The influence of ankle joint stiffness and range of motion on lower extremity biomechanics during a jump landing task.

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ABSTRACT

The influence of ankle joint stiffness and range of motion on lower extremity biomechanics during a jump landing task.
(Under the direction of J. Troy Blackburn)

Injury to the anterior cruciate ligament (ACL) is prevalent and detrimental in the physically active population. Previous research has identified lesser knee flexion displacement, greater knee valgus displacement, and greater vertical and posterior ground reaction force (GRF) as biomechanical factors that are associated with ACL injury. Triceps surae muscle stiffness may have an influence on landing biomechanics based on existing literature suggesting greater lower extremity joint displacements and lesser vertical GRF with soft landings. Similarly, existing literature has suggested lesser ankle dorsiflexion (DF) range of motion (ROM) may influence greater knee valgus displacement. Nonetheless, the influence of these two variables on lower extremity biomechanics during a jump landing has not been investigated. The purpose of this study was to determine the influence of triceps surae muscle stiffness and ankle DF ROM on lower extremity biomechanics during a jump landing task. Thirty-five physically active subjects volunteered for this study. Triceps surae muscle stiffness was assessed using the damped frequency oscillation method, ankle dorsiflexion range of motion was assessed using a goniometer, and knee biomechanics of the jump landing were assessed using an infrared high-speed camera system. Individuals who displayed lesser triceps surae muscle stiffness demonstrated lesser vertical ground reaction forces. Individuals that displayed greater passive straight-knee ankle dorsiflexion range of motion demonstrated greater knee flexion displacement, and lesser vertical and posterior ground reaction forces. As adaptations in muscle stiffness and ROM may be induced over time, the influence of stiffness and ROM on biomechanical factors associated with greater ACL injury-risk suggest these variables should be considered with ACL injury prevention.

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CHAPTER 1

INTRODUCTION

The anterior cruciate ligament (ACL) is a primary contributor to knee joint stability, thus injury to this structure may cause instability (Markolf et al., 1995). Most ACL injuries, up to 70%, have been attributed to a non-contact mechanism (Agel et al., 2005; Mountcastle et al., 2007) involving isolated planting, pivoting, and jumping or a combination thereof (Arendt et al., 1999). The majority of research studies have attempted to recreate this mechanism by using a jump landing task and measuring the associated lower extremity biomechanics. Lesser knee flexion displacement (Hewett et al., 2005; Schmitz et al., 2007; Yu et al., 2006), greater knee valgus displacement (Bell et al., 2008; Hewett et al., 2005; Vesci et al., 2007), greater vertical (Hewett et al., 2005; Schmitz et al., 2007; Sell et al., 2007; Yu et al., 2006) and posterior ground reaction force (Sell et al., 2007; Yu et al., 2006) have been suggested as biomechanical ACL injury risk factors inherent to the jump landing task.

The ankle joint motion of dorsiflexion (i.e. movement of the foot and toes toward the leg) lengthens the triceps surae (calf) muscles. These muscles respond to the imposed lengthening by generating tensile force. Stiffness refers to the ratio of change in force to change in length that a muscle experiences during contraction or joint motion (Δ force/Δ length) (Padua, 2003). Research concerning the influence of stiffness on the ACL has focused on investigating stiffness characteristics of the structures that surround the knee, reporting that greater stiffness will allow for greater biomechanical stability (Blackburn et al., 2006; Blackburn et al., 2004a; Granata et al., 2002). However, the influence of stiffness at an adjacent joint may be associated differently. Kinetic energy (i.e. ground reaction and inertial forces) during any planting, pivoting, or jumping task will be received initially at the joint that is most distal upon ground contact. Any energy that is not absorbed at this joint will continue to travel up the kinetic chain to the next proximal joint, and this process will continue until all energy has been absorbed by the body. Literature concerning different landing techniques has identified that a soft landing is characterized by greater energy absorption, lesser ground reaction force (Devita & Skelly, 1992; Self & Paine, 2001; Zhang et al., 2000) and greater sagittal plane motion at the ankles and knees (Devita & Skelly, 1992; Zhang et al., 2000). Based on the definition of stiffness, greater energy absorption and ankle motion is a result of lesser stiffness at the ankle (i.e. triceps surae). A stiff landing technique results in less motion at the ankles (Zhang et al., 2000), translating into greater stiffness (Self & Paine, 2001) and poor energy absorption. Thus greater triceps surae muscle stiffness may be associated with increased ground reaction forces and less knee and ankle motion upon landing.

Muscle extensibility is measured as the total range of motion that is available at a joint without considering the amount of resistive force while the motion is elicited (Padua, 2003). Previous literature has associated lesser triceps surae extensibility, or ankle dorsiflexion (DF) range of motion (ROM), with greater knee valgus when performing a controlled squat task (Bell et al., 2008; Vesci et al., 2007). Knee valgus during a controlled squatting task has been reported to diminish when performed on a decline wedge (Bell et al., 2008), indicating the wedge effectively shortens the triceps surae allowing greater DF ROM. Conversely, greater knee valgus was demonstrated when performing a dynamic drop landing task onto an incline wedge, due to the fact that the wedge lengthens the tricep surae resulting in lesser ankle DF ROM (Hagins et al., 2007). These data suggest that lesser ankle DF ROM may be associated with greater knee valgus displacement with a jump landing.

In summary, ACL injury has been linked to lesser knee flexion displacement, greater knee valgus displacement, and greater vertical and posterior ground reaction forces (GRF). Triceps surae muscle stiffness and ankle DF ROM have been suggested to have an influence on these ACL-injury risk factors. The purpose of this study was to investigate the influence of triceps surae muscle stiffness and ankle DF ROM on these four biomechanical variables during a jump landing task. We hypothesized that lesser triceps surae muscle stiffness and greater ankle DF ROM would be associated with greater knee flexion displacement, lesser knee valgus displacement, lesser vertical and posterior GRF.

Research Questions and Hypotheses

1. RQ_1: What is the relationship between triceps surae muscle stiffness and knee flexion displacement during the loading phase of a jump landing?

   H_1: There is a significant negative relationship between triceps surae muscle stiffness and knee flexion displacement.

2. RQ_2: What is the relationship between ankle dorsiflexion ROM and knee flexion displacement during the loading phase of a jump landing?

   H_2: There is a significant positive relationship between ankle DF ROM and knee flexion displacement.

3. RQ_3: What is the relationship between triceps surae muscle stiffness and knee valgus displacement during the loading phase of a jump landing?

   H_3: There is a significant positive relationship between triceps surae muscle stiffness and knee valgus displacement.

4. RQ_4: What is the relationship between ankle DF ROM and knee valgus displacement during the loading phase of a jump landing?

   H_4: There is a significant negative relationship between ankle DF ROM and knee valgus displacement.

5. RQ_5: What is the relationship between triceps surae muscle stiffness and peak vertical GRF during the loading phase of a jump landing?

   H_5: There is a significant positive relationship between triceps surae muscle stiffness and peak vertical GRF.

6. RQ_6: What is the relationship between ankle DF ROM and peak vertical GRF during the loading phase of a jump landing?

   H_6: There is a significant negative relationship between ankle peak ROM and peak vertical GRF.

7. RQ_7: What is the relationship between triceps surae muscle stiffness and peak posterior GRF during the loading phase of a jump landing?

   H_7: There is a significant positive relationship between triceps surae muscle stiffness and peak posterior GRF.

8. RQ_8: What is the relationship between ankle DF ROM and peak posterior GRF during the loading phase of a jump landing?

H_8: There is a significant negative relationship between ankle DF ROM and peak posterior GRF.

Predictor Variables

- triceps surae muscle stiffness (N/cm)
- ankle DF ROM (˚)

Criterion Variables

- knee flexion displacement (˚)
- knee valgus displacment (˚)
- peak vertical ground reaction force (N)
- peak posterior ground reaction force (N)

Operational Definitions

muscle stiffness (k) where $k = 4\pi^2 mf^2$ (Padua, 2003)

- m: mass of shank and foot segment
- k: active muscle stiffness
- f: damped frequency of oscillation

ankle DF ROM – passive dorsiflexion range of motion measured using a goniometer with the knee at 0˚ (straight-knee) and 90˚ of flexion (bent-knee)

knee flexion displacement – change sagittal plane knee angular position from initial contact to peak knee flexion

knee valgus displacement – change frontal/coronal plane knee angular positon from initial contact to peak knee valgus

peak vertical ground reaction force – maximum vertical force generated by the ground on the subject in the vertical direction *peak posterior ground reaction force* – maximum force generated by the ground on the subject in the posterior direction

jump landing task – a double-legged jump landing off a 30 cm box placed 40% of the subject height away from a target

initial ground contact – the instant that the subject comes in contact with the ground

loading phase – the time duration from initial ground contact to peak knee flexion during the jump landing.

Assumptions, Limitations, and Delimitations

Assumptions

1. Instruments for measuring muscle stiffness, ankle ROM, and knee kinematics and kinetics were valid and reliable.
2. Subjects and testers strictly followed the designed protocol.

Limitations

1. Methods did not attempt to control the maximum amount of physical activity subjects participated in weekly.
2. Methods did not attempt to manipulate the amount of training an individual has previously received prior to enrollment in the study.
3. Methods did not attempt to manipulate the technique the subject used to land.

Delimitations

1. Subjects consisted of recreationally active males and females between the ages of 18-30 that participate in physical activity 3 days a week for 20 minutes.
2. Subjects had no previous history of acute lower extremity injury within the last 6 months.

3. Subjects had no history of lower extremity surgery.

4. Subjects had no existing of chronic injuries.

CHAPTER 2

REVIEW OF LITERATURE

Introduction

The anterior cruciate ligament (ACL) is an important structure that is responsible for the stability of the knee (Markolf et al., 1995; Noyes et al., 1983). Over 70% of all injuries to the ACL occur during athletic participation (Smith et al., 1993), making up 20.3% of all knee injuries (Majewski et al., 2006). Injury to the ACL leads to instability and increases the risk of damage to additional structures of the knee (Markolf et al., 1995). A study conducted by Noyes et al. (1983) reported that of the individuals who suffered ACL injuries, 31% experienced difficulty with walking, 44% with activities of daily living, and 57% with straight-ahead running. Trauma to the ACL is common among the physically active, and has a detrimental effect on quality of daily living and physical function (Noyes et al., 1989).

In addition to the high frequency of injury, ACL injury rates differ across sex (Arendt et al., 1999) with females demonstrating a 1.5-6 times greater risk of injury (Hewett et al., 2005; Mountcastle et al., 2007). Females participating in the same sports as males are subject to a 3-times greater risk of suffering an ACL injury (Prodromos et al., 2007). The prevalence and sex discrepancy of ACL injury have driven researchers to identify the mechanisms and factors that are associated with ACL injury in hopes of reducing the number of occurrences, especially among women.

The literature has commonly identified the primary mechanism of ACL injury as non-contact in nature, representing a reported 70.5% (Mountcastle et al., 2007) to 78% (Noyes et al., 1983) of all ACL injuries. With respect to sex, Agel et al. (2005) reported in a 13-year epidemiological study that 58% of male and 67% of female ACL injuries involve a non-contact mechanism. Specific description of the non-contact mechanism involves planting, pivoting, and jumping, either independently or in combination (Arendt et al., 1999). The most common mechanism of injury was reported to be the combination of planting and pivoting, a mechanism that is associated with 57.1% of all ACL injuries (Arendt et al., 1999). Research studies have attempted to recreate these mechanisms through the use of controlled tasks that mimic sport movements and analysis of lower extremity biomechanics as they are performed. Some of the biomechanical factors that the literature associates with ACL injury are anterior tibial shear force (D. L. Butler et al., 1980; Chappell et al., 2002; Markolf et al., 1995; Sell et al., 2007; Yu et al., 2006), knee flexion displacement (Chappell et al., 2005; Chappell et al., 2002; Hewett et al., 2005; Sell et al., 2007; Yu et al., 2006), knee valgus displacement (Ford et al., 2005; Hewett et al., 2005), vertical ground reaction force (Hewett et al., 2005; Yu et al., 2006), and posterior ground reaction force (Sell et al., 2007; Yu et al., 2006). Specifically, lesser knee flexion displacement, greater knee valgus displacement, and greater vertical and posterior ground reaction force have been associated with a heightened risk of ACL injury.

