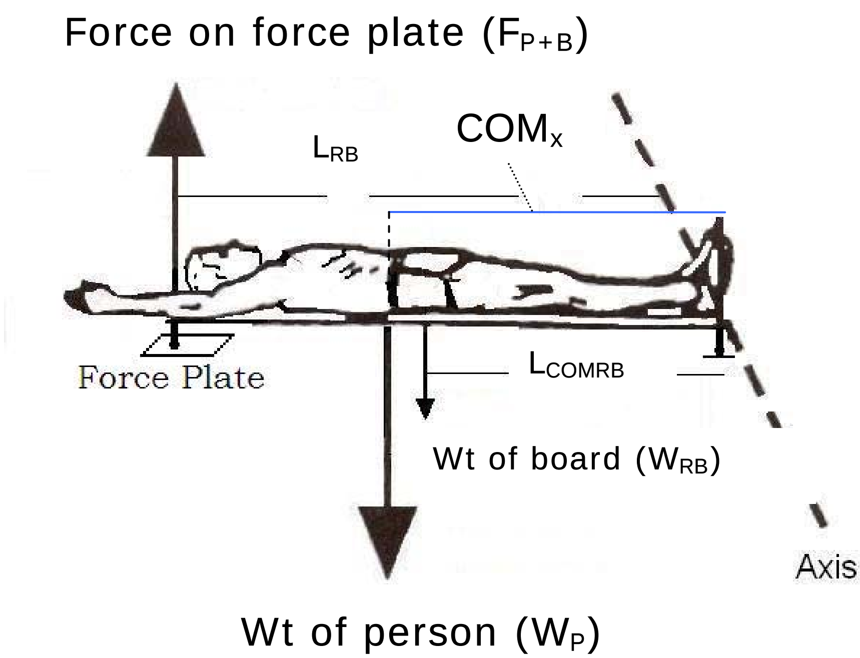

Figure 3.2. The reaction board technique. Diagram modified from http://www.twu.edu/biom/3591Labs/Center%20of%20Mass/Reaction%20Board%20Method.htm. The height of the vertical COM location is when a person is in the body position shown.

Protocol

Pre-drop test procedures:

The potential participant signed the consent form in compliance with regulations of the human subjects institutional review board. Next, the participant filled out a Health Status/Sports Participation Questionnaire (Appendix D). Once eligibility for participation was determined, the participant was randomly assigned to this or a second, related study (Arnett, Fu, Thompson, Sigurdsson, & Simpson, in submission (2007, Journal of Applied Biomechanics) Anthropometric measures were obtained using Zatsiorsky's (2002) methods: body mass, height; lengths and circumferences of body segments. As wearing laboratory shoes would likely affect performance more than wearing one's own shoes, participants wore their own running shoes.

Next, the participant was asked to lie on the reaction board to determine the vertical height of the body's COM (Hamill & Knutzen, 1995). It was assumed that at the instant the perturbation was applied, the performer was in a position in which all lower extremity joints were close to full extension, with the arms at 180° of shoulder flexion. Thus, the participant was asked to lie on the board with their hands stretched overhead about shoulder width apart, mimicking the body's position during the drop landing trials. The participant remained motionless while the force measurement was collected, which was used to calculate the vertical COM location.

After the reaction board technique, reflective markers were affixed to the participant's skin and clothing. Following marker placement, participants performed a 5-minute warm-up consisting of stationary cycling and light stationary hopping. The participant wore a Velcro™ fastened Altus Contour Weightlifting Belt® (Altus Athletic

Manufacturing Company, Inc., Altus, OK) as tight as could be tolerated during data collection. The midpoint of the vertical height of the belt was aligned relative to the body's vertical location of the COM obtained during the reaction board test.

Drop Test Procedures:

Participants performed between two to five practice trials of double-leg drop landings without and then with in-flight perturbations while attached to the apparatus. The impulse applied to performer during practice in-flight perturbation trials was 75 to 82% of the impulse applied during actual testing.

The participant was instructed that an in-flight perturbation may or may not occur during each test trial. After performing each trial, to minimize fatigue of the leg musculature, the participant rested for 15 to 20 seconds before testing resumed. Data were collected for two separate blocks of trials. The first block consisted of five acceptable trials of non-perturbed drop landings (control = CON). The second block consisted of five acceptable trials each of perturbation (PERT) and CON trials performed in a quasi-random order. To prevent participants from using anticipatory strategies, no more than two trials of the same condition were performed consecutively (Pavol & Pai, 2002). If a non-acceptable trial occurred during data collection, the participant repeated it after the remaining trials of the current trial block were completed. A trial was deemed acceptable when a participant performed a forefoot-to-heel landing, landed with the right foot touching only the force plate and the left foot touching only the ground, did not take a step forward or backward, and maintained foot position for at least 2 s after contacting the ground.

Data Reduction

The phase of interest for this study is the impact phase of landing. For each trial, the beginning of the impact phase occurred when the participant contacted the ground and ended at the minimum vertical ground reaction force (VGRF) value occurring between the first and second peak VGRF's [~100-150ms after initial contact]. See Appendix E for more detail on the definition of the impact phase.

For the kinematic data, the spatial locations of the reflective markers were reconstructed using Vicon Workstation® software (v. 4.3.1) and filtered using Woltring's generalized cross-validated spline (GCVSPL) algorithm (Woltring, 1986). A 5-link (foot, shank, thigh, pelvis, and trunk [thorax + abdomen]) rigid-body model was used for kinematic and kinetic analyses. Kinematic and biomechanical quantities (linear displacement, velocity, and acceleration; joint angles, angular velocity, and angular acceleration) were calculated using MatLab® software (v. 7.0). A Cardanic rotation sequence was used to calculate lower extremity joint angles (Z-Y-X) following the recommendations of the International Society of Biomechanics (Wu & Cavanagh, 1995). The rotation matrix representing the relative relationship between segments was then used to calculate joint angles. The first and second differentiation of the linear displacement data generated linear velocity and acceleration values, and techniques reported by Winter (2005) were used to calculate angular velocity and acceleration. Kinematic variables for the lower extremity joints were: peak angles, the relative times that these events occurred, and flexion displacement.

GRF were calculated using Body Builder™ software (Vicon, Inc., Englewood, CA). Curve analyses of the GRF were performed using in-house software (MatLab® v.

7.0). All GRF magnitude variables were scaled to body mass, and GRF time variables were scaled relative to the impact phase. The vertical ground reaction force (VGRF) variables were peak VGRF and relative time to peak VGRF. The antero-posterior ground reaction force (A-P GRF) variables of interest were: peak posterior GRF, posterior A-P impulse and relative time to peak posterior GRF. Peak vertical and antero-posterior ground reaction forces (VGRF and A-P GRF, respectively) were analyzed to test the hypotheses that the perturbation leads to increased maximum force applied in the vertical and antero-posterior directions. Posterior A-P impulse was compared between conditions to confirm that after a perturbation, during landing, greater posterior impulse must be applied to the performer.

Joint kinetics of the right ankle, knee, and hip joints were calculated using 3-D inverse dynamics (Winter, 2005). COM locations and moments of inertia for each of the lower extremity segments were determined using Zatsiorsky's anthropometric model for females (Zatsiorsky, 2002). Joint kinetic variables were peak magnitudes and the relative time to these events. Joint moments were scaled to body mass and height.

All GRF and lower extremity joint kinematic and kinetic variables are reported for the right side and all variables occurred during the impact phase of landing. Relative time (Rel-T) to selected events was calculated as a percentage of the total time of the impact phase (Rel-T = % Tot-T).

The assumptions for parametric t-tests were met, therefore, paired t-tests were used to test for differences between the CON and PERT conditions for the variables previously identified (alpha = .05). Adjusted effect sizes for paired observations (ES_{ap}) (Lipsey, 1990) were generated to determine behavioral significance.

Results

Participant characteristics are reported in Table 3.1. GRF variables are presented in Figures 3.3-3.6. See Figure E1 in Appendix E for representative GRF-time curves (Participant #3). The peak VGRF (Figure 3.3)was 1.2 N/kg lower during the PERT compared to the CON condition ($p = 0.04$) (COV = 11.2, CON; 12.8, PERT). No other significant differences were detected for GRF variables ($p>0.05$) with the coefficient of variation ranging from 42.2-121.8. Observed power for GRF variables ranged from .18-.59.

Kinematic variables about the medio-lateral axis for the hip, knee, and ankle (flexion/extension) and about the antero-posterior axis for the knee (add/abduction) are presented in Figures 3.7-3.12. See Figure F1 and F2 in Appendix F for representative joint angle-time curves (Participant #3) for CON and PERT landings, respectively. There were no significant differences between conditions for lower extremity kinematic magnitude or time variables.

Joint kinetic variables about the medio-lateral axis for the hip, knee, and ankle (flexion/extension) and about the antero-posterior axis for the knee (add/abduction) are presented in Figures 3.13-3.14. Peak hip and knee extensor moments and peak plantarflexor moments were significantly greater during the PERT compared to the CON condition. The peak plantarflexor moment occurred first followed by the knee and hip extensor moments.

Discussion

The researchers hypothesized that landing after an in-flight perturbation would require increased peak VGRF and posterior GRF, posterior A-P impulse, peak joint

Table 3.1. Participant characteristics

	Mean ± SD	Range
Age	20.3 ± 1.9	18-24
Body Mass (kg)	62.8 ± 8.2	52.5-79.4
Body Height (cm)	167.9 ± 7.3	159.6-187.7
No. yr participation		
Soccer	8.2 ± 6.0	<1-18
Basketball	7.2 ± 3.6	4-12

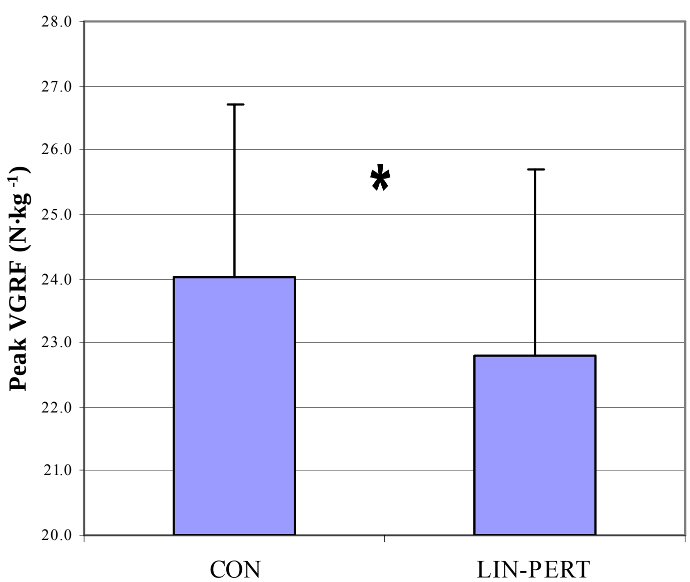

Figure 3.3. Peak VGRF for control (CON) and perturbation (PERT) landings. * indicates significant difference between conditions at $p < 0.05$.

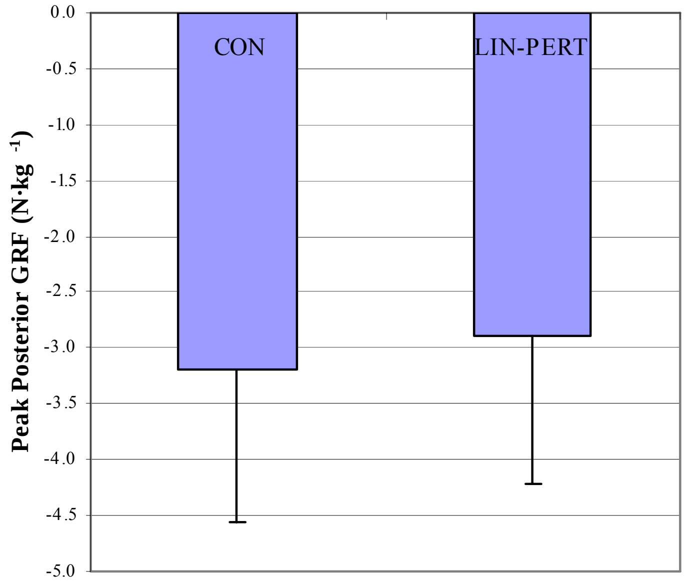

Figure 3.4. Peak posterior GRF.

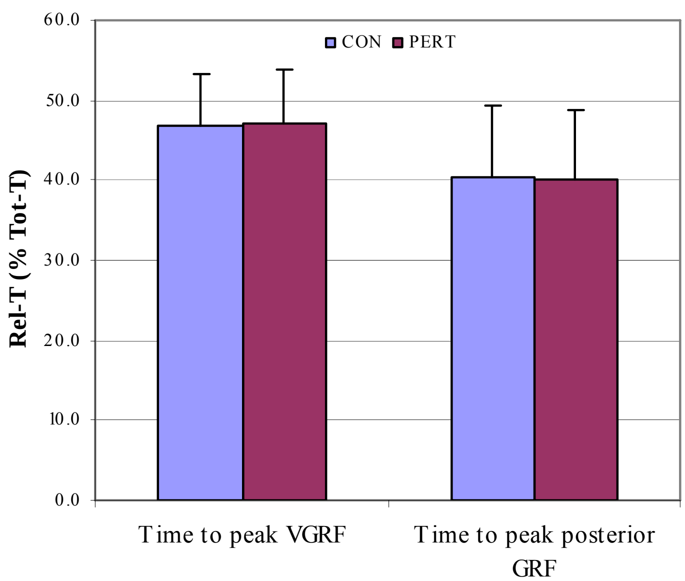

Figure 3.5. Relative times (Rel-T) to GRF.

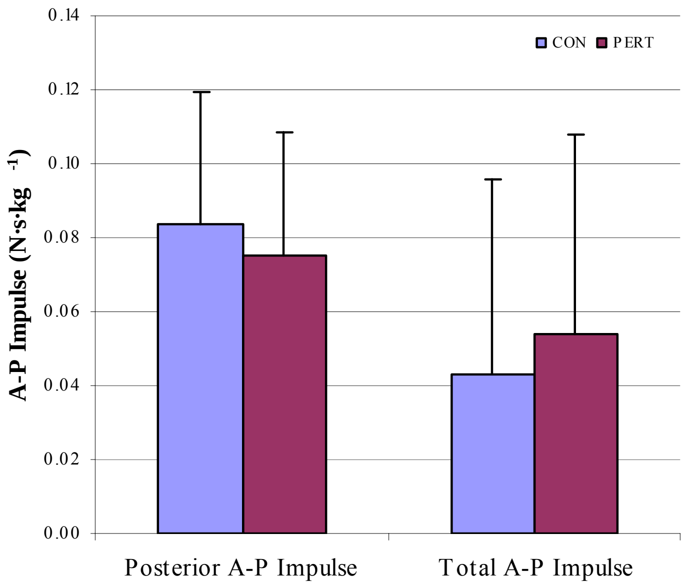

Figure 3.6. A-P impulse variables.