Biomechanical ACL Risk Factors at the Knee

Anterior Tibial Shear Force

Anterior tibial shear force (ATSF) is defined as the amount of shear force directed anteriorly at the tibiofemoral joint. Proximal anterior tibial shear force (ATSF) is reported to place direct strain on the ACL (Berns et al., 1992; Markolf et al., 1995). Markolf et al. (1995) conducted a study involving 14 fresh frozen cadaver knees. Each knee was attached to a device that controlled for flexion-extension, internal-external rotation of the tibia, valgus-varus, and anterior tibia shear force at the knee. A series of single-load (e.g. ATSF, valgus, or varus only) and paired-load tests (e.g. ATSF-varus or valgus-internal rotation) were performed starting with the knee at 90º moving toward 5º of hyperextension at 10º increments. ATSF placed the most strain on the ACL under single-load conditions, the combination of ATSF and internal tibial rotation increased ACL load near full knee extension, while ATSF and a valgus moment increased ACL load at knee flexion angles greater than 10º. A study conducted by Berns et al. (1992) reported similar results with respect to isolated and combinations loads. During clinical evaluation the ACL is the primary ligamentous restraint against isolated ATSF and provides approximately 86% of the total resistance (D. L. Butler et al., 1980). A large ATSF may load the ACL excessively, resulting in damage (Chappell et al., 2002) and the possibility of ligament rupture (Sell et al., 2007).

A study conducted by Yu et al. (2006) examined the relationships between various biomechanical factors during the landing phase of a stop-jump task. Thirty healthy male and female college students performed a stop-jump task, consisting of a two to three-step approach run followed by a double leg vertical hop. Female subjects displayed significantly greater ATSF, greater peak vertical ground reaction force, and lesser knee flexion angle. Chappell et al. (2005) conducted a similar study and investigated the effects of fatigue. Ten recreationally-active male and female subjects performed three jump tasks before and after a fatigue protocol. The fatigue protocol consisted of multiple sets of vertical jumps followed by sprints. Both sexes displayed significantly increased ATSF and decreased flexion angles at the knee when landing under fatigued conditions. The results of both of these studies suggest the association of high ATSF with decreased knee flexion angles during a jump task. It becomes logical to investigate factors that are associated with ATSF because the literature has established that ATSF places direct strain on the ACL.

Vertical Ground Reaction Force

Vertical ground reaction force (GRF) is defined as the total force exerted by the ground on the subject in the vertical direction in reaction to the landing force applied by subject to the ground. Hewett et al. (2005) conducted a prospective study to determine the association between a series of biomechanical factors and ACL injury. Two hundred and five adolescent female athletes performed a drop vertical jump task. Athletic exposure and injury were surveyed for a constant number of seasons depending upon the sport. During this time period, 9 of the 205 individuals suffered ACL injures. These individuals displayed a significantly greater vertical GRF, approximately 20%, when compared to uninjured individuals. A significant positive correlation was also found between peak vertical GRF and knee valgus angle in the injured cohort, another factor that has been linked to ACL injury. As mentioned within a previous section, Markolf et al. (1995) reported the combination of ATSF with knee valgus, as opposed to an isolated ATSF, further increases the strain placed on the ACL when the knee is in flexion. The moderate association between peak vertical GRF and knee valgus angle emphasizes the multifactorial nature of ACL injury. To provide further rationale for measuring vertical GRF, another study conducted by Yu et al. (2006) involved 60 healthy college students who each performed a series of stop-jump tasks in attempt to investigate the relationships between various biomechanical factors at the hip and knee. Yu et al. (2006) found a significant positive correlation between peak vertical GRF and peak ATSF ($r = 0.53$, $p < 0.001$). Thus larger vertical GRFs may be associated with larger ATSFs and strain on the ACL. The results of both studies provide a rationale for measuring vertical GRF within the present study.

Posterior Ground Reaction Force

Posterior GRF is defined as the amount of force exerted on the subject by the ground in the posterior direction in reaction to the force exerted by the subject on the ground. Sell et al. (2007) conducted a prediction study to identify the relationships between ATSF and a series of biomechanical variables. Thirty-six high school basketball players performed a vertical stop-jump. Peak posterior GRF was one of six variables that significantly predicted ATSF. Other variables included knee flexion/extension moment, knee flexion angle, electromyographic (EMG) activity of the vastus lateralis, and sex. The regression equation using these six variables accounted for 86.1% of the variance in ATSF. A negative correlation between peak posterior GRF and ATSF was also reported (Sell et al., 2007). As mentioned within the previous section, the study conducted by Yu et al. (2006) investigated the relationships between lower extremity biomechanical factors during a stop-jump task. Results indicated a moderate positive correlation between peak posterior and vertical GRF. A strong positive correlation was also found between peak posterior GRF and ATSF. The opposing direction of the relationships between posterior GRF and ATSF in studies by Sell et al. and Yu et al. are due to differences in defining the posterior direction. Ultimately, both investigations suggest greater posterior GRF is associated with greater ATSF. As existing literature reports that ATSF places a direct strain on the ACL, it provides rationale to account for peak posterior GRF due to the strong association reported between the two variables.

Knee Flexion Angle

Knee flexion angle is defined as the angle formed at the knee by the thigh and leg in the sagittal plane. Chappell et al. (2005) conducted a study utilizing a stop-jump task to assess kinematic and kinetic changes in recreationally active subjects before and after a specific fatigue protocol. Subjects displayed decreased knee flexion angles during a stop-jump landing under fatigued conditions. A lesser knee flexion angle increases the patella tendon insertion angle at the tibia (Zheng et al., 1998), causing a force applied through the patella tendon to have a greater anterior component (Blackburn & Padua, 2008). Thus the researchers suggested that fatigue could lead to increased strain on the ACL, induced by a decreased knee flexion angle. Hewett et al. (2005) reported in the previously mentioned study that lesser peak knee flexion angle during initial contact of a jump landing was associated with greater ACL injury risk. Sell et al. (2007) reported similar results in as study involved both male and female high school basketball players performing a vertical stop-jump task. A stepwise multiple regression model determined that knee flexion angle at peak posterior ground reaction force was one of six significant variables that when combined could strongly predict ATSF. In summary, the literature suggests that both peak knee flexion angle and knee flexion angle at peak posterior ground reaction force are associated with ACL injury.

Knee Valgus Angle

Knee valgus angle is defined as the angle formed at the knee by the thigh and leg in the frontal plane. Cadaver studies have reported valgus forces applied in positions of knee flexion place greater strain on the ACL compared to any isolated and combination of forces. (Berns et al., 1992; Markolf et al., 1995). Hewett et al. (2005) reported that individuals who sustained ACL injury exhibited knee valgus angles which were 8° greater on average compared to uninjured individuals when performing a drop-vertical jump. A logistic regression analysis revealed that knee valgus moment and knee valgus angle were significant predictors of ACL injury risk, while linear regression analysis revealed that the combination of external knee valgus moment, knee valgus angle, and side-to-side differences in these variables (i.e. dominant compared to non-dominant leg) displayed a strong ability to predict ACL injury with an R^2 value of 0.88. The results suggest that frontal plane motion plays a significant role in ACL injury, and provide rationale for such motion to be accounted for when investigating the variables that cause ACL injury.

Ford et al. (2005) utilized an unanticipated cutting maneuver to determine if valgus differences were apparent across sex. The task was initiated and controlled by a computer display and consisted of the subject in a double-leg starting stance with his or her knees flexed at approximately 45º. Upon computer prompting, the subject would jump forward and cut either left or right as directed by an indicator. The direction displayed on the indicator was purposely delayed to prevent anticipation. Lower extremity kinematic and kinetic data were collected from 54 male and 72 female middle and high school basketball players. During unanticipated cutting maneuvers, females displayed a greater mean knee valgus angle at initial contact, but neither knee flexion angle at initial contact or maximum flexion angle differed across sex. These results combined with those reported by Hewett et al. (2005) suggest that greater knee valgus may contribute to the increased ACL injury risk among females and provide rationale for the measurement of knee valgus angle in the present study.

In summary, a variety of biomechanical factors have been associated with ACL injury. The identified factors include anterior tibial shear force (D. L. Butler et al., 1980; Chappell et al., 2002; Markolf et al., 1995; Sell et al., 2007; Yu et al., 2006), peak knee flexion angle (Chappell et al., 2005; Chappell et al., 2002; Hewett et al., 2005; Sell et al., 2007; Yu et al., 2006), peak knee valgus angle ((Ford et al., 2005; Hewett et al., 2005), vertical ground reaction force (Hewett et al., 2005; Yu et al., 2006), and posterior ground reaction force (Sell et al., 2007; Yu et al., 2006). These biomechanical factors should be measured and analyzed during dynamic tasks to further determine the strength of their association with ACL injury.

Muscle Stiffness and Injury

Biomechanical stability (BS) is defined as the ability of a joint to uphold a state of equilibrium in reaction to an external force (Wagner & Blickhan, 1999). It has been proposed that superior biomechanical stability contributes to a greater ability of a joint to maintain equilibrium when perturbed (Padua, 2003). Muscle stiffness has been identified as an essential factor that is required to achieve biomechanical stability (Padua, 2003), and is defined as the ratio of the change in force to the change in muscle length (R. J. Butler et al., 2003; McNair et al., 1992). Muscle stiffness can be divided into active and passive components. Passive muscle stiffness (PMS) refers to stiffness of non-contractile tissue, while the effect of active muscle stiffness (AMS) is the result of the combined stiffness properties of contractile and non-contractile elements. Passive muscle stiffness independently has been suggested to be inefficient in achieving biomechanical stability (Wagner & Blickhan, 1999), whereas AMS has been suggested as the main contributor to biomechanical stability (Padua, 2003). The number of parallel cross-bridges formed within the involved muscle at the particular joint range of motion has been proposed to be the key factor in determining the amount of AMS (Morgan, 1977). In theory, a muscle that displays a lesser amount of stiffness will undergo greater lengthening when acted upon by a given force compared to a stiffer muscle. A joint where the surrounding musculature displays lesser stiffness will respond to an external perturbation with an increased change in joint angle when compared to one with greater stiffness. With respect to a single joint, heightened stiffness may limit the amount of accessory joint motion and inert tissue strain (Padua, 2003). A common example is where heightened stiffness of the hamstrings would be more resistive to a change in length, thus limiting the amount of anterior tibial translation and ACL strain. Achieving superior muscle stiffness at the target joint may be a key contributor in the prevention of injury.

Measurement of Muscle Stiffness

Numerous methods have been utilized for the measurement of muscle stiffness. The transient oscillation method is commonly used for active muscle stiffness measurements involving a single joint (Jennings & Seedhom, 1998; McNair et al., 1992). This method is derived from a model introduced by McNair et al. (1992) that views the target joint as the axis of a mass-spring system with a single degree of freedom. The stiffness of the surrounding musculature at the target joint dictates the overall stiffness properties of the spring. When the system receives an external pertubation, the system will oscillate at a frequency dictated as a function of the mass and stiffness of the spring. McNair et al. (1992) noted that the viscoelastic properties of the involved musculature provide a damping element that causes exponential decay of the oscillatory motion when the system is perturbed. The equation posed by McNair et al. (1992) is written as $k = 4\pi^2 mf^2 + c^2 4/m$ where *k* represents total stiffness, *m* represents the mass of the system, *f* represents the damped frequency of oscillation, and *c* represents the coefficent of damping. As Jennings and Seedhom (1998) have reported that the damping element provides less than a 5% difference to the final stiffness result, it has been omitted in many research studies thereafter (Blackburn et al., 2004b; Blackburn et al., 2006; Blackburn et al., 2004a; Jennings & Seedhom, 1998). Padua (2003) mentioned a major limitation of this method lies in the assumption that only the target muscles are being activated during testing. Numerous research studies have utilized EMG in an attempt to ensure antagonists are not activated.