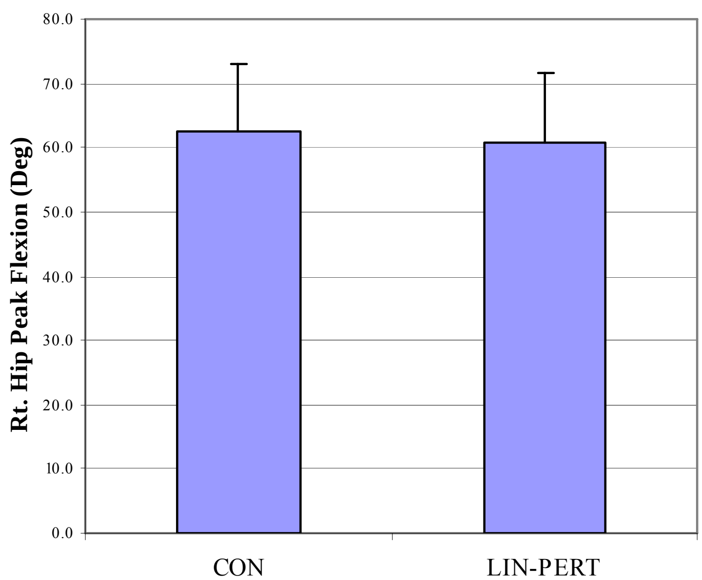

Figure 3.7 Peak hip flexion.

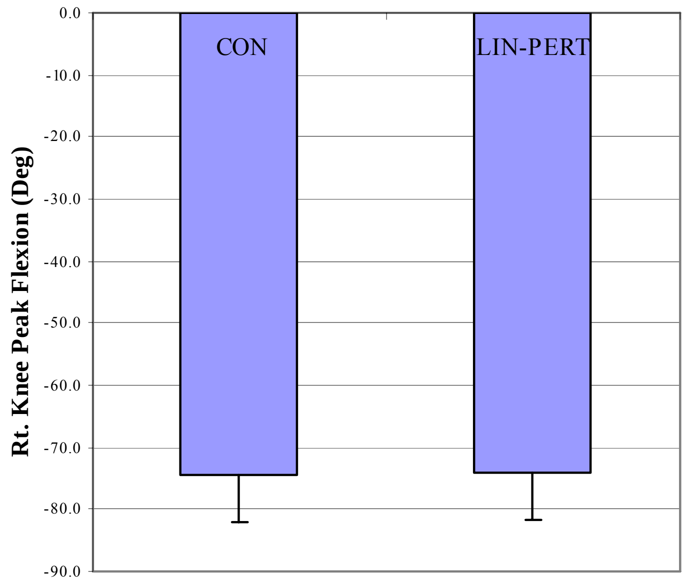

Figure 3.8. Peak knee flexion.

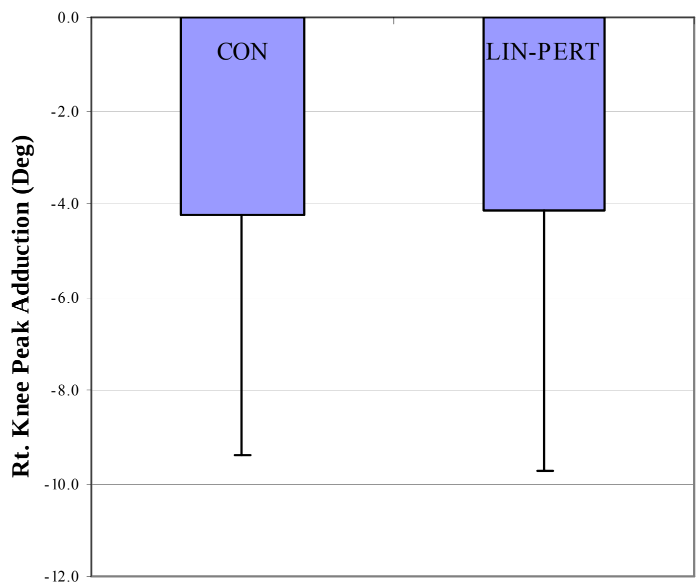

Figure 3.9. Peak knee adduction.

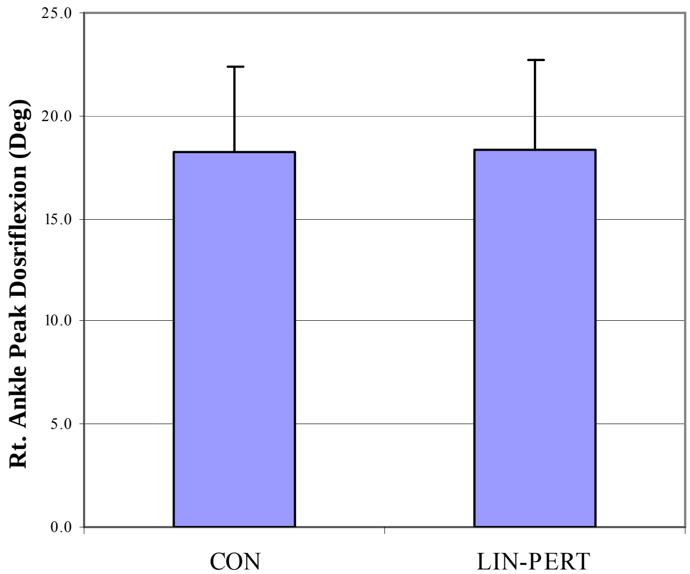

Figure 3.10. Peak ankle dorsiflexion.

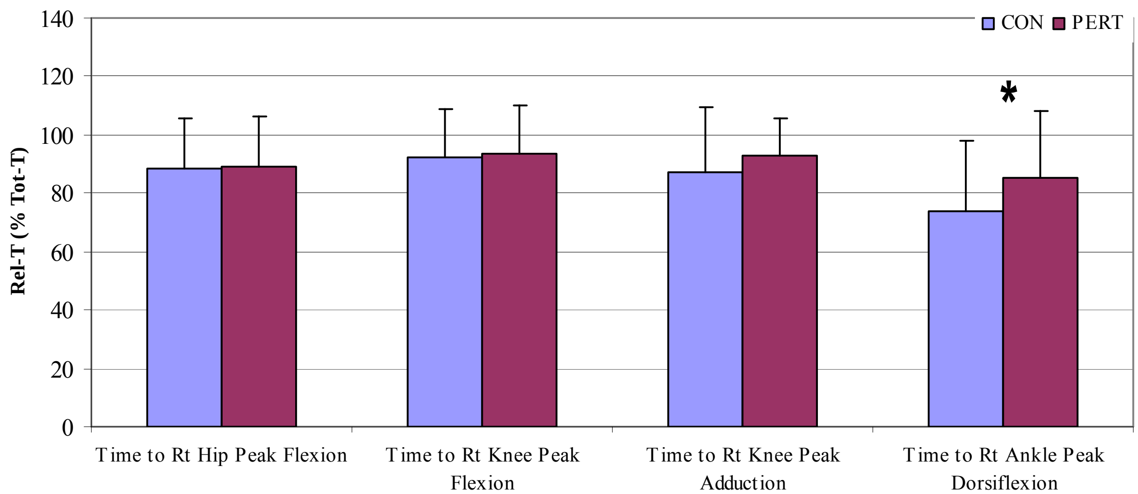

Figure 3.11. Relative time to peak joint angles. * indicates significant difference between conditions at $p < 0.01$.

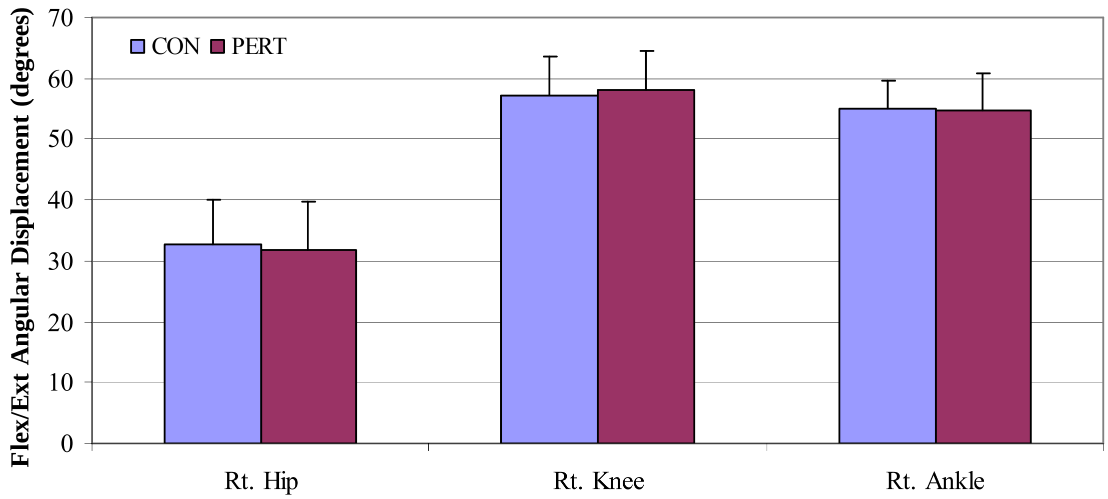

Figure 3.12. Lower extremity flexion angular displacements.

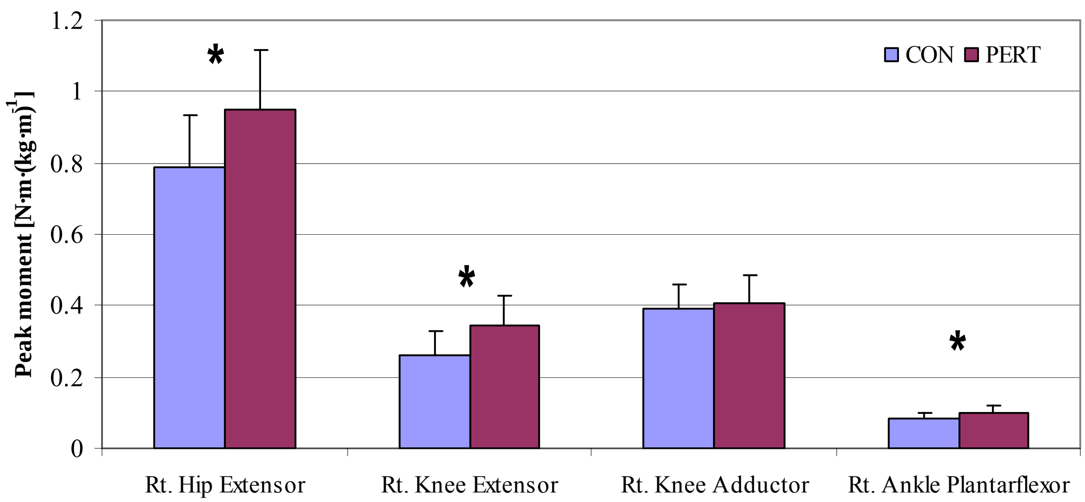

Figure 3.13. Peak joint moments. * indicates a significant difference between conditions at $p < 0.01$.

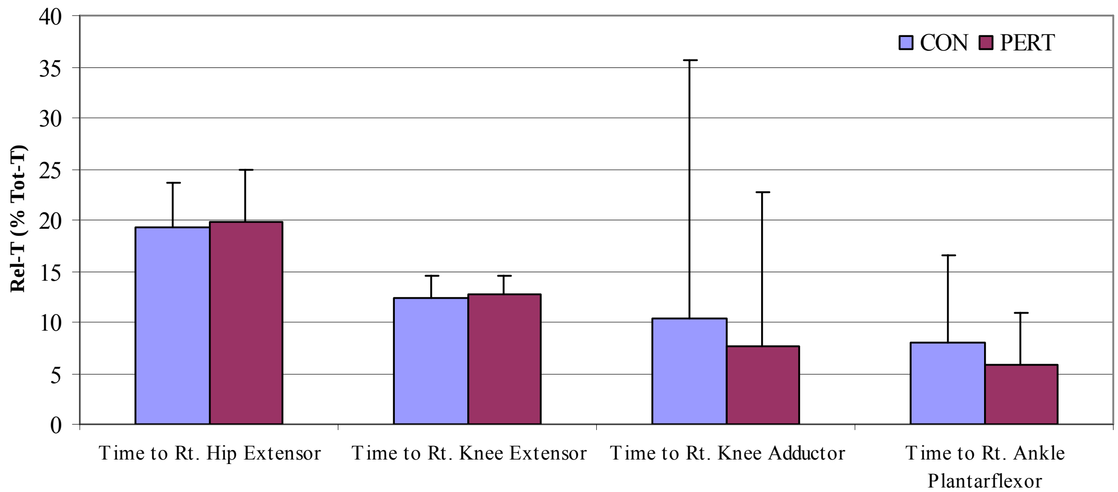

Figure 3.14. Relative time to lower extremity peak joint moments.

angles, and peak extensor moments of the lower extremity joints. Outcomes for peak extensor moments supported our predictions. However, anticipated VGRF posterior GRF, posterior A-P impulse, and kinematic differences between CON and PERT conditions were not identified.

It was surmised that increased extensor moments of the lower extremity would occur during PERT compared to CON landings because of the momentum gained during flight. In turn, if increased joint moments did occur, it could reflect increased mechanical loading on the ACL. Indeed, increased peak net muscle extensor moments of the hip, knee, and ankle were displayed during the PERT compared to the CON landings. These outcomes were believed to be behaviorally significant, as the effect sizes ranged from 7.8 to 12.7 (ES_{ap})(Partial η^2 : 0.69-.086). Thus, the increased extensor moments during PERT landings were crucial in dissipating the body's kinetic energy gained during flight, to reduce the rotational momentum of the lower extremities created by the joint reaction forces after landing began (Simpson & Kanter, 1997), and to prevent joint collapse after ground contact (Devita & Skelly, 1992). If lower extremity flexion cannot be controlled during the initial impact phase (0 ms to 148 ± 29 ms) after an in-flight perturbation, loss of stability to the athlete may occur, increasing the risk of high-impact collision injuries and/or soft tissue damage.

Increased knee extensor moments demonstrated during PERT compared to CON landings may also reflect an increased quadriceps moment, relative to knee flexor (hamstrings) moment. Although not discernible from knee moments alone, it is possible that increased knee extensor moments could reflect increased anterior shear forces acting on the tibia (DeMorat et al., 2004; Simpson & Kanter, 1997). Particularly, as the knee extensor joint moment peaked within the first 12.8 ± 1.8% (19 ms) of the impact phase, which is within the range of time noted when ACL injuries occur after initial contact with the ground (Krosshaug et al., 2007). In addition, these moments occurred when the knee angle was moving through 16 ± 6˚ to 27 ± 9˚. The ACL is more likely to be injured when the knee is close to full extension at initial contact, such as displayed in this study (Boden et al., 2000). Further, the knee adduction joint moment peaked just prior to the peak knee extensor moment (7.6 ± 15%), thus indicating that the knee adductors were also still creating adduction motion during this time. These adduction moments are likely to place stress on the ACL as it is a secondary restraint to knee adduction (Inoue et al., 1987; Whiting & Zernicke, 1998). In addition, knee valgus moments have been implicated as ACL injury mechanisms in female athletes (Hewett et al., 2005; McLean et al., 2005).

It is likely that increased ankle extensor joint moments were generated during PERT landings to counteract the rotational momentum of the body gained during the impact phase due to inertial torque of the body. Assuming a simple model of the body in which the foot is fixed to the ground, according to d'Alembert's principle, when the foot contacts and becomes fixed to the ground, the rest of the body will accelerate in the anterior direction, hence an inertial torque acts on the body about the flexion/extension axis of the ankle joint. Consequently, ankle extensor joint moments would be required to slow down the flexion rotation of the tibia. Otherwise, the person would continue to rotate forward, risking loss of balance.

This same rationale may also apply to the knee and hip joints. As the flexion rotation of a segment (e.g., tibia) slows down, the adjacent, proximal segment (upper leg) continues to move forward, that is, rotate, unless an opposing joint moment acts at their common articulation (knee).

Knee and hip extensor moments also may have contributed to maintaining a relatively large principle moment of inertia of the body about the flexion/extension axis of the ankle, thereby inversely affecting the body's angular velocity. By creating extensor joint moments of the lower extremities, the flexion angular displacements of the lower extremity joints only ranged from 32° (hip) to 58° (knee). Consequently, the flexion/extension axis moment of inertia remained greater than if more flexion had occurred.

Indeed, the angular displacement of the hip, knee, and ankle, respectively, were 33°, 57°, and 55° compared to 32°, 58°, and 55° for the CON and PERT conditions, respectively. Viewing the group analysis of these variables does not support that differences existed between conditions. However, it has been suggested that movement strategies vary by individual and therefore warrant an individual by individual analysis (Bates, 1996; James et al., 2003).

As ACL injury etiology is likely multifactorial (Hewett et al., 2006), it would seem logical that some individuals would be at greater risk than others. As it is also known that landing strategies vary among people (James et al., 2003), understanding individual participant responses therefore is paramount. Thus, a discussion of qualitative observations for differences and similarities in responses to perturbations follows.

The group findings for angular kinematics displayed almost no differences between the PERT and CON landings, likely due to participants exhibiting behavioral differences from one another. Indeed, there were some interesting angular kinematic results demonstrated among the participants. The maximum standard deviation for any angle or angular displacement was < 2˚. Therefore, for each participant, for the PERT and CON means to be different, 2˚ was used as the critical difference. About the mediolateral axis of the knee, 39% of the participants displayed 2˚ or more displacement during the PERT compared to CON condition, while 15% exhibited the opposite (2˚ or more during the CON condition), with the remaining participants exhibiting similar values during both conditions. Furthermore, between the PERT and CON conditions only 8% of participants had 2 ° or more angular displacement about the longitudinal axis of the knee and no participants demonstrated AB/ADD knee displacement > 2˚. Although increased angular displacement does not preclude an individual to an ACL injury, it supports previous findings of inter-individual variation in movement strategies.

Unexpectedly, peak VGRF was significantly lower during the PERT compared to the CON condition (p=0.04). The original hypothesis was that peak VGRF would be greater during the PERT compared to the CON condition. One explanation for this reverse outcome could be that participants were anticipating the in-flight perturbation and adapted their biomechanical strategy by increasing lower extremity flexion to increase the linear displacement of the COM during landing. However, joint kinematic data do not support that this occurred during the impact phase. A second explanation may be due to increased time to maximum ankle dorsiflexion during PERT landings. This strategy may have caused the force applied to the ground to act over a greater period of time, thereby potentially reducing the impact phase peak VGRF.

As there was high inter-participant variability for both A-P GRF variables, no difference between the two conditions were detected for peak posterior GRF or posterior

A-P impulse, even though the statistical power was 0.95. Therefore, further formal testing of individual participant biomechanical response strategies (James et al., 2003) may demonstrate unique participant strategies. It is possible that the perturbation in this study caused the feet to move back slightly as demonstrated in Fig. 3.15. If this movement occurs, then impulse in the A-P direction may not change much between conditions because the backward velocity of the feet may counteract the velocity gained from the perturbation.

However, the alternative explanation is that the magnitude of the impulse applied to the participants was insufficient to produce A-P impulse differences between the perturbation conditions. The anterior-directed impulse created in this study was standardized to each participant's body mass, had an average magnitude of 4.8 ± 1.8 N s, and lasted for 119.1 ± 25.0 ms. For obvious reasons, we deliberately selected a value for impulse that would be sufficient to change the performer's momentum, yet was safe. Thus, it was desired that we applied an impulse less than that experienced during an ACL injury episode. The range of magnitudes of perturbations applied to athletes during physical activity is not known. Thus, the magnitude of perturbation, applied to the participants during this study relative to actual perturbations also are not known.

There is no question that there is some critical combination of magnitude of force, time force is applied, and point of application of the perturbation to the performer's body that likely increases risk or causes an ACL injury. However, developing a better understanding of small perturbations has merit. In-flight perturbations are a common event occurring during sports participation, and suffering an ACL injury is the exception, rather than the rule, to a given athlete during every day practice and

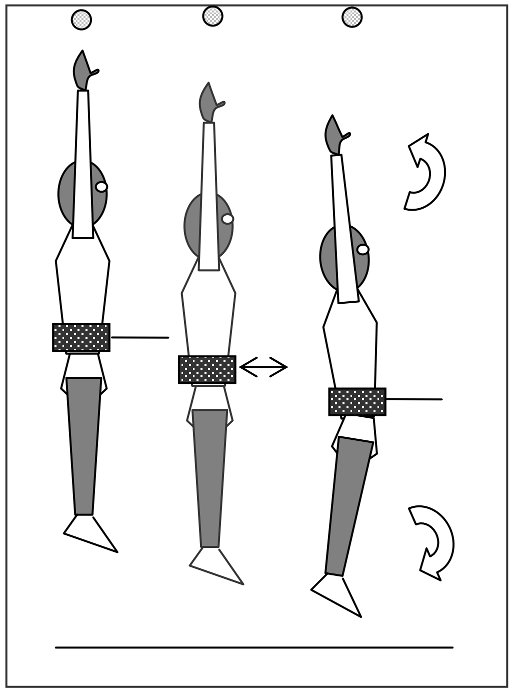

Figure 3.15. Effect of linear in-flight perturbation.

competition. Thus, small magnitude in-flight perturbations applied at the body's COM may not produce deleterious effects on landing biomechanics, as suggested by the results of this study. Further, it is important to understand how individuals successfully cope with perturbations to prevent injury.

Limitations of this study need to be mentioned. First, previously mentioned is the potentially low magnitude of the perturbation. Second, is that the perturbations were laboratory created, which could have provided the participants opportunity to use anticipatory strategies. However, to minimize this possibility, participants were blinded as to whether an in-flight perturbation would occur during a given trial, and the application of a perturbation among sham trials was randomized within blocks of trials to minimize anticipation. In addition, compared to participating during sport, participants knew the direction and approximate time during flight when an in-flight perturbation could occur.

In conclusion, an in-flight perturbation applied at the body's COM does lead to altered landing biomechanics. Increases in peak joint moments, specifically the peak knee extensor net muscle moment, could lead to increased loads placed on the ACL. Although the in-flight perturbations in this study appear to be relatively light in magnitude, it is evident that these events do affect the biomechanical strategy utilized during landing.

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CHAPTER 4

THE EFFECT OF IN-FLIGHT PERTURBATIONS ON LANDING BIOMECHANICS: II. ROTATIONAL PERTURBATION[1]

[1]S.W. Arnett, Y. Fu, R. Thompson, P. Sigurdsson, & K.J. Simpson. To be submitted to *Journal of Applied Biomechanics*.

Abstract

Perturbations have been implicated as a mechanism for ACL injury. One type of perturbation athletes encounter are in-flight perturbations. Although these events are common and have been associated with the occurrence of ACL injuries, the effects of these events on drop landing biomechanics are not known at this time. Therefore, the purpose of this study was to determine the effect of a rotational in-flight perturbation on landing biomechanics.

Twelve college-aged female soccer and basketball athletes performed drop landings with (PERT) and without (CON) in-flight perturbations being applied. Three-dimensional ground reaction forces and lower extremity joint kinematics and kinetics were analyzed using paired t-tests ($\alpha = 0.05$).

Compared to CON landings, peak vertical ground reaction force was decreased during PERT landings. There were no significant differences for lower extremity joint kinematic or kinetic magnitudes between landing conditions. The relative times to peak hip and knee flexion occurred later and relative time to the peak ankle plantarflexor moment occurred earlier during the PERT compared to the CON condition.

Decreased peak VGRF during PERT landings was diametrically opposite to the original prediction. This may have occurred due to a longer time to apply force to the ground, as evidenced by later relative times of knee and hip flexion during PERT landings. The decreased VGRF during PERT landings could also be interpreted as the participants altering their landing strategy due to anticipation of the perturbation. The lack of significant differences found between conditions led the researchers to qualitatively analyze the data at the individual participant level. There appeared to be

support for the prevalence of individual variation in landing strategies. These individual variations were present in both knee joint kinematics and kinetics. The effect of these variations on ACL injury incidence is difficult to discern at this time. However, the individual variation demonstrated by the participants of this study supports that strategies differ among and within participants and these variations in landing strategy could predispose a certain set of individuals to an increased risk of ACL injury. The low magnitude of the in-flight perturbation created in this study may represent typical perturbations encountered in the real world but are insufficient to cause injury. A limitation is that performing a perturbation task could result in a participant developing anticipatory strategies different than would emerge during a sport situation. Nevertheless, in-flight perturbations are common during sports participation and low magnitude rotational perturbations appear to influence the temporal pattern of landing biomechanics and the individual participant responses to perturbations.

Keywords: landing, ACL, perturbation, joint kinetics

Introduction

Sports involving jumping and landing (e.g., basketball rebounding, soccer heading, netball goal) often result in athletes colliding in the air with another athlete, i.e., an in-flight perturbation. Consequently, environmental perturbations have been reported to be part of the multifactorial etiology of ACL injuries (Hewett et al., 2006; Krosshaug et al., 2007). However, during everyday game and practice situations, athletes are not likely to encounter only one type of perturbation but a multitude of different environmental perturbations.

At this time, there is little research investigating the effects of environmental perturbations on lower extremity biomechanics (Shultz et al., 2000; Shultz et al., 2001). The effect of a linear in-flight perturbation on landing biomechanics was investigated in Part I (Arnett, Fu, Thompson, Sigurdsson, & Simpson, in submission, 2007). In the previous investigation, perturbation trials resulted in increased peak extensor hip and knee moments and ankle plantarflexor moment. The increased peak knee extensor moment could have led to an increased shear force on the proximal end of the tibia potentially increasing the mechanical loading of the ACL, compared to that experienced during CON landings. Temporal differences in joint kinematics were also demonstrated during the perturbation trials with peak ankle flexion and the peak plantarflexor moment occurring later and earlier, respectively, during the perturbation trials.

The point of application of an in-flight perturbation determines the resulting change in the performer’s momentum. As discussed in Part I, a linear in-flight perturbation would lead to an increase in linear momentum (Arnett, Fu, Thompson, Sigurdsson, & Simpson, in submission, 2007). However, an in-flight perturbation applied above the body's COM would lead to an increase in rotational and linear momentum ('rotational in-flight perturbation'). Therefore, for this study, a rotational in-flight perturbation (PERT) was defined as an external impulse applied 20% of trunk length superior to the body's vertical center of mass (COM_y) location during the flight phase of a drop landing.

Thus, the PERT applied to the participant should lead to an altered landing posture at contact compared to typical landings and landings during PERT trials in Part I (Arnett, Fu, Thompson, Sigurdsson, & Simpson, in submission, 2007) due to the rotational momentum gained during flight. After the PERT, it is predicted that the performer would have to produce increased joint extensor moments, compared to typical landings, to slow down the rotation of the respective proximal segment.

We hypothesized that the PERT compared to the CON condition would result in increased angular displacement, increased peak angles, and increased peak extensor moments for the hip, knee, and ankle. In addition, we hypothesized that posterior GRF impulse and peak VGRF and posterior GRF would be increased during the PERT compared to the CON condition. Peak knee valgus angles and moments were not expected to vary by more than .5 SD for the PERT compared to the CON condition. Therefore, the purpose of this study was to determine if an in-flight perturbation applied above the vertical location of the body's COM (PERT) resulted in altered landing biomechanics, compared to the biomechanics produced during a non-perturbed condition (CON). This information, in addition to the results of Part I (Arnett, Fu, Thompson, Sigurdsson, & Simpson, in submission, 2007), would aid in understanding the implications of different types of in-flight perturbations on landing performance.

Consequently, the PERT applied in this study was anticipated to require altered landing mechanics, compared to typical landings, that could place the ACL at higher risk for injury. In addition, understanding these alterations in mechanics may help us understand mechanisms of ACL injury following an in-flight perturbation.

Methods

The methods used in this study are similar to those used in Part I (Arnett, Fu, Thompson, Sigurdsson, & Simpson, in submission, 2007). Therefore, only brief descriptions of the methodology will be presented here.

Design

The design of this study was quasi-experimental. The two conditions of the independent variable were, perturbation (PERT), rotational PERT applied to the person and control (CON), the initial block of drop landing trials performed prior to any perturbation trials.

Participants

Participants consisted of 12 college-aged, recreationally active females. Participants were recruited from the local community and university. The participant inclusion/exclusion criteria were the same as those used in Part I (Arnett, Fu, Thompson, Sigurdsson, & Simpson, in submission, 2007): healthy, noninjured, without neurological problems and had been/currently were participating in competitive basketball or soccer and had not performed lower extremity resistance training within the past 48 hours. Participants were randomly assigned to this or the linear perturbation, Part I study (Arnett, Fu, Thompson, Sigurdsson, & Simpson, in submission, 2007). Using an adjusted effect size for a repeated measures statistical test (Lipsey, 1990), based on several outcome variables from an initial subset of the participants in this study, a sample of 12 participants was predicted to elicit adequate power (0.80) for the kinetic variables. Two biomechanical variables (kinematic) would require a sample size in excess of 90 to obtain adequate statistical power.