Research Studies in Muscle Stiffness

Research studies concerning muscle stiffness have compared stiffness of the knee flexor and triceps surae musculature across sex. Blackburn et al. (2004a) conducted a study using healthy subjects to investigate differences in active and passive knee flexor stiffness. Fifteen males and 15 females were individually set up on an isokinetic dynamometer. The device assessed PMS while the subject was instructed to relax by measuring resistance against an isokinetic lever arm moving at 5º per second. EMG activity of the knee flexors was measured and compared to baseline to ensure no active muscle force was generated. The slope of the moment-angle curve was defined as PMS.

During active muscle stiffness measurements, the subject was instructed to maintain the same level of contraction while an external perturbation was applied. Oscillatory frequency of the system was determined from the tangential acceleration of an accelerometer that had been distally attached to the system and was used to determine AMS. Females displayed less mean AMS and PMS when compared to males (Blackburn et al., 2004a). Blackburn et al. (2004b) analyzed the same data using a stepwise multiple regression and reported a significant moderate relationship between PMS and AMS with an R^2 value of 0.249. Additional research was conducted by Blackburn et al. (2006) to compare structural stiffness and material modulus across sex, where material modulus refers to a measure of stiffness where anthropometric differences between sex are accounted for. Twenty male and 20 female subjects were placed on a triceps surae loading device and seated with 90º of hip and knee. The device involved an adjustable lever arm that rested on the subject's distal femur and was loaded with a weight that was equivalent to 30 ± 5% of the ground reaction force obtained during maximum contraction. The metatarsal heads were placed on the edge of a wooden plank that was rigidly secured to a force plate. With the subject blindfolded and wearing headphones that played static noise to prevent anticipation, an external perturbation was applied randomly during each 10-second interval. Subjects were instructed to maintain the same level of contraction. The vertical ground reaction force output generated from each trial was used to determine the damped oscillation frequency. Structural stiffness was calculated using the equation derived by McNair et al. (1992) described previously. To account for anthropometric differences across sex, material modulus was calculated as the ratio of stress to strain using estimates to determine the cross sectional area for the triceps surae.

Blackburn et al. (2006) reported that males tend to display greater active muscle stiffness and material modulus (Blackburn et al., 2006; Kubo et al., 2003). A similar study was conducted by Kubo et al. (2003) using a ramped maximal voluntary isometric contraction in plantar flexion, while measuring tendon elongation using ultrasonography and calculating stiffness as the ratio of change in isometric force to change in tendon length. Kubo et al. (2003) reported findings that were consistent to those reported by Blackburn et al. (2006) and suggested that males demonstrate greater active triceps surae muscle stiffness.

The bulk of the literature has focused on establishing differences in muscle stiffness without studying the physiological implications of these differences. Although the literature reports that males tend to display greater knee flexors (Blackburn et al., 2004b; Blackburn et al., 2004a), triceps surae (Blackburn et al., 2006; Kubo et al., 2003), and total leg stiffness (Granata et al., 2002) compared to females, no empirical evidence has been reported as to how these differences affect factors that have been associated with injury. Theoretical links have been made stating that because increased stiffness contributes to increased biomechanical stability and decreased accessory joint motion the chances of injury should decrease. These links have been made with respect to the musculature directly surrounding the target joint. In other words, it has been suggested that greater stiffness of the knee flexors is associated with a lesser risk of ACL injury. This theory may not hold true, however, when considering how stiffness of an adjacent joint affects the target joint. The transfer of kinetic energy throughout the body must be considered for such a theory to be developed.

Energy Absorption During Landing

Kinetic energy during any planting, pivoting, or jumping task will first be received at the joint that is most distal upon ground contact. Any energy that is not absorbed at this joint will be transferred up the kinetic chain to the next proximal joint, and this process will continue until all energy has been absorbed by the body. Self & Paine (2001) measured lower extremity kinematics and kinetics during a landing from a 12-inch height with multiple techniques. The four landing styles included a flexed-knee with natural plantarflexion, stiff-knee with natural plantarflexion, stiff-knee with rigid plantarflexion, and stiff-knee with a heel-first ground strike. Landing with a more plantarflexed technique, as opposed to a dorsiflexed technique, resulted in decreased vertical ground reaction force, increased Achilles' tendon force, and lower Achilles' tendon stiffness. Zhang et al. (2000) conducted a similar study where three techniques and drop heights (0.32 m, 0.62 m, 1.03 m) were used. A soft landing technique was associated with a significantly smaller vertical ground reaction force and greater range of motion at the ankle, knee, and hip when compared to a stiff landing. The results of these two studies suggest that greater ankle joint range of motion may be associated with improved energy absorption. Furthermore as previous literature indicates lesser stiffness is associated with greater extensibility, decreased stiffness at the ankle may allow for increased muscular lengthening or range of motion, increased energy absorption, and a decrease in energy transferred to the knee.

Muscle Extensibility and Injury

Muscle extensibility is defined as the "total range of motion at a joint, without eliciting the generation of muscular force" (Gleim & McHugh, 1997) or the "available range of motion at a joint and does not take into consideration the amount of resistive force during muscle lengthening" (Padua, 2003). Based on the definition of stiffness (stiffness = Δforce / Δlength), Blackburn et al. (2004a) theorized that greater muscle extensibility would be associated with lesser stiffness and later found that PMS represented a moderate negative relationship with active extensibility (Blackburn et al., 2004b). At the other end of the spectrum, decreased muscle extensibility may be related to musculotendinous injury (Thacker et al., 2004) due to improper energy absorption within the body. Optimal muscle extensibility will result in the maximal ROM of a joint and allow for functional activities to be performed (Gajdosik, 2007). Hirth (2007) has suggested that overhead functional squat testing may be used clinically to effectively identify muscular tightness and weakness. In particular, medial knee displacement (MKD) or knee valgus while performing such a task has been associated with lateral gastrocnemius, soleus, and peroneal tightness (Hirth, 2004).

Ankle dorsiflexion (DF) range of motion (ROM) has been identified as a factor that may influence ACL injury (Bell et al., 2008; Vesci et al., 2007). Bell et al. (2008) conducted a study that identified strength and flexibility measures among individuals who displayed excessive knee valgus when performing a controlled double-leg squat task and compared them to individuals who did not. Individuals who demonstrated greater knee valgus tended to demonstrate lesser DF ROM. Additionally, lesser knee valgus was observed when increasing DF ROM artificially by placing a decline wedge under the heels of these subjects. A similar study conducted by Vesci et al. (2007) showed consistent findings as those reported by Bell & Padua (2008). Hagins et al. (2007) applied similar concepts to a dynamic drop-landing task and reported increased knee valgus when using an incline wedge. Thus previous literature suggests lesser DF ROM may influence greater knee valgus with both controlled and dynamic tasks. Because knee valgus diplacement has been suggested to be a biomechanical factor that is associated with ACL injury, the influence of ankle DF ROM on knee valgus displacement should be considered with respect to a jump landing.

Influence of Triceps Surae Stiffness and Ankle Dorsiflexion Range of Motion on Landing Biomechanics

The literature has established that a stiff landing technique will result in greater vertical ground reaction forces and lesser ankle dorsiflexion as compared to a soft landing technique (Self & Paine, 2001; Zhang et al., 2000). Yu et al. (2006) found that subjects who landed with greater vertical ground reaction forces also produced lesser knee flexion angles. It may be suggested that greater triceps surae muscle stiffness results in greater vertical ground reaction force and represented by lesser knee flexion and ankle dorsiflexion. In addition, vertical ground reaction force has been reported to display a positive correlation with posterior GRF (Yu et al., 2006), while limits in ankle dorsiflexion have been associated with knee valgus (Bell et al., 2008; Vesci et al., 2007). Thus greater triceps surae stiffness may limit ankle dorsiflexion and result in greater knee valgus, vertical and posterior ground reaction forces. The literature has associated less knee flexion, greater knee valgus, and greater vertical and posterior ground reaction forces with a heightened risk of ACL injury.

Summary

Injury to the anterior cruciate ligament (ACL) is detrimental to physical function and is common among the physically active population. Females have a higher incidence of ACL injury compared to men who participate within the same activities (Hewett et al., 2005). ACL injuries are usually involved with a non-contact mechanism, involving planting, pivoting, and jumping in combination or independently (Arendt et al., 1999). Various biomechanical factors have been associated with ACL loading and ATSF including lesser knee flexion displacement, greater knee valgus displacement, greater vertical and posterior GRF.

Heightened active muscle stiffness of the surrounding joint musculature has proposed to be essential in achieving biomechanical stability and decreasing the risk of injury. Sex differences seem to exist when comparing active triceps surae muscle stiffness and when accounting for anthropometric measures (i.e. material modulus). However, the implications of these findings have not been empirically investigated to identify the association between stiffness and ACL risk factors and potential influence on the greater ACL injury risk in females.

The distribution of energy throughout the lower extremity begins at the foot and ankle and travels up the kinetic chain during a planting, pivoting, or jumping task. The literature loosely establishes that increased joint ROM at the target joint when landing may be associated with more effective energy absorption and lower stiffness at the ankle may also play a role at the knee.

Muscle stiffness is not synonymous with muscle extensibility, but the two are related (Blackburn et al., 2004b). Bell et al. (2008) and Vesci et al. (2007) have found that individuals who display excessive knee valgus during a controlled double-leg squat tend to exhibit decreased ankle dorsiflexion (DF) range of motion (ROM). This knee valgus diminished when DF ROM is artificially increased (Bell & Padua, in press).

Because of its reported effect on knee valgus, the influence of DF ROM on knee valgus and other biomechanical ACL injury factors should be further investigated.

The literature has established that differences exist with respect 1) to active muscle stiffness (AMS) when comparing across sex, 2) ankle dorsiflexion ROM when comparing between individuals with excessive knee valgus and control, and 3) lower extremity biomechanics when comparing across sex and the presence of ACL injury. However, no studies have integrated these findings and determined if associations exist between stiffness, range of motion, and knee biomechanics. Furthermore, the literature does not describe the influence of triceps surae muscle stiffness and DF ROM on knee biomechanics. The purpose of this study is to investigate the association between these factors. In determining the relative influence of each factor, conclusions may be used to explore interventions to decrease the likelihood of ACL injury.

CHAPTER 3

METHODOLOGY

Subjects

Thirty-five recreationally active subjects (17 males, 18 females) ranging from ages 18 to 30 years were recruited from the student population at the University of North Carolina at Chapel Hill for this study. Subjects were asked to complete a questionnaire to confirm that they conform to the selection criteria prior to formal data collection. All subjects read and signed an Informed Consent Form approved by the Institutional Review Board of the University of North Carolina at Chapel Hill.

Inclusion Criteria

Subjects were required to be between the ages 18 and 30. All subjects participated in some sort of physical activity for a minimum of 20 minutes at least three times per week. All subjects were affiliated with the University of North Carolina at Chapel Hill.

Exclusion Criteria

Subjects were required to have no history of any lower extremity injury within the past six months, any type of lower extremity surgery, suffer from any lower extremity chronic injury, nor have a history of any neurological disorders.