Drop Landing Procedures

Test Task:

One drop landing task was performed for each trial. The drop landing task was performed with the arms overhead and the shoulder joint close to full flexion. As in Part I, the proprietary device ("BIFIC" in Figure 4.1) was used to create in-flight perturbations. The components of this device have been described previously in Part I (Arnett, Fu, Thompson, Sigurdsson, & Simpson, in submission, 2007). An average force of 1.15 times body mass was applied to the performer 126.8 ± 26.9 ms during the flight phase when the cable attached to the performer was close to horizontal (-7 to 10°, absolute angle from horizontal). A load cell (MLP-200, Transducer Techniques, Temecula, CA) was used to collect the force signals produced by BIFIC. Raw force signals were collected (1200 Hz) conditioned, and amplified (gain = 1).

The experimental set-up was similar to that used in Part I (Arnett, Fu, Thompson, Sigurdsson, & Simpson, in submission, 2007) and a brief description of the equipment used follows. A seven-CMOS 4.1 megapixel, Vicon MX™ camera system and Workstation™ software (v. 5.2.4) (Vicon, Inc., Englewood, CA) were used to capture the spatial locations of reflective markers (sampling frequency = 240 Hz, shutter speed = 1/1000 s). For video acquisition, 47 reflective markers were placed on the participant's skin and clothing using the designated marker set (Table B1, Appendix B) (Lu &

O'Connor, 1999), and two markers were placed on the apparatus cable 0.61 m from the participant.

An AMTI force platform (Model OR6-6-1, Advanced Mechanical Technology, Inc., Newton, MA) was used to collect ground reaction force (GRF) signals. Raw GRF signals from the antero-posterior (AP-GRF) and vertical (VGRF) directions were collected (1200 Hz) and amplified (gain = 4000) prior to being saved on a desktop computer.

As done in Part I (Arnett, Fu, Thompson, Sigurdsson, & Simpson, in submission, 2007), a classic reaction board technique (Fig. 4.2) (Hamill & Knutzen, 1995) was used to determine the vertical location of the participant's COM for a static position similar to the body position that was exhibited during the flight phase. The belt worn by the participant was then aligned at 20% of the participant's trunk length (mid PSIS to C7) superior to the vertical COM location.

Protocol

Pre-drop test procedures:

The pre-drop test procedures were the same as those reported for Part I (Arnett, Fu, Thompson, Sigurdsson, & Simpson, in submission, 2007). The potential participant signed the consent form in compliance with regulations of the human subjects institutional review board, and filled out a health status/sports participation questionnaire. Anthropometric measures were obtained (Zatsiorsky, 2002), estimated flight phase COM_y location determined, and reflective markers were placed on the participant. The participants performed a 5-minute warm-up consisting of stationary cycling and light stationary hopping, then two to five practice trials each of no perturbation and then PERT landings were performed while attached to the apparatus. The impulse applied to performer during practice in-flight perturbation trials was 75-82% of the impulse applied during actual testing.

Drop Test Procedures:

The participant was instructed that an in-flight perturbation may or may not occur during each test trial. After performing each trial, to minimize fatigue of the leg musculature, the participant rested for 15-20 s. The first block of trials consisted of five trials of CON landings, then a second block of five trials each of perturbation (PERT) and five non-perturbation sham trials quasi-randomly ordered. To prevent participants from using anticipatory strategies, no more than two trials of the same landing condition were performed consecutively (Pavol & Pai, 2002). A trial was deemed acceptable if the participant: performed a forefoot-to-heel landing, landed with the right foot touching only the force plate and the left foot touching only the ground, did not take a step forward or backward, and maintained foot position for at least 2 s after contacting the ground. If a non-acceptable trial occurred, the trial was repeated after the remaining trials of the current trial block were completed.

Data Reduction

As in Part I (Arnett, Fu, Thompson, Sigurdsson, & Simpson, in submission, 2007), the impact phase was of interest: from foot contact to the time after the highest peak vertical ground reaction force (VGRF) occurred when the local minimum was displayed [~100-150ms after initial contact]. (See Appendix E for more detail on the definition of the impact phase.)

For the kinematic data, the spatial locations of the reflective markers were reconstructed using proprietary methodology within the Vicon Workstation® software (v. 4.3.1) and smoothed using Woltring's generalized cross-validated spline (GCVSPL) algorithm (Woltring, 1986). A 5-link (foot, shank, thigh, pelvis, and trunk [thorax + abdomen]) rigid-body model was used for kinematic and kinetic analyses. Kinematic and biomechanical quantities (linear displacement, velocity, and acceleration; joint angles, angular velocity, and angular acceleration) were calculated using MatLab® software (v. 7.0). Following the International Society of Biomechanics guidelines (Wu & Cavanagh, 1995), a Cardanic rotation sequence was used to calculate lower extremity segment rotations (Z-Y-X). Joint angles were then calculated from the rotation matrix representing the relative relationship between segments. Linear velocity and acceleration values were generated by taking the first and second differentiation of the linear displacement data. Angular velocity and acceleration values were generated using techniques reported by Winter (2005). Kinematic variables for the lower extremity joints were: peak angles, the relative times that these events occurred, and flex/ext displacement.

For antero-posterior ground reaction forces (A-P GRF) and VGRF, magnitudes were scaled to body mass, and time was scaled relative to impact phase time. Peak VGRF and A-P GRF were analyzed to test the hypotheses that the perturbation leads to increased maximum force applied in the vertical and antero-posterior directions. Posterior A-P impulse was compared between conditions to confirm that after a perturbation, during landing, greater posterior impulse must be applied to the performer.

Joint kinetics of the right ankle, knee, and hip joints were calculated using 3-D inverse dynamics (Winter, 2005). COM locations and moments of inertia for each of the lower extremity segments were determined using Zatsiorsky's anthropometric model for females (Zatsiorsky, 2002). To test the prediction that during PERT landings peak extensor and plantarflexor moment magnitudes and temporal patterns would be altered, peak magnitudes (scaled to body mass and leg length) and the relative time to these events were calculated. Relative time (Rel-T) to selected events was calculated as a percentage of the total time of the impact phase (Rel-T = % Tot-T).

Paired t-tests were used to compare the values of the CON and PERT conditions (alpha = .05). To determine behavioral significance, adjusted effect sizes for paired observations (ES_{ap}) were calculated (Lipsey, 1990). Due to limited group findings between PERT and CON landings, a-posteriori qualitative analyses were conducted on joint kinematic and kinetic data.

Results

Participant characteristics are reported in Table 4.1. GRF variables are presented in Figures 4.3-4.6. See Figure E2 in Appendix E for representative GRF-time curves (Participant #7). The peak VGRF (Figure 4.3) was 1.2 $N{\cdot}kg^{-1}$ lower during the PERT compared to the CON condition. No significant differences existed between the remaining GRF magnitude or time variables.

Kinematic and kinetic values are presented in Figures 4.7-4.12 and 4.13-4.14, respectively. See Figure F3 and F4 in Appendix F for representative joint angle-time curves (Participant #7). There were no significant group differences between conditions for lower extremity kinematic or kinetic variables. For knee flex/ext angular

Table 4.1. Participant characteristics

	Mean ± SD	Range
Age	21.2 ± 1.5	18-24
Body Mass (kg)	64.6 ±7.9	52.9-76.2
Body Height (cm)	171.9 ± 5.5	160.8-180.1
No. yr participation		
Soccer	11.6 ± 5.0	2-17
Basketball	9.2 ± 7.1	1-17

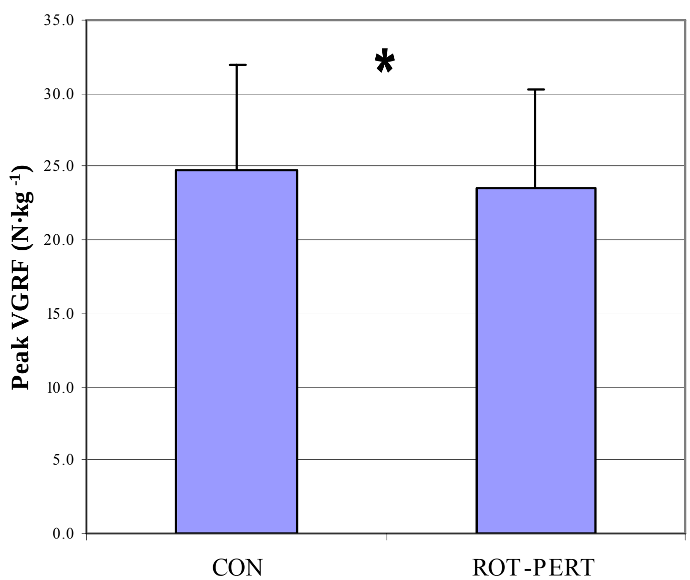

Figure 4.3. Peak VGRF for control (CON) and perturbation (PERT) conditions. * indicates significant difference at $p < 0.05$.

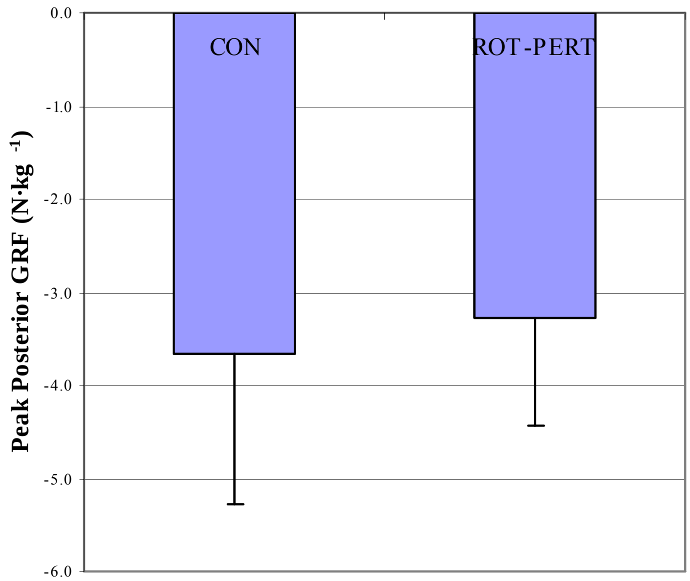

Figure 4.4 Peak posterior GRF.

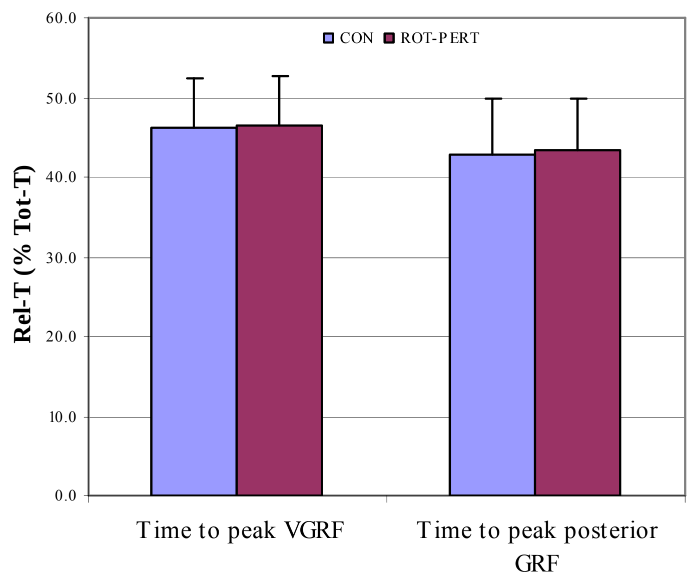

Figure 4.5. Relative time (Rel-T) to GRF variables for CON and PERT conditions.

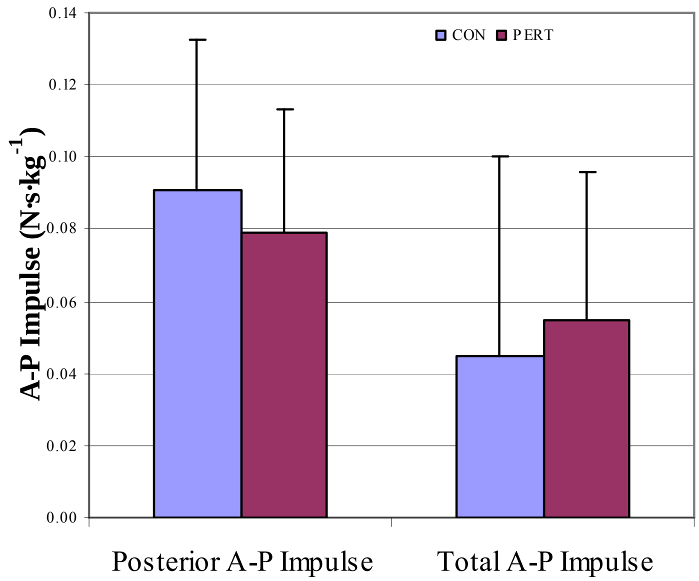

Figure 4.6. A-P impulse variables.

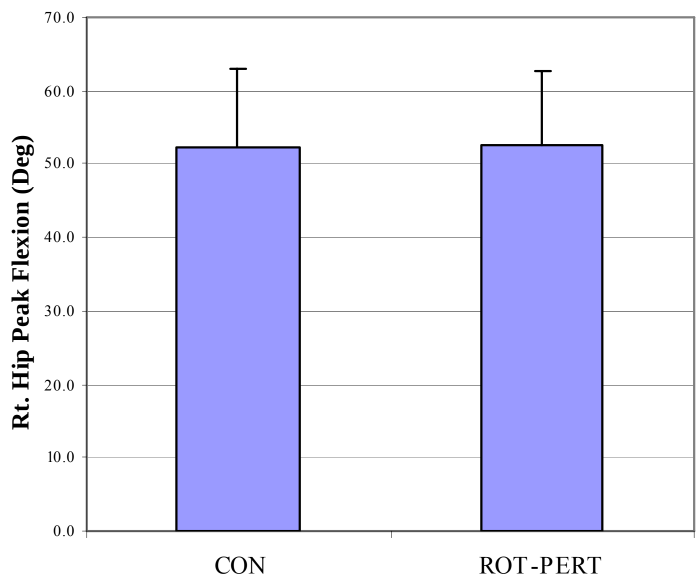

Figure 4.7. Peak hip flexion.