Experimental Design

All data were collected in the Sports Medicine Research Laboratory during a single testing session lasting 1.5 hours. Subjects were asked to perform three separate tasks in a counterbalanced order including 1) ankle dorsiflexion range of motion, 2) triceps surae musculotendinous stiffness, and 3) a jump landing task. All data were collected for the subject's dominant leg, which was defined as the preferred extremity a subject used to kick a soccer ball for maximum distance.

Instrumentation

A standard 12" plastic goniometer was used to measure ankle dorsiflexion range of motion. A custom-made triceps surae loading device interfaced with a force plate (Bertec Corp, Columbus, OH) was used to measure active triceps surae muscle stiffness. This device was identical to the one used by Blackburn et al. (2006) and consisted of an adjustable wooden lever arm fixed about a vertical post. A seven camera motion capture system was used to sample kinematic data during the jump landing task (Vicon Motion Systems, Centennial, CO). Data were analyzed using the Vicon Nexus software specifically configured for the system. A Bertec 4060-08 aluminum force plate was used to collect ground reaction force data during the jump landing task.

Procedures

Ankle Dorsiflexion Range of Motion Assessment

Ankle dorsiflexion range of motion measurements were taken using a standard 12" plastic goniometer. The subject was first seated on a padded table with his or her knees hanging off the edge at a 90° flexion angle. The subject was measured for dominant leg passive bent-knee ankle DF ROM while he or she was completely relaxed and the examiner moved the ankle into DF by applying overpressure until a soft end-feel was observed. The axis of the goniometer was placed over the lateral malleolus at the axis of rotation. The stationary arm was aligned with the fibular shaft, while the moving arm was aligned with the head of the fifth metatarsal. Five trials were performed. The subject was then seated with his or her knees fully extended at 180° with their ankles are hanging off the edge of the table. Another five trials were performed in this position.

Triceps Surae Musculotendinous Stiffness Assessment

The procedure used to measure active triceps surae muscle stiffness was identical to that used by Blackburn et al. (2006). Each subject was seated with the hips, knees, and ankles all at 90° angles. The metatarsal heads were placed on the edge of a wooden plank secured to a force plate (Bertec model 4060-08, Bertec Corp., Columbus, OH) such that any plantarflexion force applied to the plank registered as a vertical ground reaction force. A custom-made loading device was used to apply resistance to the system. This device was identical to the one used by Blackburn et al. (2006) and consists of an adjustable wooden lever arm fixed about a vertical post. The distal portion of the lever arm of the loading device was placed over the distal femur. A visual of the setup is presented in Figure 1. A load equaling 10% of the subject's mass was placed on the distal end of the lever arm. The subject was instructed to actively plantarflex the ankle just enough to support the weight of the loading device and maintain a 90° angle at the ankle joint. A carpenter's level was placed on the lever arm to ensure the joint angle. The subject was instructed to close his or her eyes. During the 10-second trial, the distal portion of the lever arm was randomly perturbed downward such that the ankle ankle was forced into dorsiflexion, lengthening the triceps surae and initating oscillatory flexion and extension about the ankle. The subject was instructed "not to intervene" with the perturbation and to maintain the same level of triceps surae contraction as if before the perturbation. The damped frequency of oscillation (f) was calculated as 1 / (t2 – t1), where t1 and t2 represent the initial and secondary peaks of GRFv data immediately after perturbation. An example of the force output is presented in Figure 2. Stiffness will be calculated using the equation $k = 4\pi^2 mf^2$ where *k* represents active muscle stiffness, *m* represents the mass of the foot and shank segment, and *r* is the radius. The mass of the system was assumed as 16.1% of the total body mass (Dempster et al., 1959), where 10 of the 16.1% is attributed the mass that was added to the device and 6.1% is attributed to the mass of the shank and foot segment. Five trials were performed.

Jump Landing Task Assessment

Subjects performed a jump landing task from a 30 cm height. Each subject was fitted with spandex shorts and shirt. Retro-reflective markers were then placed bilaterally over the acromion process, anterior superior iliac spine, greater trochanter, anterior thigh, medial and lateral epicondyle of the knee, anterior shank, medial and lateral malleolus of the ankle, calcaneus, and the first and fifth metatarsal heads using double-sided tape. A marker was also placed over the sacrum at the level of L5-S1. Any reflective areas of the shoes and/or clothing were covered with non-reflective tape. All points were digitized during a static trial while the subject was standing in a calibration area over both force plates to create a segment-linkage model. The knee joint center was defined as the midpoint between markers of the medial and lateral epicondyle. The ankle joint center was defined as the midpoint between the markers of the medial and lateral malleolus. The markers of the medial malleoli and medial epicondyles were then removed for jump landing trials. A box 30 cm height was placed a distance equal to 40% of the subject's height away from the edge of the force plates. Each subject began by standing with the feet shoulder width apart on top of the box facing the force plates. During the task, the subject jumped off the box and landed with both feet on the force plates. If the subject landed with any portion of the foot off the force plate, the trial was discarded and repeated. Lower extremity kinematics and kinetics were sampled during the jump landing task. Subjects were allowed up to three practice trials to familiarize themselves with the task. The first 5 successful trials were used for data analysis.

Data Sampling and Reduction

All force plate data were sampled at a rate of 1500 Hz, while data captured using the Vicon system were sampled at a rate of 150 Hz. The axis system for Vicon was set up such that positive X, Y, and Z values represented forward, leftward and upward directions respectively. Knee joint angles were calculated as Euler angles (YXZ sequence) defined by the shank reference frame relative to the femur reference frame such that flexion, varus and internal rotation of the knee represented positive angular displacements for subjects who were right leg dominant. Knee flexion, valgus and external rotation represented positive angular displacements for subjects who were left leg dominant. Knee flexion displacement, knee valgus displacement, peak vertical and posterior GRF were identified during the loading phase using the Motion Monitor motion capture software (Innovative Sports Training, Chicago, IL). The loading phase was defined as the time from initial contact to peak knee flexion. All ground reaction force data were normalized to body weight measured in newtons. Knee flexion and valgus displacements were calculated as the difference between the minimum and maximum values during the loading phase. Force plate and Vicon data were processed using a fourth order, zero phase lag low-pass Butterworth filter with a cutoff frequency of 10 Hz.

Statistical Analysis

All ROM, biomechanical, and stiffness data for each subject were averaged over all 5 trials. All peak posterior GRF and knee valgus displacement data were adjusted to positive values for statistical analyses. Data were analyzed using the Statistical Package for Social Sciences 16.0 (SPSS, Inc. Chicago, IL). Twelve separate Pearson bivariate correlation analyses were conducted to evaluate the relationships between triceps surae stiffness and ankle DF ROM (predictor variables) and knee flexion displacement, knee valgus displacement, peak vertical and posterior GRF, respectively. Statistical significance was established *a priori* as $\alpha \leq 0.05$.

CHAPTER 4

RESULTS

A total of thirty-five physically-active, healthy subjects (17 males, 18 females) completed data collection. Subject descriptive statistics are presented in Table 1. The normalized stiffness data for one female subject were eliminated from all ankle stiffness correlation analyses as the value was considered to be an outlier (more than three standard deviations above the mean). Descriptive statistics for each dependent variable are presented in Table 2. Scatter plots and trendlines of each potential relationship between data are presented in Figures 1-12.

Primary Research Questions

Stiffness

The results of Pearson bivariate correlation analyses between normalized stiffness and the biomechanical variables of interest are summarized in Table 3. A significant positive correlation was observed between normalized stiffness and normalized vertical GRF ($r = 0.411$, $p = 0.016$), indicating that subjects with greater normalized stiffness demonstrated greater vertical GRFs during the jump landing task. However, no significant associations were observed between normalized stiffness and normalized posterior GRF ($r = 0.319$, $p = 0.066$), knee flexion displacement ($r = -0.123$, $p = 0.489$), or knee valgus displacement ($r = 0.316$, $p = 0.068$). As the correlations between normalized stiffness and posterior GRF and knee valgus displacement, respectively, approached significance, post hoc power analyses were conducted for these data. Observed powers of 0.35 for posterior GRF and 0.47 for knee valgus displacement were detected indicating that 101 and 71 subjects, respectively, would be needed for the posterior GRF and knee valgus displacement analyses in order to achieve a power of 0.80.

Passive Straight-Knee Ankle DF ROM

Correlational analyses between passive straight-knee ankle DF ROM and the biomechanical variables of interest are shown in Table 4. Significant associations were observed between passive straight-knee ankle DF ROM and knee flexion displacement ($r = 0.464$, $p = 0.029$), normalized vertical GRF ($r = -0.411$, $p = 0.014$), and posterior GRF ($r = -0.412$, $p = 0.014$). These results suggest that individuals with greater passive straight-knee ankle DF ROM display greater knee flexion displacement and lesser vertical and posterior GRF. No significant association was observed between passive straight-knee ankle DF ROM and knee valgus displacement ($r = -0.290$, $p = 0.091$).

Passive Bent-Knee Ankle DF ROM

Correlational analyses between passive bent-knee ankle DF ROM and the biomechanical variables of interest are shown in Table 5. No significant associations were observed between passive bent-knee ankle DF ROM and knee flexion displacement ($r = 0.327$, $p = 0.055$), knee valgus displacement ($r = -0.330$, $p = 0.053$), normalized vertical GRF ($r = -0.311$, $p = 0.069$), or normalized posterior GRF ($r = 0.295$, $p = 0.085$). As each of these correlations approached statistical significance, post hoc power analyses were conducted to investigate the non-significant relationships with all the biomechanical variables of interest. Observed powers of 0.51 for knee flexion displacement, 0.51 for knee valgus displacement, 0.38 for vertical GRF, and 0.46 for posterior GRF were detected indicating that 68, 66, 96, and 75 subjects, respectively, would be needed for each analysis to achieve a power of 0.80.

CHAPTER 5

DISCUSSION

Our primary findings suggest that triceps surae stiffness is significantly related to peak vertical ground reaction force during the loading phase of a jump landing. Specifically, lesser triceps surae stiffness is associated with lesser peak vertical GRF. In addition, our results indicate that greater passive straight-knee ankle DF ROM is associated with greater knee flexion displacement, and lesser vertical and posterior GRFs during this same task.

The positive relationship between triceps surae stiffness and peak vertical GRF is in agreement with our hypothesis. Direct comparison with previous literature is difficult, as we are unaware of any previous literature that has evaluated the influence of triceps surae stiffness on knee biomechanics during a jump landing using a methodology similar to the current study. However, several studies on landing techniques have suggested that soft, or less stiff, landings are characterized by lesser vertical GRF, and therefore support the current results (Devita & Skelly, 1992; Self & Paine, 2001; Zhang et al., 2000).

Based upon the definition of musculotendinous stiffness, $k = \Delta$ Force/ Δ Length, stiffness is related to and influences both the change in force and change in length of a muscle-tendon unit. For example, given a set change in length of a muscle-tendon unit, a less stiff unit will exhibit a lesser change in force compared to a stiffer muscle-tendon unit. Alternatively, for any given set change in force, a less stiff muscle-tendon unit will exhibit a greater change in length compared to a stiffer muscle-tendon unit. This suggests that lesser triceps surae stiffness should allow for greater ankle joint displacement during a jump landing. This idea is supported by previous research on landing techniques that has found that soft, or less stiff, landings are characterized by greater sagittal plane displacements at the primary joints of the lower extremity (Devita & Skelly, 1992; Self & Paine, 2001; Zhang et al., 2000). However, our results indicate that there is no relationship between triceps surae stiffness and knee flexion and valgus displacements, or peak posterior GRF. Additionally, supplementary correlation analyses suggest that there is no significant relationship between normalized triceps surae stiffness and hip flexion ($r = 0.80$, $p = 0.651$) or ankle DF ($r = 0.273$, $p = 0.118$) displacements. As a whole, our results appear to contradict the previous literature, but there are several factors that we have identified that may have contributed to the observed discrepancies.

First, there is an inherent difference in the concept of stiffness as described in our study and the stiffness described by Devita & Skelly (1992) and Zhang et al. (2000). Those authors defined a stiff versus a soft landing as a function of knee joint displacement during landing, and manipulated this displacement within subjects to evaluate the effects on both kinematics and kinetics during a jump landing. These studies are in sharp contrast to the current investigation that directly assessed linear stiffness of the triceps surae complex in a standardized position before measuring the biomechanics associated with the preferred landing strategy of subjects. As a result, the relationship between joint displacement and “stiffness” in the works of Zhang and Devita are more than likely the result of the fact that “stiffness” was defined as a function of displacement and therefore it is expected that these variables would be associated.

Second, the degree of neuromuscular activity utilized by a subject has the ability to alter the stiffness properties of a muscle, where increased neuromuscular activity results in increased musculotendinous stiffness (Wagner & Blickhan, 1999). In the current investigation, triceps surae stiffness was assessed under a standardized load relative to total body mass to facilitate similar amounts of neuromuscular activity during this assessment for all subjects as has been done previously. Blackburn et al. (2006) assessed triceps surae stiffness using a standardized load relative to neuromuscular activity, approximately 30% of maximal voluntary contraction, and reported similar activation levels. Unfortunately, EMG of the triceps surae was not measured during the jump landing task in this investigation, and it is unknown whether subjects used the same amount of neuromuscular activation during the jump landing task and the stiffness assessment. As a result, the triceps surae stiffness of a subject measured in the standardized assessment may not be truly representative of the stiffness strategy utilized by that subject during the jump landing task if different neuromuscular activation levels that would have altered the stiffness properties of the tissue were utilized in the two conditions. Ultimately, the capacity to which each individual utilized stiffness and recruited neuromuscular activity during the landing task is unclear. An individual who possesses greater stiffness may not necessarily have recruited a similar level of neuromuscular activity during landing as an individual who possesses lesser stiffness. Thus, the individual who possesses greater stiffness may not have landed with the lesser knee flexion displacement, greater knee valgus displacement and posterior GRF we would expect.

Finally, our definition of the loading phase of the jump landing task may have affected the relationship between sagittal plane displacements and triceps surae stiffness.

It is unclear how long the loading phase, defined as the time from initial ground contact to peak knee flexion, lasted for each subject. Similarly, it is not known the time that it took for each subject to reach peak vertical GRF during this loading phase. These two time elements may have dramatically affected our results due to the concept of impulse which is representative of the total amount of force necessary to change the motion of an object (Serway & Jewett, 2004). During landing, the ground reaction force acts on the body to bring the velocity to zero. The total amount of force that is necessary to achieve this is known as the impulse and is equal to the force applied to the body times the time period over which it is applied (force * Δtime). By definition the product of these two components is equivalent to the mass of the subject times the change in velocity of the subject. As subjects in our study completed a jump landing from a standardized height and distance, it is reasonable to assume that their velocity immediately before initial contact would be similar. Then, after normalizing for subject mass, it could be argued that all subjects would have needed a similar impulse during landing. However, the time over which the force was applied to achieve this impulse is unknown. Specifically, it is unclear how long the loading phase lasted and the time it took to reach peak vertical GRF in each subject. The lack of significant correlation between triceps surae stiffness and sagittal plane joint displacement suggests that individuals with high and low levels of stiffness demonstrated similar displacements. If individuals with greater stiffness demonstrate similar ankle DF displacement compared to those with lesser stiffness, landing over a shorter period of time would result in a greater peak force and present a positive correlation between stiffness and vertical GRF. This idea is supported by Devita & Skelley (1992) reported a shorter impact phase and greater vertical GRF in stiff landings compared to soft landings. Thus, stiffness may have an influence on loading phase duration, as opposed to knee flexion displacement. However, this notion is speculative, and future research is necessary to evaluate its plausibility.

The relationships between passive straight-leg ankle dorsiflexion ROM and knee flexion displacement, and normalized vertical and posterior GRFs are in agreement with our hypotheses. Individuals who displayed greater ankle DF ROM demonstrated lesser vertical and posterior GRF, and greater knee flexion displacement. Supplementary correlation analyses indicate that greater ROM is also related to greater hip flexion, but not ankle DF displacement. Thus, greater passive straight-knee DF ROM is generally associated with a less erect posture during landing. Increased joint displacement of the lower extremity likely increases the loading phase duration and enhances shock absorption (Devita & Skelly, 1992; Zhang et al., 2000). Previous literature also suggests that landing in a less erect posture encourages longer muscle moment arms in the lower extremity, thus an individual will not have to produce as much muscular force to attenuate the landing forces compared to landing in a more erect posture (Devita & Skelly, 1992). Additionally, the center of gravity (COG) of an individual landing in a less erect posture will demonstrate greater vertical displacement during the loading phase compared to a more erect posture (Blackburn & Padua, 2009). The potential influence of greater COG vertical displacement on lesser vertical GRF may be driven by greater loading phase duration and energy absorption, similar to our findings with hip and knee joint displacement.

However, it is difficult to further suggest whether greater knee and hip flexion displacements are driven by greater ankle DF ROM as no significant correlations exist between ankle DF displacement and ankle DF ROM. Our findings only suggest that greater ankle DF ROM influences lesser GRF and greater hip and knee flexion displacement.

The lack of correlation between ankle ROM and DF displacement may be explained by between-subject differences in landing style. Descriptive statistics in our study indicate a mean ankle initial contact angle of 55º (sd = 15º, range = 60º). The large variability for ankle angle at initial contact suggests that subjects may have adopted landing styles that range from a more dorsiflexed initial contact angle that does not allow for further ankle DF displacement to a more plantarflexed contact angle that allows for a lot of dorsiflexion. This variability in landing style likely contributed to the lack of a significant relationship between ROM and displacement.

Our findings further suggest that no influence exists between passive bent-leg ankle DF ROM and the biomechanical variables of interest. The lack of association between these variables and passive DF ROM in the bent-knee condition may potentially be explained by the mean knee flexion angle at initial contact during the jump landing and the contribution of the gastrocnemius at that relative joint angle. The mean knee flexion angle at initial contact was 11.10º ± 6.62º, while the peak value was 80.18º ± 13.31º. The straight-knee ROM measurement, performed in 0º of knee flexion, assesses the extensibility of both gastrocnemius and soleus while the bent-knee measurement, performed at 90º of knee flexion, primarily accounts for soleus extensibility only. The correlation between straight-knee ROM and the biomechanical variables of interest but not with bent-knee ROM suggests the importance of the gastrocnemius contributions to the jump landing task. Similarly, this idea may have contributed to the lack of relationship between triceps surae stiffness and the biomechanical variables of interest. Because triceps surae stiffness was assessed in the same knee position as the bent-knee ROM assessment, the stiffness contributions of the soleus may have been much greater than that of the gastrocnemius. A straight-knee position may more properly assess gastrocnemius stiffness and potentially be associated with the biomechanical variables of interest.

A lack of relationship between passive straight-knee DF ROM and knee valgus displacement was contrary to our initial hypothesis. Bell et al. (2008) previously reported that individuals who displayed excessive knee valgus (MKD) with a controlled squat task tended to demonstrate lesser ankle DF ROM. As such, we hypothesized that individuals who demonstrated lesser ankle DF ROM would display greater knee valgus during the jump landing task. However, a number of factors likely influenced the discrepancies between our results and those reported by Bell et al. First, Bell et al. reported no significant DF ROM differences between control and MKD groups in their study. The differences between groups reported by Bell et al. with the bent-knee condition approached statistical significance and are similar to our findings ($p = 0.053$). Both studies suggest a strong trend between ankle bent-knee DF ROM and knee valgus displacement. Overall, the lack of a significant difference reported by Bell et al. in ankle DF ROM between groups who displayed differences in knee valgus may suggest that ankle ROM has a limited influence on knee valgus. Second, Bell et al. also reported significantly less plantarflexion (PF) strength in the MKD group compared to the control group. Because existing literature suggests that the medial gastrocnemius acts as a dynamic stabilizer in preventing knee valgus (Lloyd & Buchanan, 2001), lesser plantarflexion strength may have a stronger influence on knee valgus displacement than ROM. Third, Bell et al. reported a mean of 8.5º for passive straight-knee ankle DF ROM in the MKD subjects while the mean for our study was 14.3º. The greater restriction reported by Bell et al. may be attributed to the placement of a bolster under the subject's heel when measuring straight-knee ankle DF ROM, forcing knee hyperextension and further restricting the gastrocnemius. The extent of the effect due to straight-knee positioning differences on ankle DF ROM measurements is unknown. Overall, the MKD group in the study by Bell et al. possessed approximately 6º (40%) less DF ROM, suggesting that DF ROM only influences frontal plane knee motion when severely restricted. There is the potential that had subjects exhibited greater restriction of the triceps surae complex in the current study, we may have observed an association with increased knee valgus during our task. Finally, the inclusion criteria for MKD used by Bell et al. may have only applied to a limited number of individuals in our study. Subjects were assigned to the MKD group if the MKD demonstrated during the squat task diminished when performing the same task on a decline wedge. Only 18 of 75 (24%) of the potential subjects met this criterion. This further suggests that only a small percentage of individuals demonstrated a strong MKD trend due to a poor ankle DF ROM. Lastly, differences in task characteristics may have influenced the potential association between ROM and knee valgus displacement. The influence of lesser ankle DF ROM may have a greater impact with controlled tasks such as a squat compared to the dynamic landing task used in our study.

Limitations & Further Research

The primary limitation to the current study involves the failure to monitor neuromuscular activity of the triceps surae via EMG during the jump landing. It is unclear how potential differences in neuromuscular activity, if any, altered the relationships between stiffness and landing biomechanics. Another limitation is that subjects were not given any specific instructions on how to land, leading to a large range of ankle DF displacement values and landing strategies.

Further research should incorporate a number of additions to the current protocol to observe the potential association between stiffness and ROM with the biomechanical variables of interest within this study. EMG data should be recorded during the jump landing to measure the amount of muscle activity present at the triceps surae. The data may be used to determine relationships in neuromuscular activity between groups who demonstrate higher and lower stiffness during the standardized assessment and the influence on biomechanics. A grouping criterion may be incorporated to compare biomechanical differences with individuals who display high and low stiffness or greater and lesser ROM during a jump landing. Ankle ROM may be artificially manipulated by adopting the research method performed by Hagins et al. (2007) in which a condition is added requiring individuals to land on an incline wedge and are monitored for specific biomechanical variables. Finally, further associations between triceps surae muscle stiffness, ankle ROM, knee flexion or valgus displacement, and vertical and posterior GRF associated with a jump landing may be identified by expanding the sample size to increase statistical power of the analysis.

Clinical Application

Greater knee flexion displacement, lesser knee valgus displacement, and lesser vertical and posterior GRF are biomechanical factors associated with a reduced risk of ACL injury. The current study suggests that lesser triceps surae muscle stiffness and greater passive straight-knee ankle DF ROM are associated with lesser vertical GRF during a jump landing. In addition, greater passive straight-knee ankle DF ROM is associated with greater knee flexion displacement and lesser posterior GRF. The implication of this study suggests clinicians should advocate techniques to increase passive straight-knee ankle DF ROM as this is associated with reduced ACL injury risk factors. The increase in ROM may lead to lesser vertical and posterior GRF and greater knee flexion displacements, thus reducing the likelihood of ACL injury.