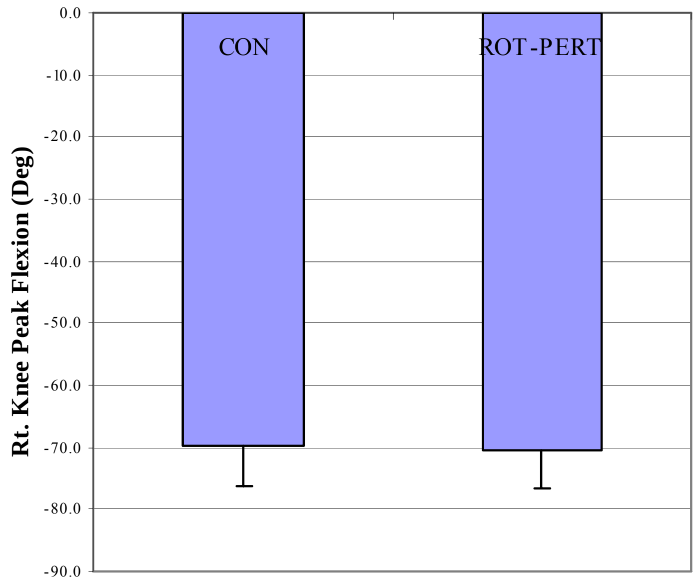

Figure 4.8. Peak knee flexion.

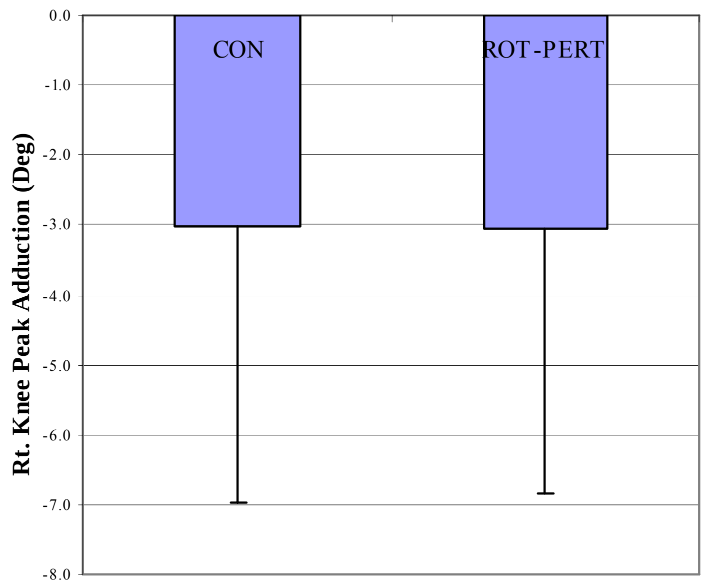

Figure 4.9. Peak knee adduction.

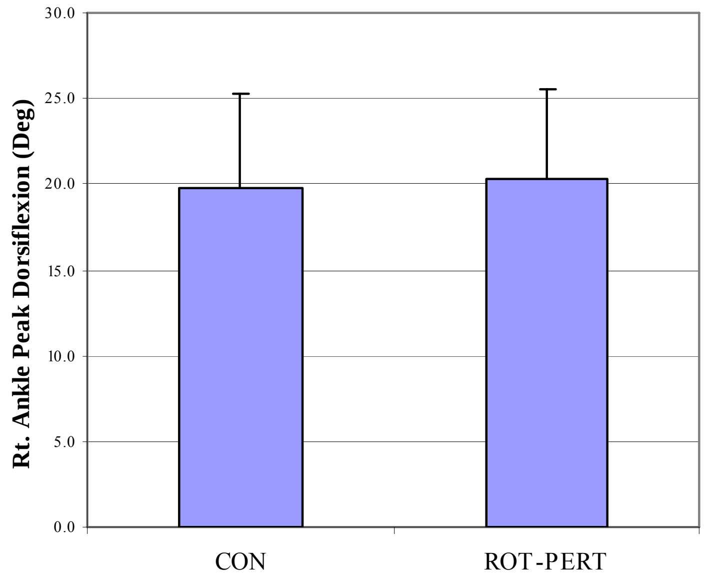

Figure 4.10. Peak ankle dorsiflexion.

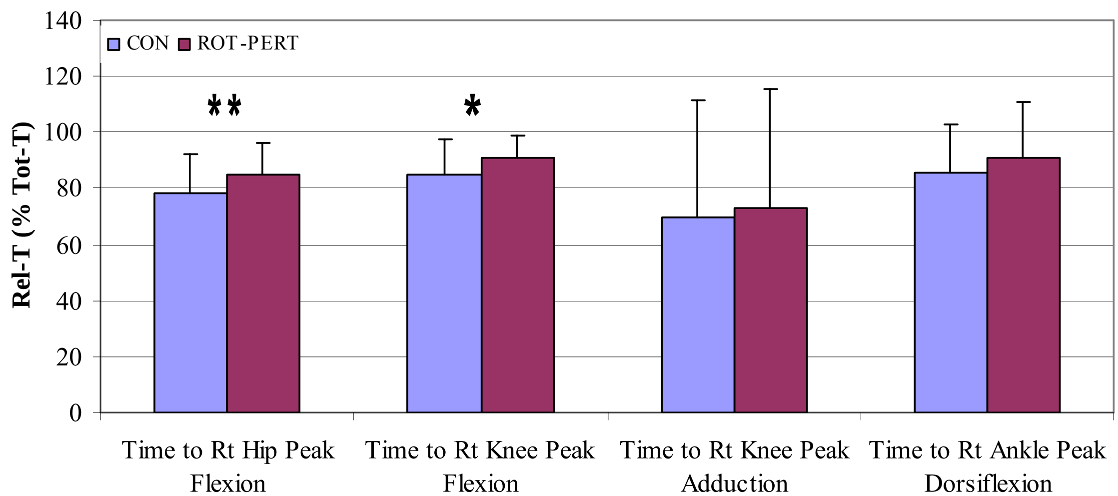

Figure 4.11. Relative time to peak joint angles for the CON and ROT-PERT conditions. * indicates significant difference at $p<0.05$, ** $p<0.01$.

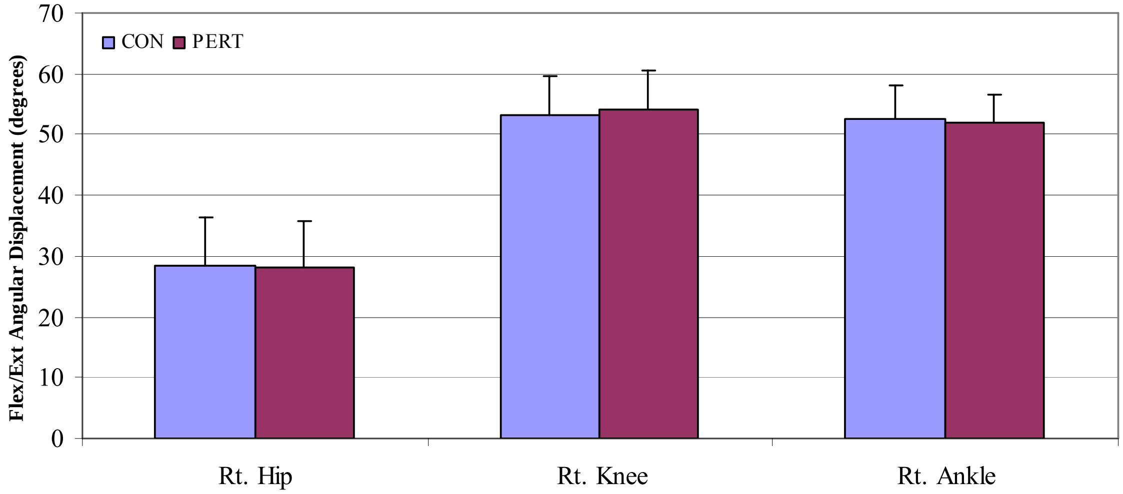

Figure 4.12. Flexion/extension displacement.

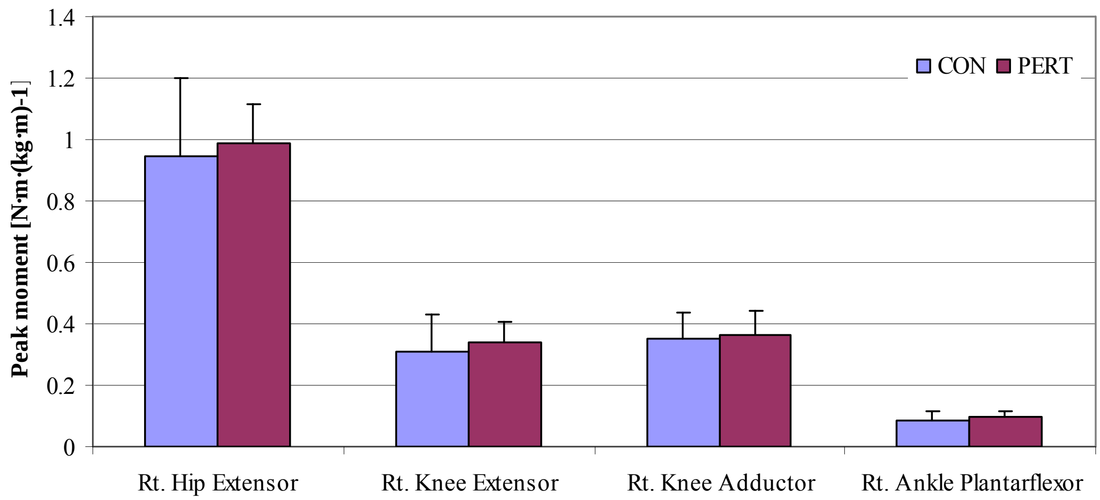

Figure 4.13. Peak joint moments.

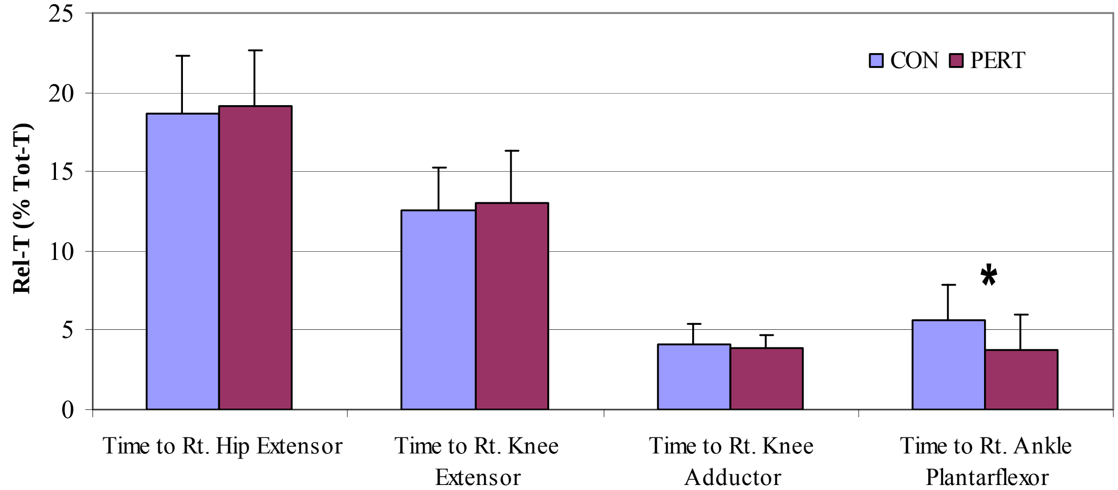

Figure 4.14. Time to peak moments. * indicates significant difference between conditions at $p < 0.01$.

displacement, 42% of the participants had 2° or more displacement during PERT compared to CON landings and 8% had 2° or more displacement during the CON compared to PERT landings. Individual participant variations were visually noted in the joint moment curves.

Discussion

The goal of this study was not to elicit injuries, so creating an impulse similar to those leading to an ACL injury was not performed. Furthermore, the differences between the perturbation created in this study and perturbations in real life are not known.

It is possible that the magnitude of the impulse applied to the participants was insufficient to produce differences in A-P GRF, peak joint angle, and peak joint moment magnitudes between conditions. The anterior-directed impulse created in this study was standardized to each participant's body mass, had an average magnitude of 6.5 ± 4.0 N·s, and lasted for 126.8 ± 26.9 ms. In light of the findings from Part 1, the impulse value chosen for this study was low in magnitude as safety remained a concern due to the perturbation having a different point of application.

It was hypothesized that landing after an in-flight perturbation would require increased posterior A-P impulse to reduce the body's linear and angular momenta. Thus, increased A-P forces were anticipated to create the additional impulse needed. However, differences for A-P GRF variables did not demonstrate differences between PERT and CON landings, likely a result of individual participant variation.

Similar to Part I , we were surprised to find that peak VGRF was significantly lower rather than higher during the PERT compared to the CON condition (p=0.04). For both studies, by anticipating the in-flight perturbation, during PERT landings, the participants may have increased lower extremity flexion to increase the linear displacement of the COM during landing. The perturbations were laboratory created, which could have provided the participants opportunity to use anticipatory strategies. However, to minimize this possibility, participants were blinded as to whether an in-flight perturbation would occur during a given trial, and the application of a perturbation among sham trials was randomized within blocks of trials to minimize anticipation. Furthermore, compared to participating during sport, participants knew the direction and approximate time during flight when an in-flight perturbation could occur. However, joint kinematic data do not support that this occurred during the impact phase. Second, and likely the more important and interesting explanation, is the differences in individual participant responses.

Individual participant differences also were the most likely explanation for lack of group differences between PERT and COND landings for the joint kinematics and moments, too. It was surmised that, although women are more likely than men to undergo an ACL injury, that it was more likely that it was a subset of women who have a particular combination of morphological and biomechanical factors that interact to place them at higher risk for an ACL injury. As it was known that landing strategies vary among individuals (James et al., 2003) and ACL injury etiology is likely multifactorial (Hewett et al., 2006), it is of the utmost importance to understand individual participant responses, particularly within research focusing on injury. In addition, individual landing strategies were noted in Part I (Arnett, S., Fu, Y, Thompson, R., Sigurdsson, P., & Simpson, K.J., in submission to Journal of Applied Biomechanics, 2007). Therefore, the following is a discussion of qualitative observations of participant responses to perturbations.

The group findings for angular kinematics displayed almost no differences between the PERT and CON landings, likely due to individual variation in landing strategy (James et al., 2003). The critical difference used in this study, 2°, was similar to that used in Part I (Arnett, S., Fu, Y, Thompson, R., Sigurdsson, P., & Simpson, K.J., in submission to Journal of Applied Biomechanics, 2007) due to the maximum standard deviation for any angle or angular displacement being < 2˚. About the mediolateral axis of the knee, 42% of the participants displayed 2˚ or more displacement during the PERT compared to CON condition, while 8% exhibited the opposite (2˚ or more during the CON condition), with the remaining participants exhibiting similar values during both conditions. However, between the PERT and CON conditions no participants demonstrated INT/EXT or AB/ADD knee displacement > 2 °. Although increased flex/ext angular displacement of the knee is not noted as an ACL injury mechanism, it supports previous findings of inter-individual variation in landing strategies (James et al., 2003).