Table 1: Subject Descriptive Statistics

Variable	Mean	SD
Age (years)	20.54	1.50
Height (cm)	177.01	10.46
Mass (kg)	73.42	14.11
Weight (N)	720.00	138.41

Table 2: Dependent Variable Descriptive Statistics

Dependent Variable	Mean	SD
Normalized Stiffness (N/cm/N)	0.128	0.019
Ankle Dorsiflexion Range of Motion		
– Passive at 0º	14.280	5.457
– Passive at 90º	18.931	5.900
Knee Flexion Displacement (deg)	69.084	12.004
Knee Valgus Displacement (deg)	7.044	5.009
Normalized Peak Vertical GRF (N/N)	2.156	0.569
Normalized Peak Posterior GRF (N/N)	0.589	0.197

Table 3: Pearson Correlation Analyses Between Normalized Stiffness and Landing Biomechanics

Dependent Variable	**r**	**r^2**	**p-value**
Knee Flexion Displacement	-0.123	0.015	0.489
Knee Valgus Displacement	0.316	0.100	0.068
Normalized Peak Vertical GRF	0.411	0.169	0.016*
Normalized Peak Posterior GRF	0.319	0.102	0.066

*significant correlation between variables

Table 4: Pearson Correlation Analyses Between Passive Straight-Knee Ankle DF ROM and Landing Biomechanics

Dependent Variable	r	r^2	p-value
Knee Flexion Displacement	0.464	0.215	0.029*
Knee Valgus Displacement	-0.290	0.084	0.091
Normalized Peak Vertical GRF	-0.411	0.169	0.014*
Normalized Peak Posterior GRF	-0.412	0.170	0.014*

*significant correlation between variables

Table 5: Pearson Correlation Analyses Between Passive Bent-Knee Ankle DF ROM and Landing Biomechanics

Dependent Variable	r	r^2	p-value
Knee Flexion Displacement	0.327	0.107	0.055
Knee Valgus Displacement	-0.330	0.109	0.053
Normalized Peak Vertical GRF	-0.311	0.097	0.069
Normalized Peak Posterior GRF	-0.295	0.087	0.085

*significant correlation between variables

Table 6: Supplemental Correlation Analyses Between Normalized Stiffness and Sagittal Joint Displacements

Dependent Variable	r	r^2	p-value
Ankle DF Displacement	-0.273	0.075	0.118
Knee Flexion Displacement	0.123	0.015	0.489
Hip Flexion Displacement	0.080	0.006	0.651

*significant correlation between variables

Table 7: Supplementary Correlation Analyses Between Passive Straight-Knee Ankle DF ROM and Sagittal Joint Displacements

Dependent Variable	r	r^2	p-value
Ankle DF Displacement	0.150	0.023	0.390
Knee Flexion Displacement	0.464	0.215	0.029*
Hip Flexion Displacement	0.357	0.127	0.035*

*significant correlation between variables

Figure 1: Tricep Surae Muscle Stiffness Assessment Setup

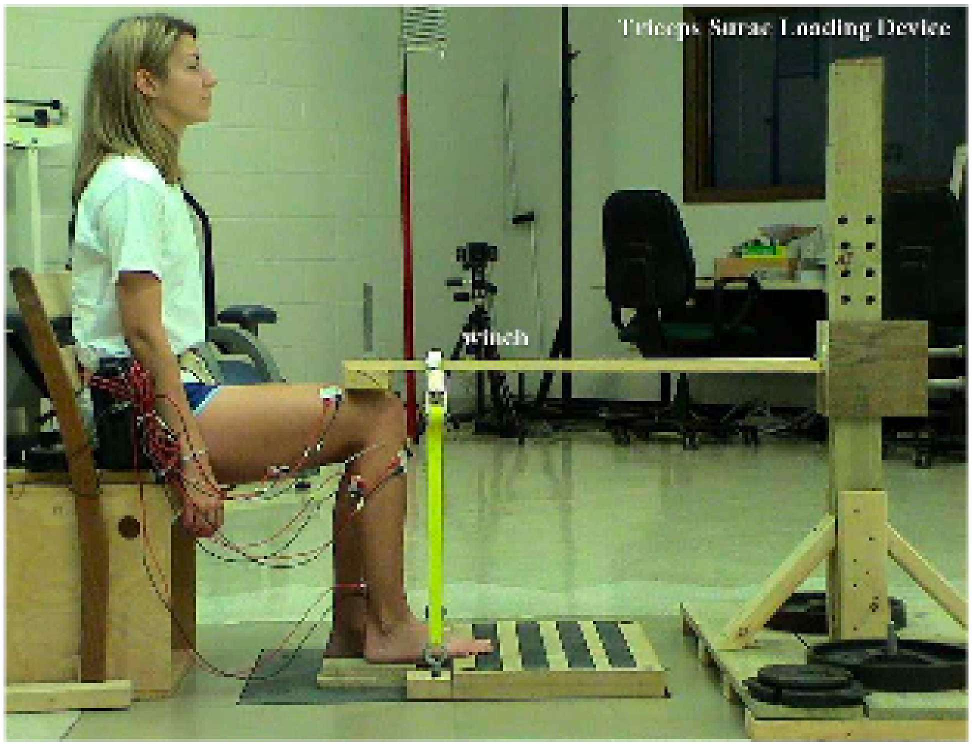


Figure 2: Damped Frequency Oscillation Method Example Output

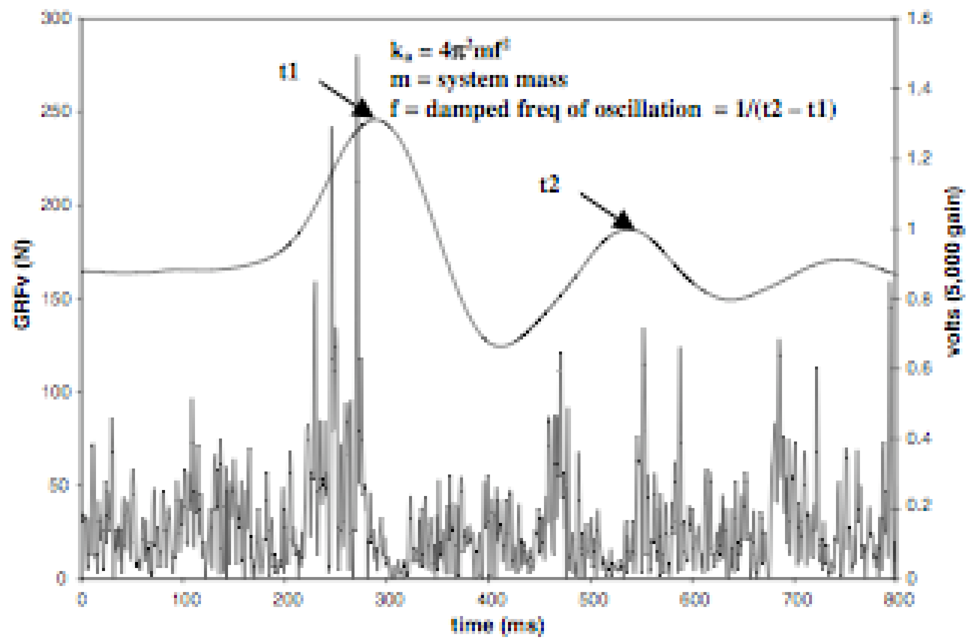


Figure 3: Scatter Plot and Trendline of Relationship Between Normalized Stiffness and Knee Flexion Displacement

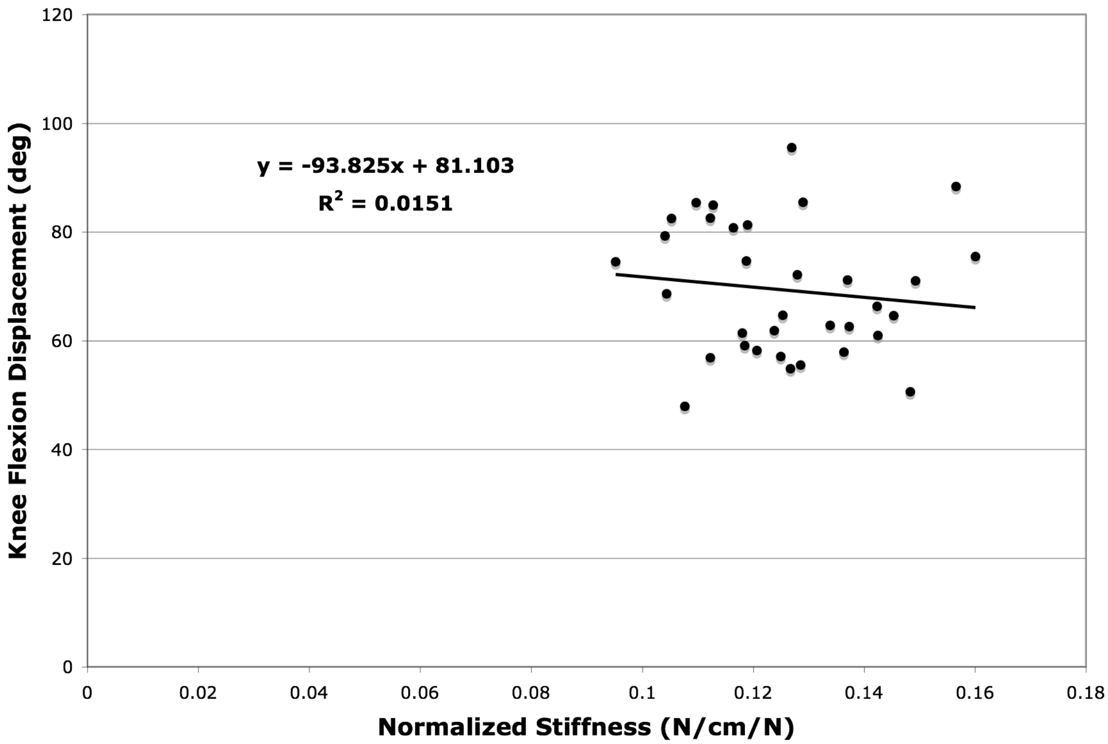
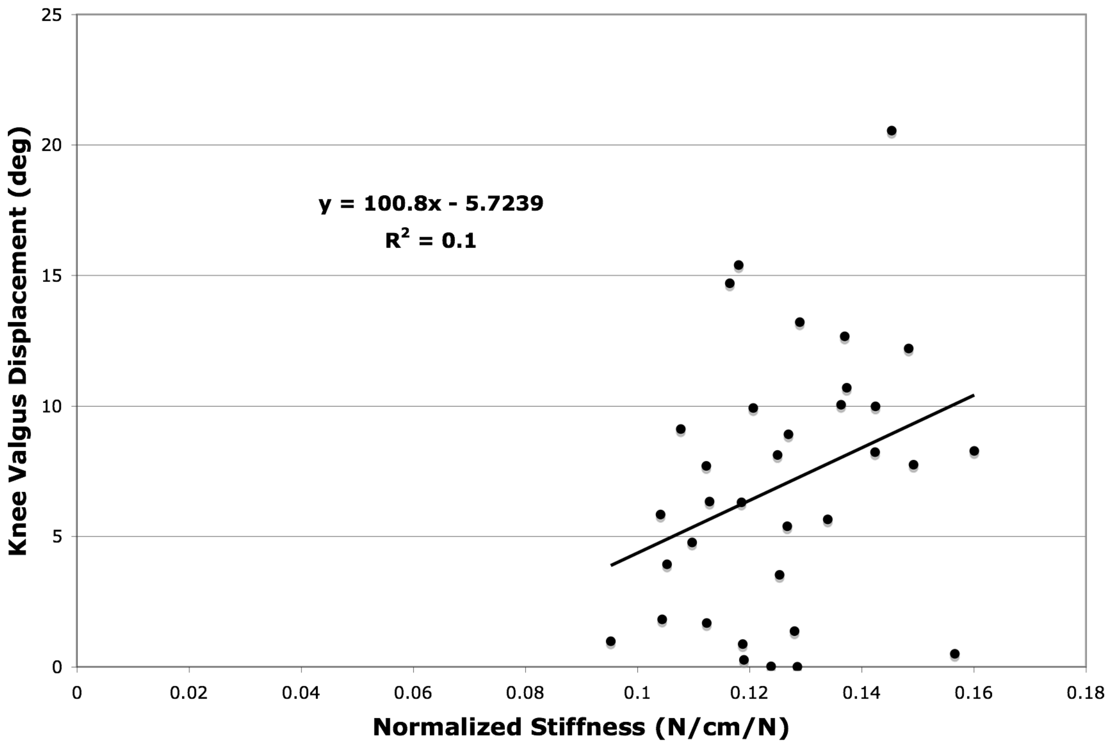