The patterns of the knee joint moment curves prompted further qualitative inspection to determine if individual differences between conditions occurred. Indeed, individual variation in the pattern of the joint moment curves did occur between conditions (Figures 4.15-4.20). The variation occurring at the knee joint further supports that individuals utilized different strategies during landing (James et al., 2003). In addition it supports that within individuals, the PERT condition resulted in an individual modifying the joint kinetics used during landing. These changes were not in terms of

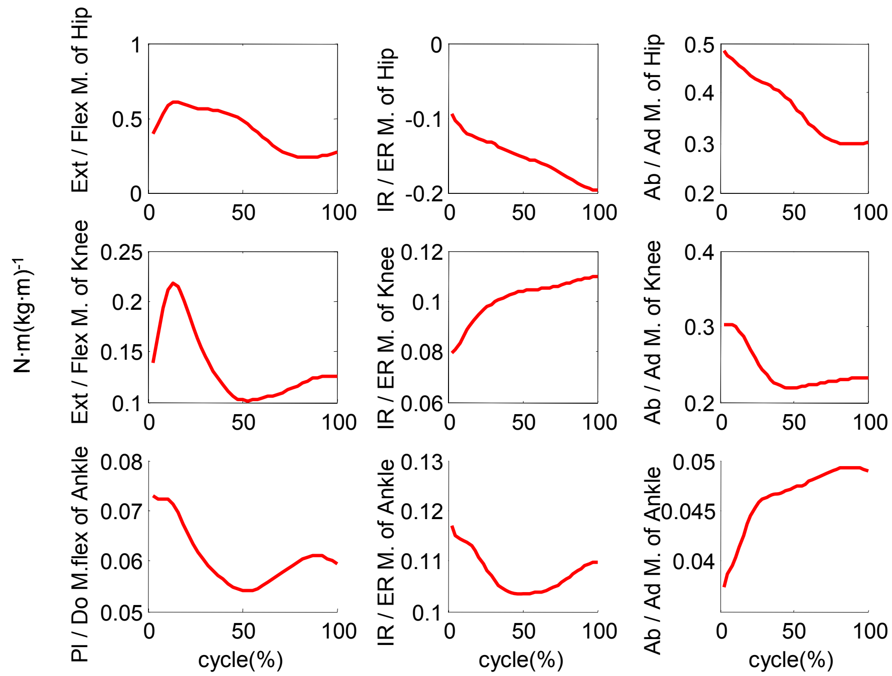

Figure 4.15. Representative join moment curves for CON condition (Participant 7).

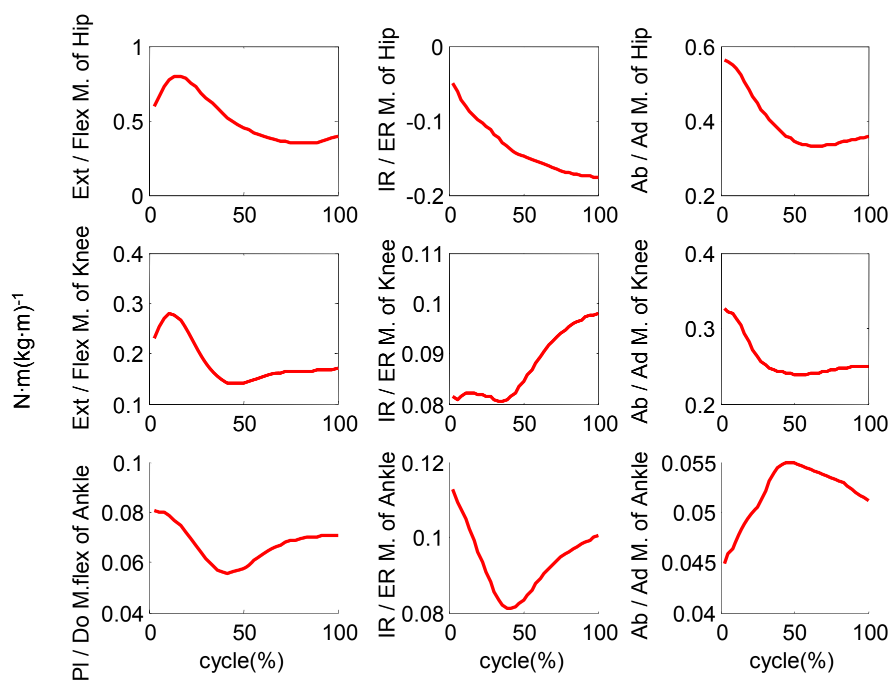

Figure 4.16. Representative joint moment curves for PERT condition (Participant 7).

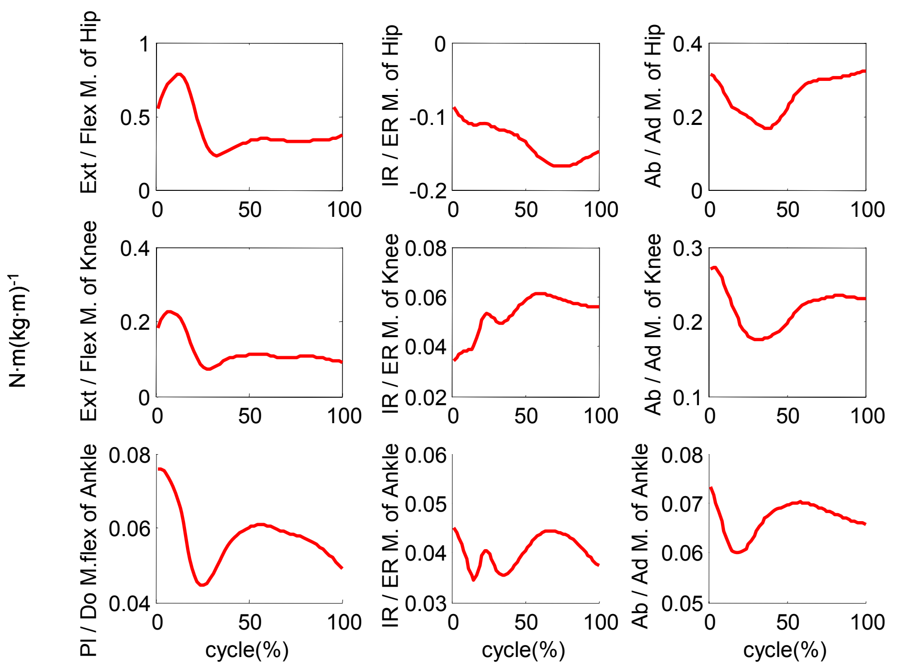

Figure 4.17. Representative joint moment curves for CON condition (Participant 10)

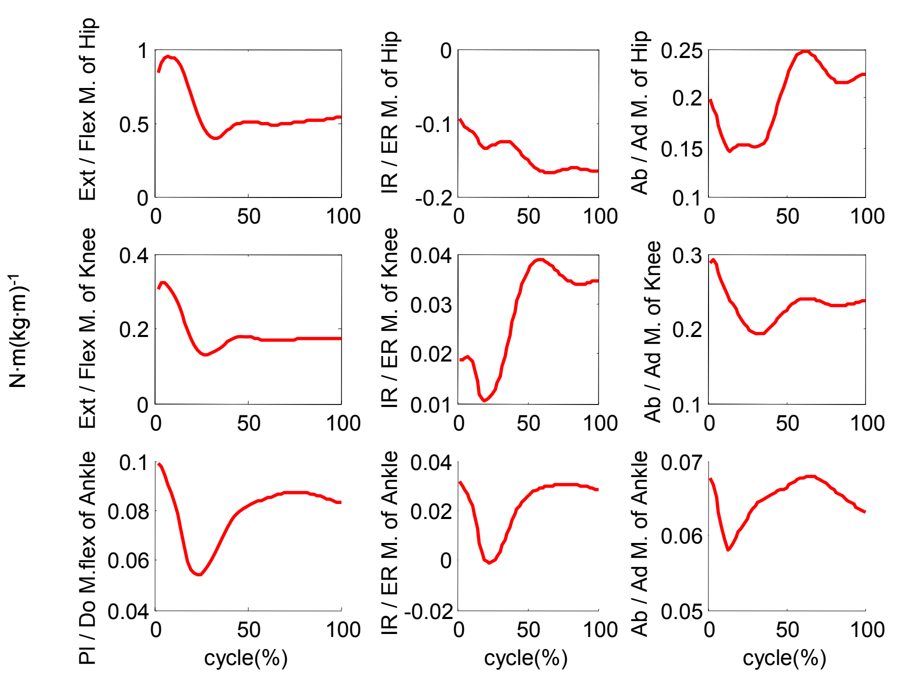

Figure 4.18. Representative joint moment curves for PERT condition (Participant 10).

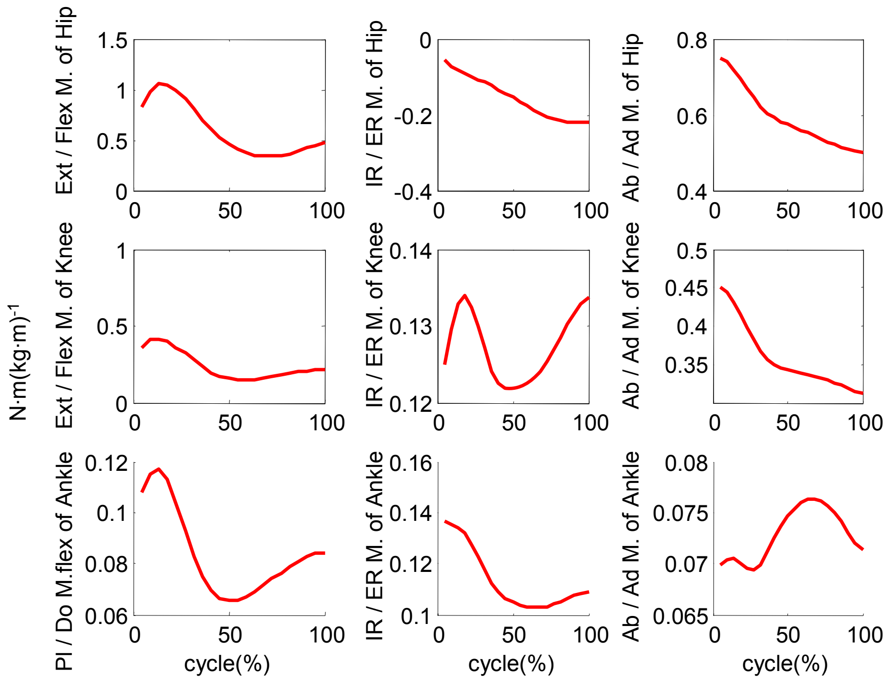

Figure 4.19. Representative joint moment curves for CON condition (Participant 17).

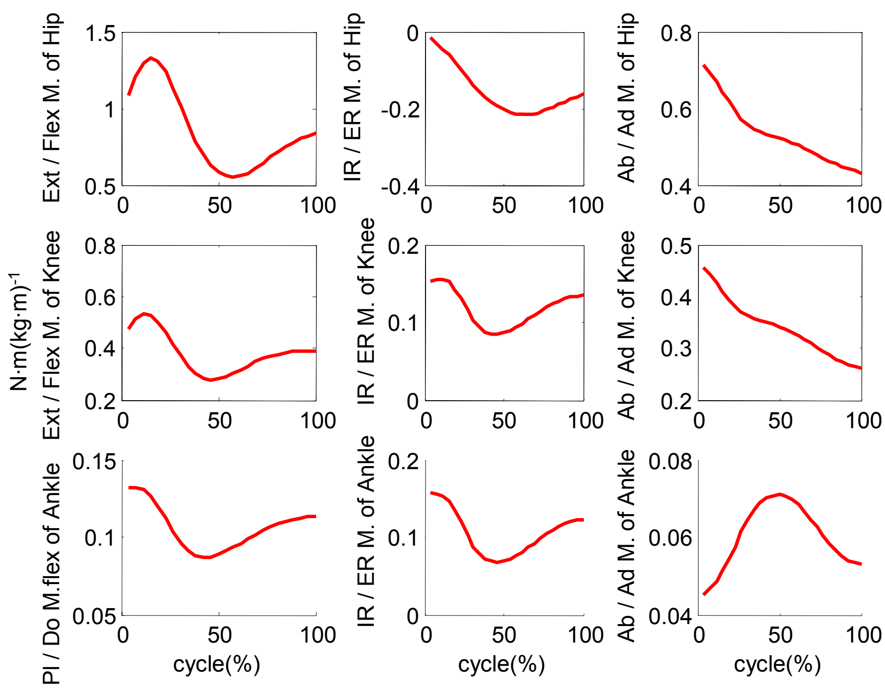

Figure 4.20. Representative joint moment curves for PERT condition (Participant 17).

magnitude for all knee moments, but specifically for the pattern of the curve during the impact phase. This finding would help explain the lack of differences seen in the group analysis between conditions. However, the importance of including some type of individual analysis technique, especially regarding ACL injury research, is supported by this finding.

It is possible that upper extremity biomechanics play an important role in responding to a rotational perturbation, as the trunk, head and upper extremities comprise a large portion of the mass of the body and movements of the trunk have a large influence on the location of the body's COM. When the PERT was applied to the participant's trunk, the trunk was anticipated to rotate in a flexion direction while also translating (hence, causing the entire body to translate) in the anterior direction. Both responses thus, were surmised to place the performer's body into a lower stability situation at ground contact. A reaction by the performer in response to the perturbation might be decreased trunk flexion and increased arm extension after ground contact. These movement strategies counteract the anterior rotation of the trunk, thereby helping maintain stability during the impact phase of landing.

The results of this study combined with the results of Part I (Arnett, S., Fu, Y, Thompson, R., Sigurdsson, P., & Simpson, K.J., in submission to Journal of Applied Biomechanics, 2007) support that in-flight perturbations affect landing biomechanics. In addition, it is likely that certain combinations of the components of an in-flight perturbation (magnitude of force, time force is applied, and point of application to the performer's body) produce individual participant results that are unique, and potentially more likely to increase the likelihood of an ACL injury than other combinations of force applications.

Developing a better understanding of small in-flight perturbations acting at different points of application on the body remains important. Athletes commonly have to cope with these events during daily practice and competitive situations, yet ACL injuries do not always occur following an in-flight perturbation. Thus, small magnitude in-flight perturbations applied above the body's COM may not produce deleterious effects on landing biomechanics, as suggested by the results of this study. In order to help athletes prevent ACL injuries, it is important to understand individual participant variations in joint motions and moments used to cope with these events.

In conclusion, an anterior-directed in-flight perturbation applied superior to the body's COM appears to lead to altered temporal patterns in landing biomechanics. Based on qualitative analyses of joint kinematics and kinetics, individual participant variations in landing biomechanics were demonstrated during PERT landings. Increases in flex/ext angular displacement at the knee may help reduce ACL loading by preventing the knee from being close to full extension, an ACL injury mechanism (Boden et al., 2000). Differences in the pattern of int/ext rotation moments at the knee may lead to increased ACL loading due to its role in preventing internal rotation of the tibia, relative to the femur (Inoue et al., 1987).

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CHAPTER 5

SUMMARY, CONCLUSIONS, AND RECOMMENDATIONS

Summary

Due to the continued prevalence of ACL injuries in athletics, it remains imperative to help define factors and mechanisms predisposing individuals to these types of injuries. In-flight perturbations have been implicated in ACL injuries during practice and game settings, yet the effect of these events on landing biomechanics are not known. Additionally, it has been predicted that in-flight perturbations lead to altered biomechanical landing strategies resulting from the effect of these events on a human during flight. Therefore, the purpose of these two studies was to investigate the effect of in-flight perturbations on landing biomechanics.

Twenty five participants volunteered for this project which consisted of two studies; a linear in-flight perturbation study (n=13) and a rotational in-flight perturbation study (n = 12). Within each study the participants were blinded to the in-flight perturbation (PERT) randomly mixed with a non-perturbation event (CON). Ground reaction forces and 3-D lower extremity joint kinematics and kinetics were collected while participants performed drop landings with and without in-flight perturbations. Paired t-tests were used to examine differences between CON and PERT conditions.

Reaction to a linear in-flight perturbation led to altered biomechanical strategies during landing. The results of the first study supported in part the major premise of the study, which was that peak joint moments would increase in order to help decrease the momentum gained during flight. It was surmised that the perturbation pulled the individual's center of mass (COM) forward while the upper and lower extremities moved independently of the COM. This response may be explained by the fact that the human body is a multi-link chain and not a rigid object.

The results of the second study did not support the major premise of the study. Reaction to a rotational in-flight perturbation failed to elicit altered magnitudes of the biomechanical variables investigated. However, temporal differences occurred between conditions. It was possible that the rotational perturbation influenced trunk motion in such a way that the response to each perturbation was highly variable among participants. Therefore, differences for joint motion and moment patterns of the lower extremity were not able to be discerned among all participants, making it difficult to define common, biomechanical responses.

In addition, it was possible that the altered strategy during PERT landings were reflected by trunk flexion and/or shoulder rotations. However the upper extremity biomechanics used during PERT landings in this project were not investigated. The lack of an altered lower extremity biomechanical strategy points to the importance of understanding upper extremity biomechanics during PERT landings due to these segments constituting a large portion of the mass of the body. However, the size and distribution of the mass of these segments, in addition to the lack of rigidity in the trunk, should provide greater rotational inertia. Hence, the upper body may produce greater resistance to in-flight perturbations, resulting in less alterations in landing biomechanics. Therefore, it is possible that the soft tissues of the body, tendons, ligaments, menisci, etc, may have absorbed some of the effects of the PERT.

Of interest in both parts of the project were the individual variations in joint kinematics and kinetics at the knee joint, as those elements are related to the strain and stress placed on the ACL during landings. It was likely that individuals who exhibited variations in joint kinematics and kinetics at the knee joint, compared to those demonstrating no change, between PERT and CON landings may be more susceptible to an ACL injury, or vice versa. The individual variation in knee flex/ext angular displacement and joint moment patterns between PERT and CON landings supports that individual variations in landing biomechanics occur after a perturbation. Identifying trends within these individual variations are examples of focus areas still in need of further investigation.

The results for the peak VGRF and posterior GRF did not support the original prediction in either part of the project. The possible explanations were increased time to peak joint flexion and anticipation of the perturbation from the participants. As the GRF for the sham trials demonstrated differences from the CON conditions, it is likely that the participants were anticipating the in-flight perturbation at some level.

The in-flight perturbations created in this study were designed to move a rigid object, weighing approximately 68 kg, 2-3 cm anteriorly. However, as alluded to previously, the flight behavior of a rigid object is not the same as a multi-link chain, such as the human body. This assumption, in part, might help explain the lack of differences occurring for the posterior AP impulse. If the human body behaved like a rigid object, the posterior AP impulse would have to have increased during the PERT compared to CON landings. One explanation, therefore, of the lack of group differences between the landing conditions is that the relative velocity of the feet may have been less during PERT landings, thereby resulting in no effect on the AP GRF variables.

Conclusions

In conclusion, in-flight perturbations appeared to influence landing biomechanics and for rotational perturbations, joint moment responses appeared to be more dependent on individual participants. In-flight perturbations are common in many sports and do not always lead to injury. Understanding the mechanics of low-magnitude perturbations will help us understand eventually the long-term effects of perturbation loading on the lower extremity joints during sport participation and set the foundation for understanding abnormal joint loading and strains during ACL injury. Conclusions from these studies lead us to believe that perturbations with low magnitudes acting in an anterior direction may not atypically affect lower extremity landing biomechanics of all of our female participants.

Recommendations

Future studies should examine the effects of in-flight perturbations acting in different directions, with greater magnitudes, and on upper extremity biomechanics. Many unanswered questions continue to exist in this area of biomechanical research. This project has assisted in facilitating increased knowledge of how to better recreate in-flight perturbations in a laboratory setting in order to narrow the focus on which explanatory factors need further examination.

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Appendices

Appendix A

Impulse Calculation

To represent the total impulse generated, it is more realistic to calculate the amount of impulse generated during very small increments of time, then sum the individual impulses. An impulse, I(t), generated at time t for a very short time increment (dt) is:

$I(t) = F(t) \times dt$[1]

where F(t) is the perturbation force acting on an individual during flight in the antero-posterior direction at time t. The total impulse (I_{tot}) generated is the sum of the impulses applied between two intervals of time (t_1 to t_2):

$$I_{tot} = \sum_{t_1}^{t_2} I(t)$$

and the total impulse generated produces the equivalent amount of change in momentum.

[1] $I = \int_{t1}^{t2} \Sigma F \, dt$

Appendix B

Table B1. Marker locations for each segment.

Body segment	*Label*	*Descriptions*
Head	LFHD	Left head marker above the ear
	RFHD	Right head marker above the ear
Trunk	SN	Sternum
	XP	Xiphoid process
	RTBK	Right scapular inferior angle
	C7	The seventh cervical spinous process
	T10	The tenth thoracic spinous process
Left upper limb	LACJ	Left acromioclavicular joint
	LHUM	Left humerus marker
	LELB	Left lateral humerus epicondyle
	LFRM	Left forearm marker
	LWR1	Left styloid process marker on the radius
	LWR2	Left styloid process marker on the ulna
Right upper limb	RACJ	Right acromioclavicular joint
	RHUM	Right humerus marker
	RELB	Right lateral humerus epicondyle
	RFRM	Right forearm marker
	RWR1	Right styloid process marker on the radius
	RWR2	Right styloid process marker on the ulna

Pelvis	LASI	Left anterior superior iliac spine (ASIS)
	RASI	Right anterior superior iliac spine marker
	LPSI	Left posterior superior iliac spine (PSIS)
	RPSI	Right posterior superior iliac spine (PSIS)
Left femur	LTRO	Left greater trochanter
	LTHI	Left thigh marker
	LLFC	Left lateral femoral epicondyle center
	LMFC	Left medial femoral epicondyle center
Right femur	RTRO	Right greater trochanter
	RTHI	Right thigh
	RLFC	Right lateral femoral epicondyle center
	RMFC	Right medial femoral epicondyle center
Left shank	LTT	Left tibial tuberosity
	LFHM	Left fibular head
	LLMA	Left lateral malleolus
	LMMA	Left medial malleolus
Right shank	RTT	Right tibial tuberosity
	RFHM	Right fibular head
	RLMA	Right lateral malleolus
	RMMA	Right medial malleolus
Left Foot	LHEE	Left heel
	LFMT	Left fifth metatarsal head

	LNTC	Left Navicular tubercle
	LTOE	Left middle foot of distal metatarsal
Right Foot	RHEE	Right heel
	RFMT	Right fifth metatarsal head
	RNTC	Right Navicular tubercle
	RTOE	Right middle foot of distal metatarsal

Appendix C

COM Calculation

F_{P+B} = force produced by the participant and the reaction board

L_{RB} = length of the reaction board

W_P = participant's body weight

COM_x = distance of participant's COM from non-scale end of reaction board

W_{RB} = Weight of the reaction board

L_{COMRB} = distance of COM of reaction board from non-scale end

$\Sigma T = 0$

$$-F_{B*}L_{RB}+W_B*L_{COMRB} = 0 \quad (1)$$

$$-F_{P+B}*L_{RB}+W_P*COM_x+W_{RB}*L_{COMRB} = 0 \quad (2)$$

$$COM_x = ((F_{P+B}*L_{RB})-(W_{RB}*L_{COMRB}))/W_P \quad (3)$$

The force produced by the reaction board alone is measured (F_B) and used in equation 1 to solve for W_B*L_{COMRB}. Equation 3 is derived from equation 2. The value obtained from equation 1 is used in equation 3. The person lies on the reaction board and the second force measurement is collected (F_{P+B}). This value is inserted into equation 3 and solved for COM_x.

Appendix D

Health Status/Sports Participation Questionnaire

Participant #: _____
Date: _____________
Invest. initials: ______

Health Status/Sports Participation Questionnaire

Please respond as completely as possible. Your responses to this questionnaire will be kept confidential and will only be reviewed by the research investigators.

Age _______

Health Condition

1. Please identify how you would evaluate your health overall (circle best choice below)

   Excellent   Good   Fair   Somewhat poor   Very poor

Do you have any current medical problems (e.g., hypertension, back strain, lower extremity injury, ankle sprain, torn meniscus, lower extremity malalignment, etc.)? Circle: yes no If yes, complete the following. Use one row for each medical problem.

Describe the problem	Has it been diagnosed by a physician? (yes/no)	Treated by a physician or other medical professional? (yes/no)	Does the problem affect your balance, strength, vision, movements; or produce nausea or dizziness? (yes/no)	Are you currently taking medication for problem? (yes/no). If yes, list the medication.
1.				
2.				
3.				

Are you currently taking medication(s) and have you changed the dosages of any medication(s) (prescription or nonprescription) you currently take that have any of the following side effects?
(Circle Yes or No)

a. Balance problems? Yes / No
b. Feel sick or nauseated during physical activity? Yes / No
c. Dizziness? Yes / No
d. Vision problems? Yes / No
e. Feel coordination is off? Yes / No
f. Ability to think or follow directions is impaired? Yes / No
g. Other side effects: Yes / No If yes, describe side effect: _________________________

____

Have you ever experienced the following? Please circle "yes" or "no." If yes, then check off the appropriate spaces.

Yes No Broken bone? If so: ___ right leg ___ left leg ___ spine ___ foot (rt. lt.)

Yes No Surgery? If so: ___ right leg ___ left leg ___ spine ___ foot (rt. lt.)

Yes No Sprain to the following: ____ hip (rt. lt.) ___ knee (rt. lt.) __ ankle (rt. lt.)

Yes No Are you currently experiencing any lower extremity, or back, pain/discomfort/injury? If yes, describe:

____

Yes No Are you currently recovering from an illness or an injury? If yes, or not sure, describe:

____

Yes No Have you been diagnosed with a lower extremity malalignment? If yes, describe:

____

Yes No Have you had a previous lower extremity, back, or head injury within the last year, regardless if medical attention was sought? If yes, describe:

____

Yes No Have you been diagnosed with a cardiovascular or pulmonary condition? If yes, describe:

Yes No Have you been diagnosed with a visual, vestibular, neurological or other condition affecting your balance? If yes, describe:

Yes No Have you performed a lower extremity resistance workout within the last 48 hours? If yes, describe:

Yes No Is there any other information related to your health that we should know? If yes, describe:

Other comments:
Please check the sport(s) that you have participated in and the level of participation.

	High School Varsity	College Intramural	College Varsity	Recreational (Organized)	Recreational (Unorganized)
Basketball					
Soccer					

For the sport(s) checked above, when did you last participate and how long have you participated?

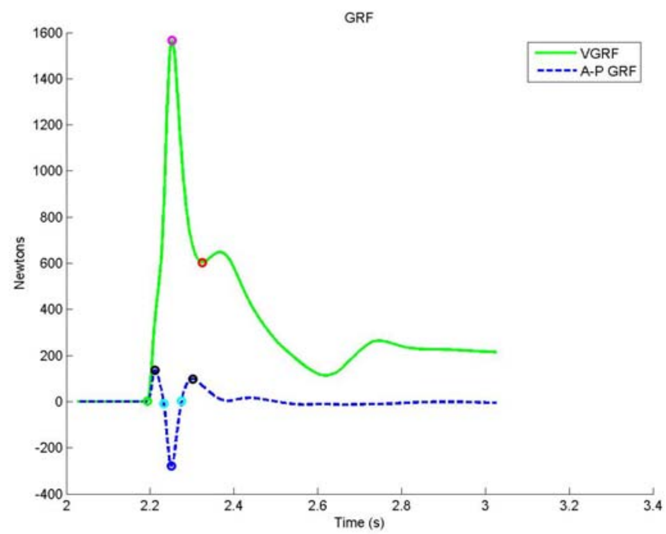
	Date of last participation	Months	Years
Basketball			
Soccer			

Physical Activity Questionnaire

Activity	***Total Yrs*** you've engaged in this activity during your life	***How often*** do you now engage in this activity? **1** = rarely; **2**= 5-10 times/yr **3** = 2-3 times/mo. **4** = 2-3 times/wk **5** = 4 or more times/wk	What is the length of time ***(in minutes)*** you spend each time you engage in this activity?	How would you perceive your level of physical exertion during this activity: **1** = very low effort, **2 =** moderate effort, **3** = fairly effortful; **4** = extremely effortful	*Skill level*: If the activity is a sport, rank your highest level of skill attained: **NA** = not applicable **1** = never have competed; **2** = recreational competition; **3** = high school competition; **4** = college competition; **5** = greater skill than college level
Baseball					
Basketball					
Bicycling					
Bowling					
Dance					
Equestrian					
Fitness class					
Gardening					
Golf					
Housework					
Racquetball (and/or squash)					
Running					
Swimming					
Tennis					
Walking					
Weight Lifting					
Yard work					
Others: list below					

Appendix E

Representative GRF Curves



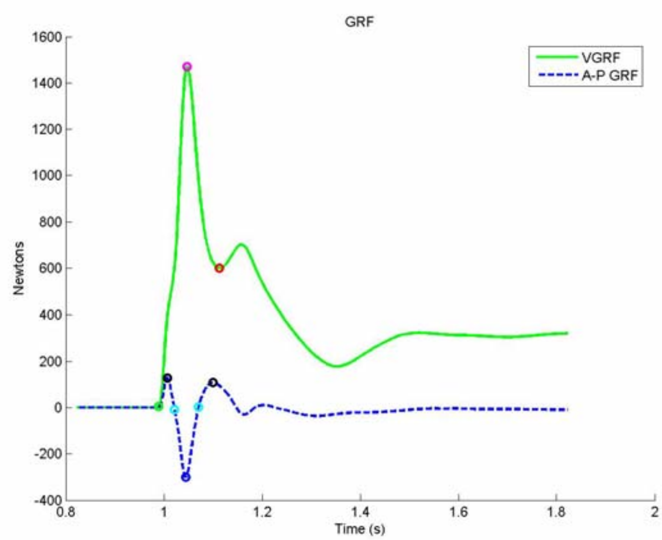
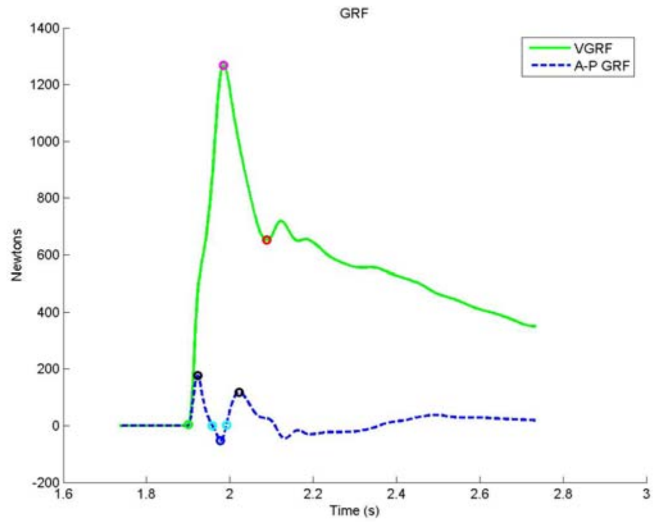


Figure E1. GRF curves for study #1. CON(top) and PERT(bottom) conditions. The beginning and end of the impact phase is defined by the green and red circles, respectively. The pink circle represents the peak VGRF and the royal blue circle represents the peak posterior GRF.