Figure 4: Scatter Plot and Trendline of Relationship Between Normalized Stiffness and Knee Valgus Displacement



(+)Y represents knee valgus displacement

Figure 5: Scatter Plot and Trendline of Relationship Between Normalized Stiffness and Normalized Vertical GRF

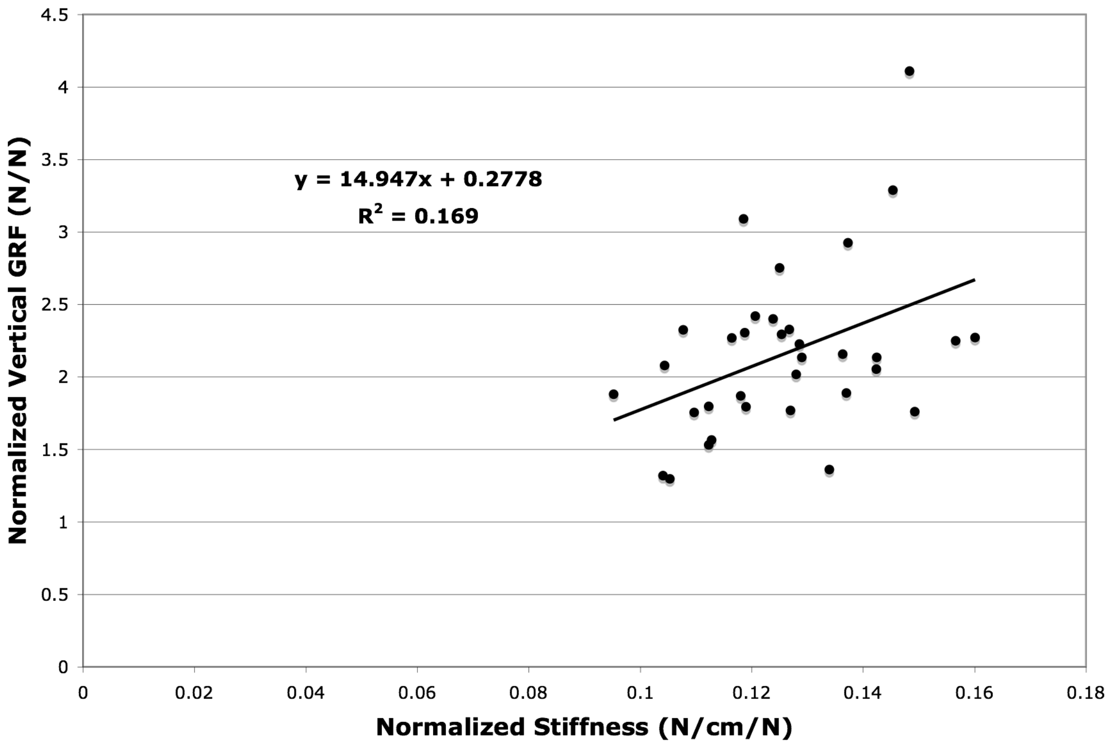


Figure 6: Scatter Plot and Trendline of Relationship Between Normalized Stiffness and Normalized Posterior GRF

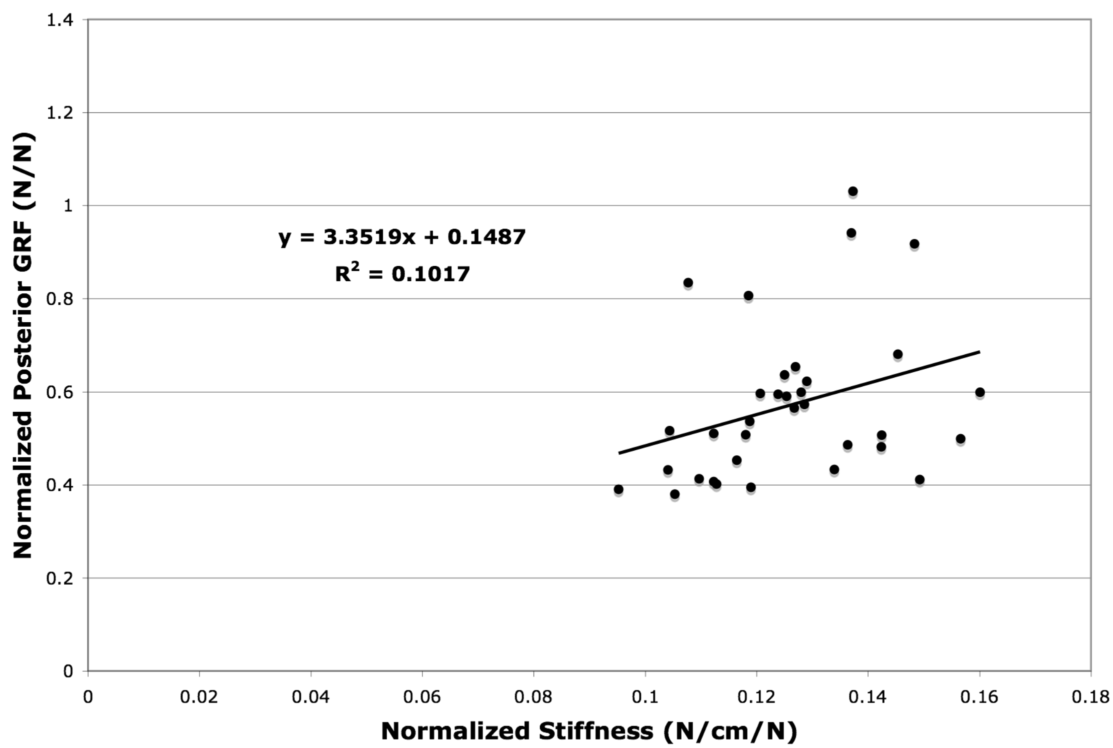


Figure 7: Scatter Plot and Trendline of Relationship Between Passive Straight-Knee Ankle DF ROM and Knee Flexion Displacement

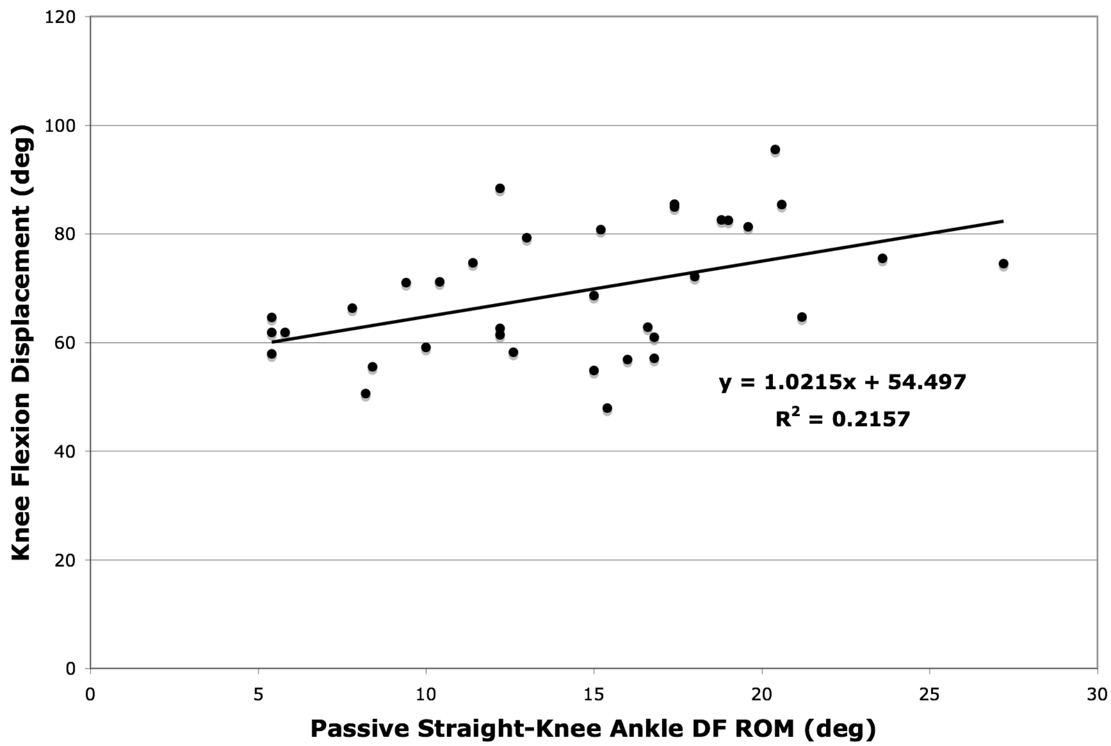
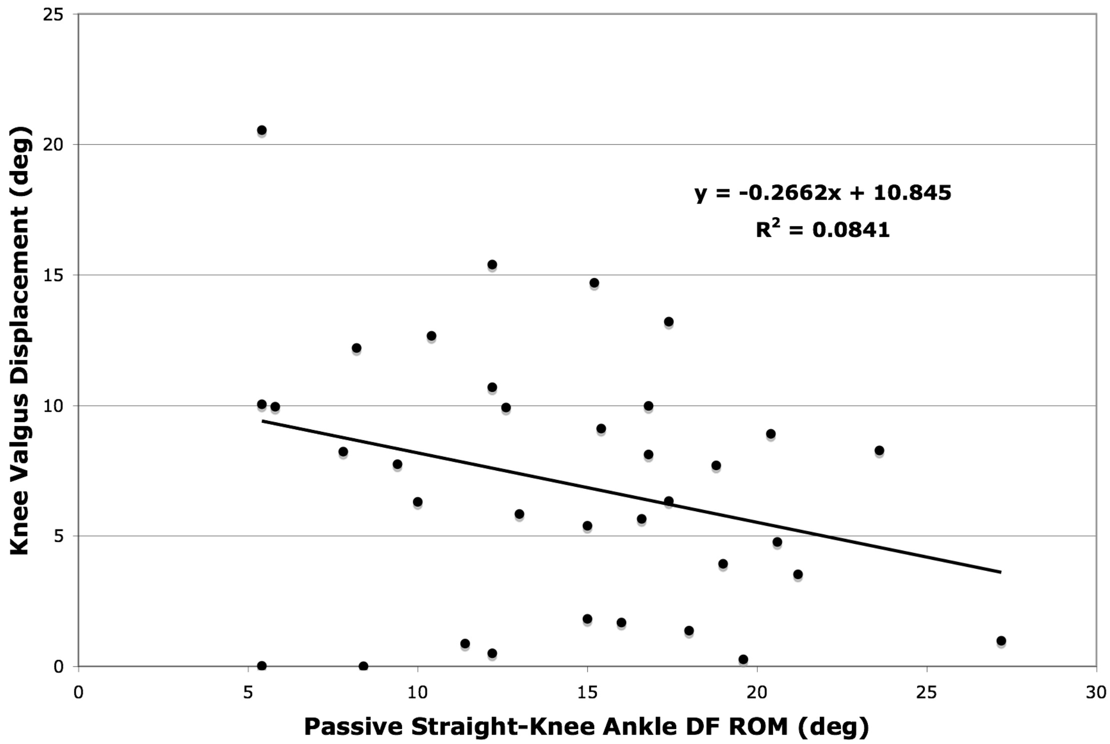