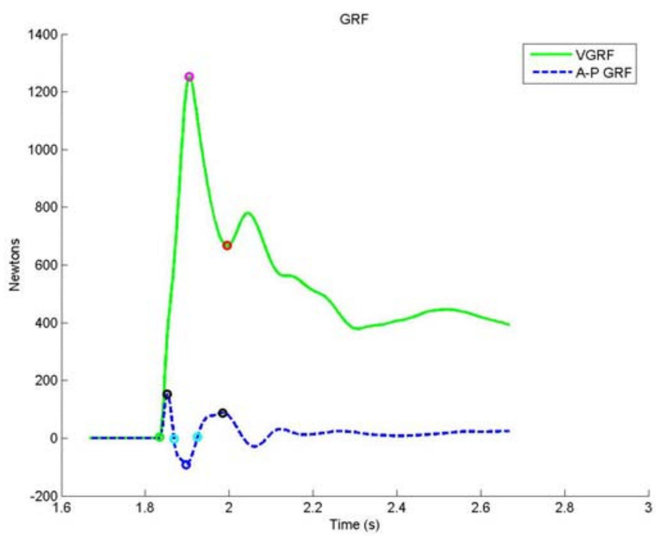


Figure E2. GRF curves for study #2. CON(top) and PERT(bottom) conditions. The beginning and end of the impact phase is defined by the green and red circles, respectively. The pink circle represents the peak VGRF and the royal blue circle represents the peak posterior GRF.

Appendix F

Representative Joint Angle Curves

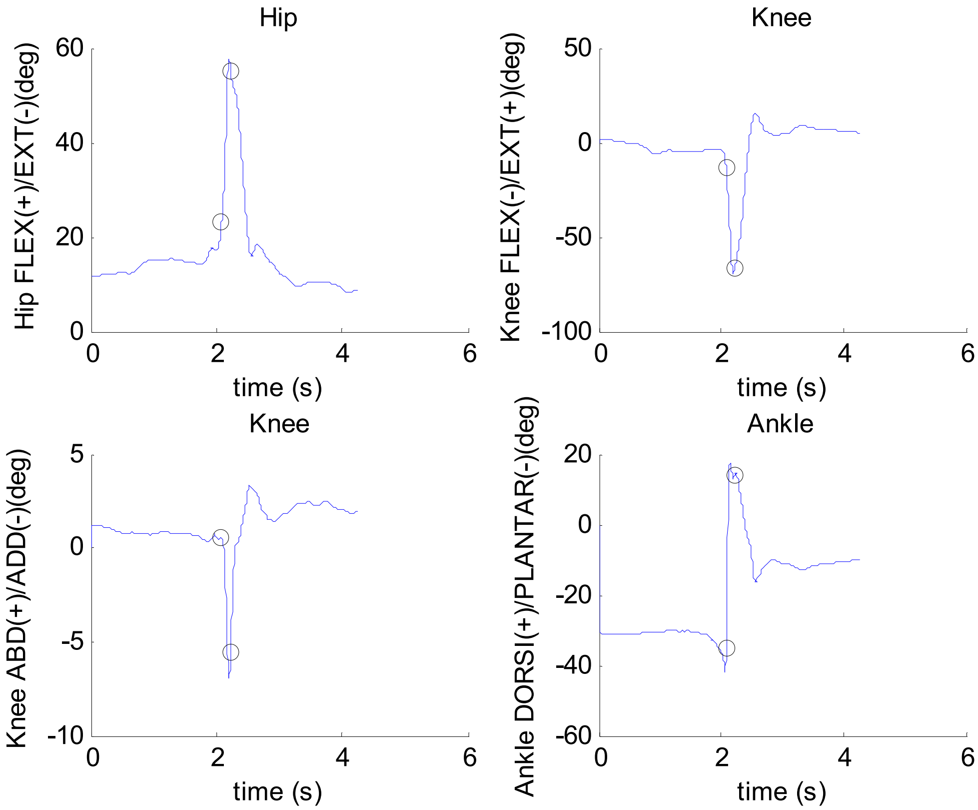


Figure F1. Representative curves for lower extremity joint angles for study #1 (CON). All figures are of the right lower extremity. Hip FLEX/EXT (full extension = 0), Knee FLEX/Ext (full extension = 0), Knee ABB/ADD (neutral = 0), and Ankle DORSI/PLANTAR (neutral = 0).

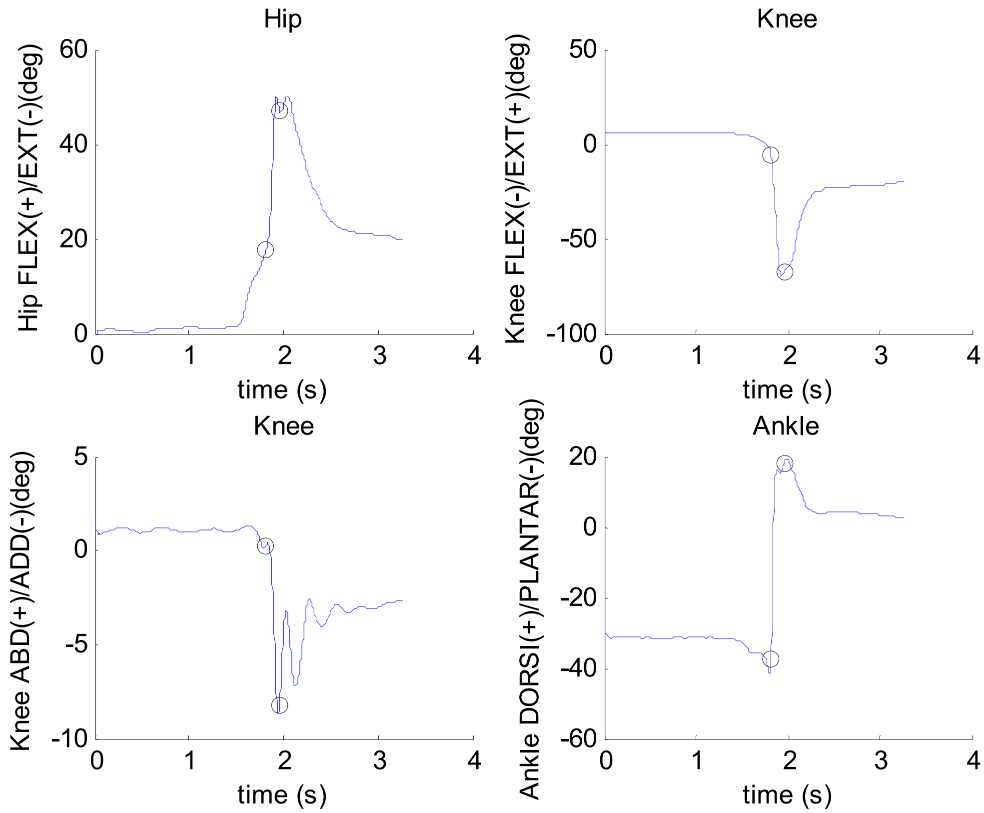


Figure F2. Representative curves for lower extremity joint angles for study #1 (PERT). All figures are of the right lower extremity. Hip FLEX/EXT (full extension = 0), Knee FLEX/Ext (full extension = 0), Knee ABB/ADD (neutral = 0), and Ankle DORSI/PLANTAR (neutral = 0).

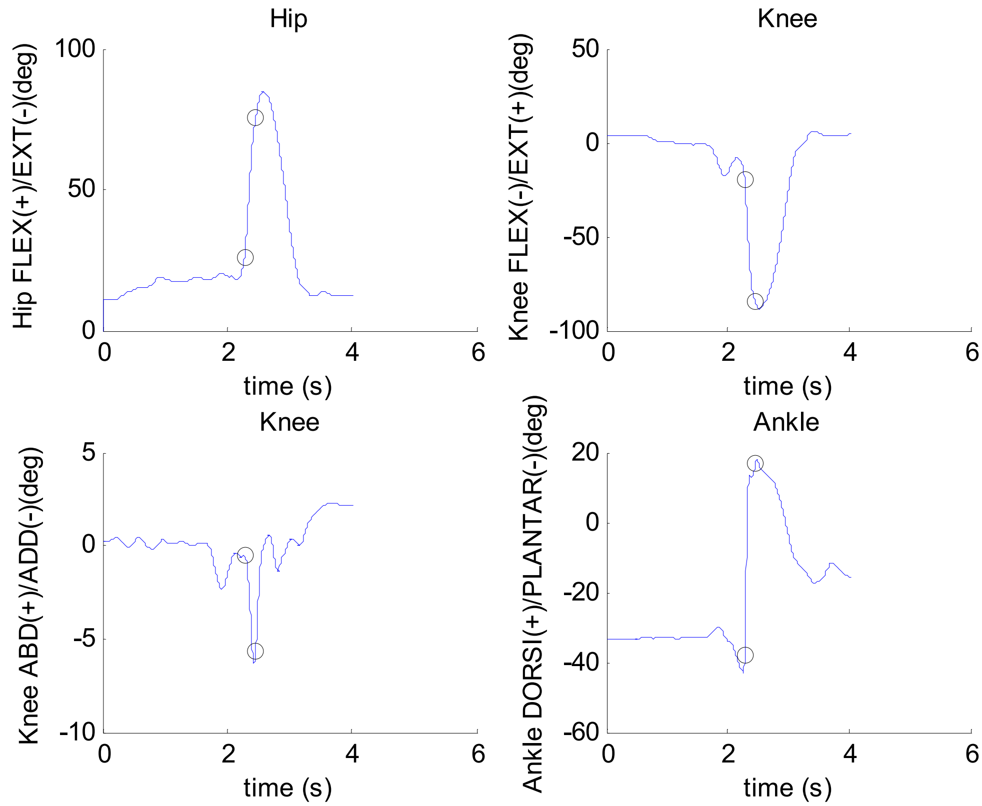


Figure F3. Representative curves for lower extremity joint angles for study #2 (CON). All figures are of the right lower extremity. Hip FLEX/EXT (full extension = 0), Knee FLEX/Ext (full extension = 0), Knee ABB/ADD (neutral = 0), and Ankle DORSI/PLANTAR (neutral = 0).

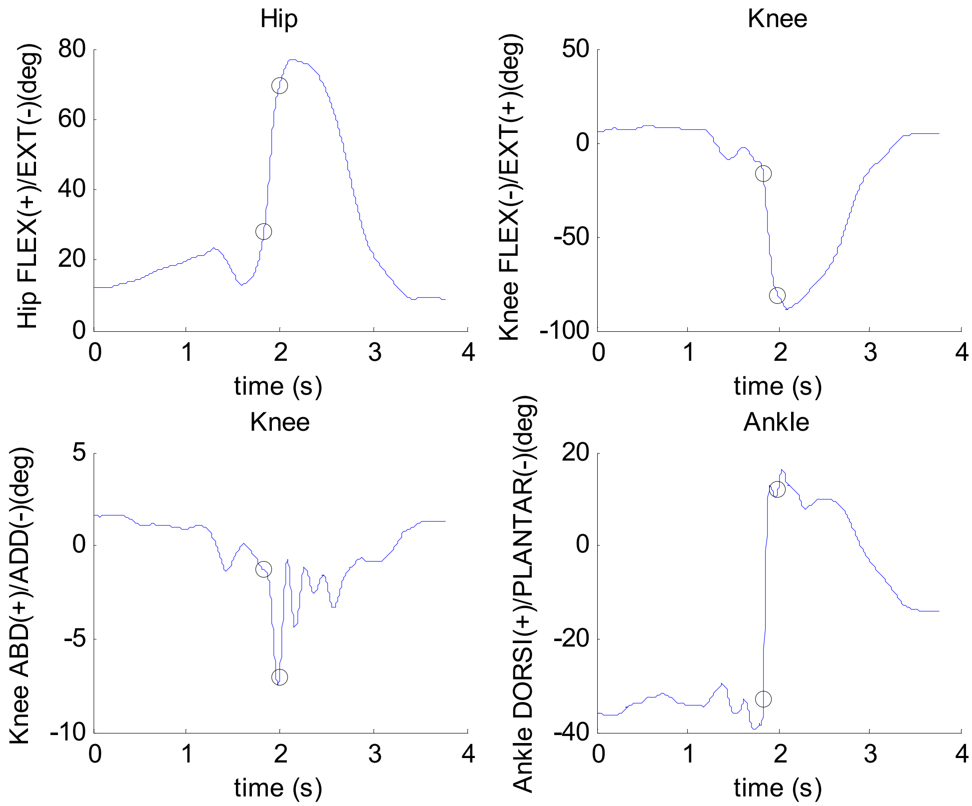


Figure F4. Representative curves for lower extremity joint angles for study #2 (PERT). All figures are of the right lower extremity. Hip FLEX/EXT (full extension = 0), Knee FLEX/Ext (full extension = 0), Knee ABB/ADD (neutral = 0), and Ankle DORSI/PLANTAR (neutral = 0).