Figure 8: Scatter Plot and Trendline of Relationship Between Passive Straight-Knee Ankle DF ROM and Knee Valgus Displacement



(+)Y represents knee valgus displacement

Figure 9: Scatter Plot and Trendline of Relationship Between Passive Straight-Knee Ankle DF ROM and Normalized Vertical GRF

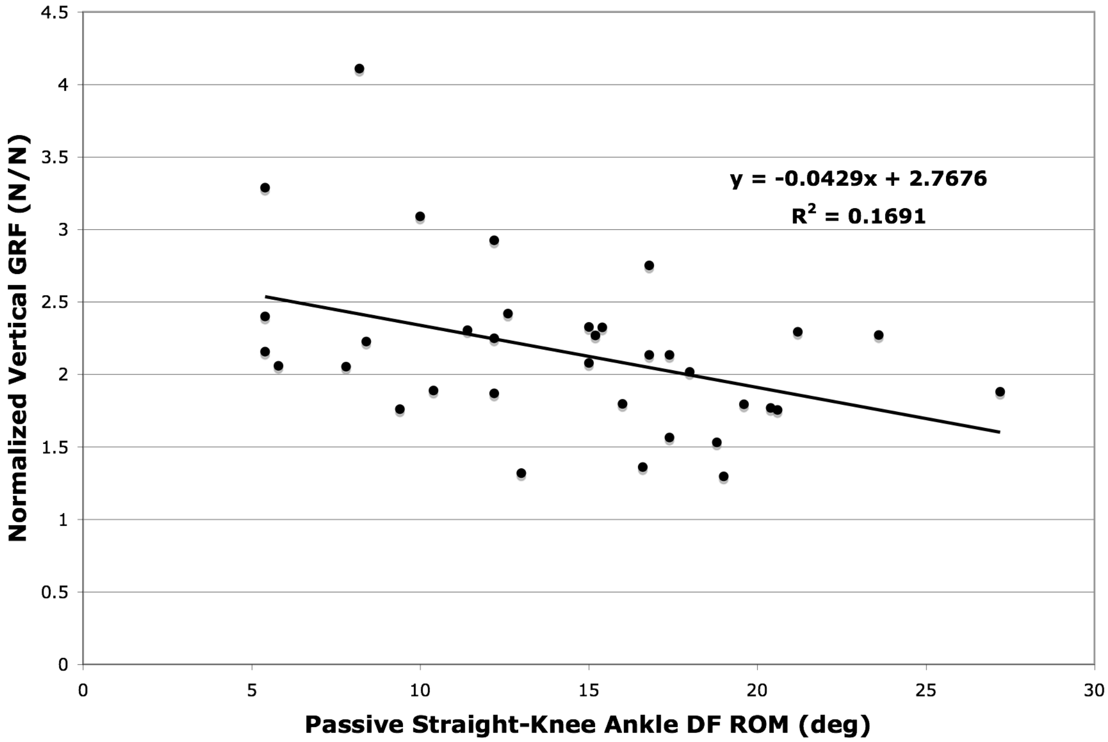


Figure 10: Scatter Plot and Trendline of Relationship Between Passive Straight-Knee Ankle DF ROM and Normalized Posterior GRF

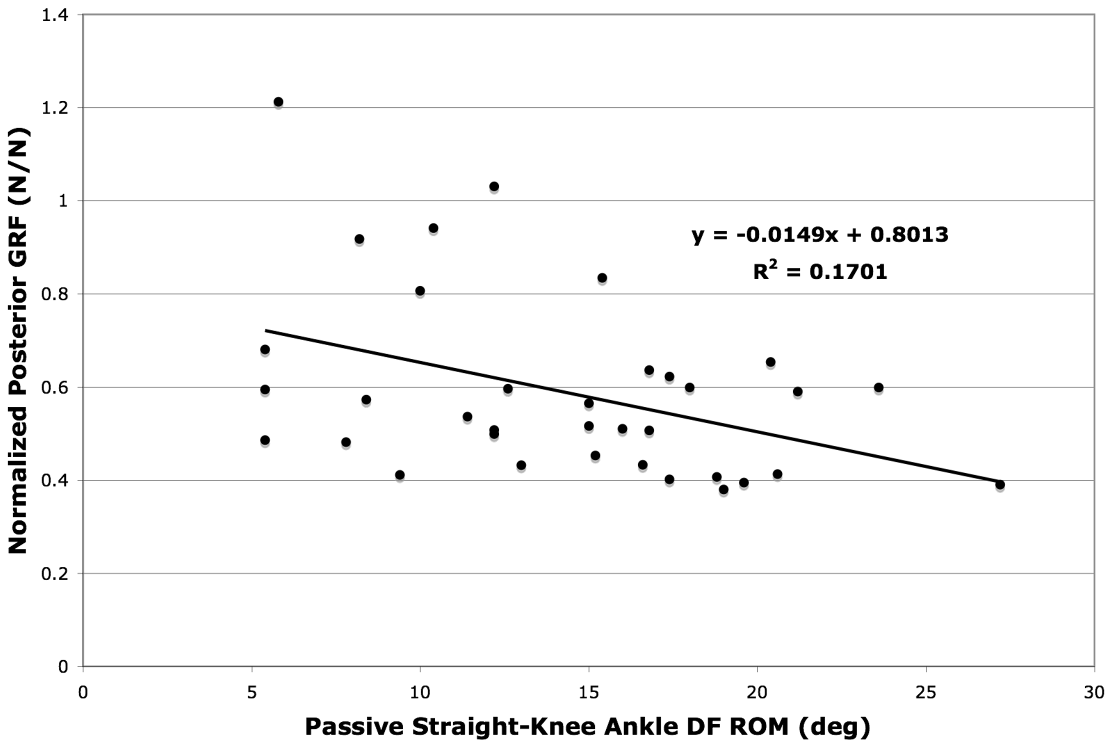


Figure 11: Scatter Plot and Trendline of Relationship Between Passive Bent-Knee Ankle DF ROM and Knee Flexion Displacement

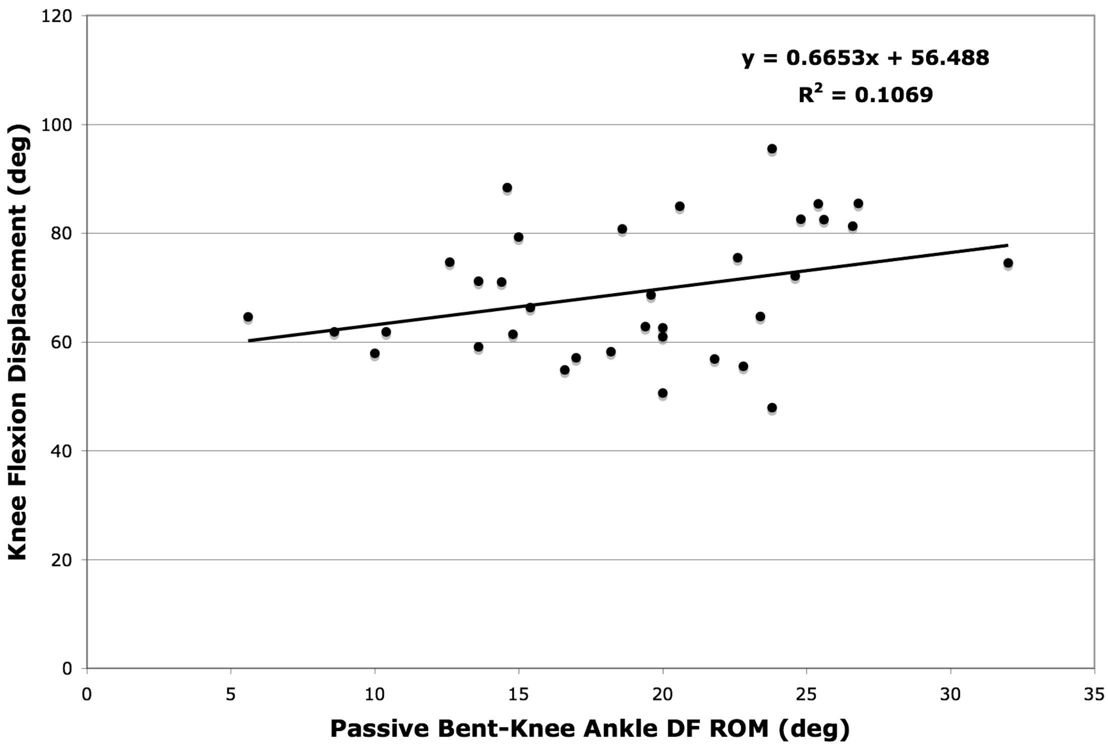
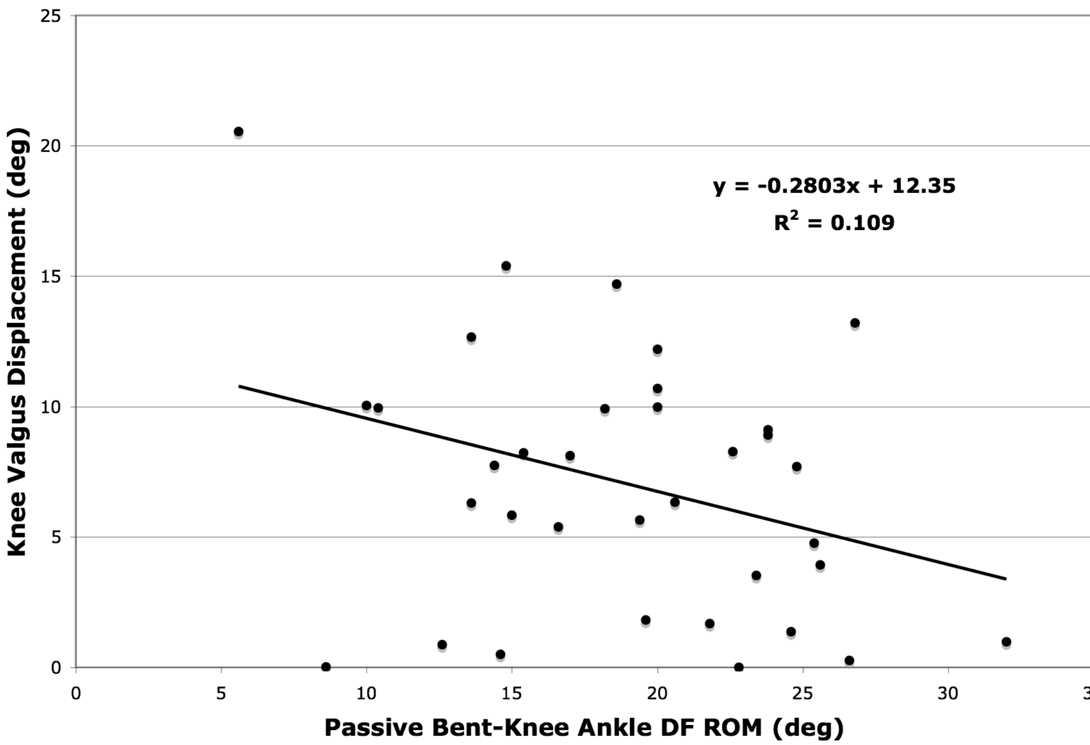


Figure 12: Scatter Plot and Trendline of Relationship Between Passive Bent-Knee Ankle DF ROM and Knee Valgus Displacement



(+)Y represents knee valgus displacement

Figure 13: Scatter Plot and Trendline of Relationship Between Passive Bent-Knee Ankle DF ROM and Normalized Vertical GRF

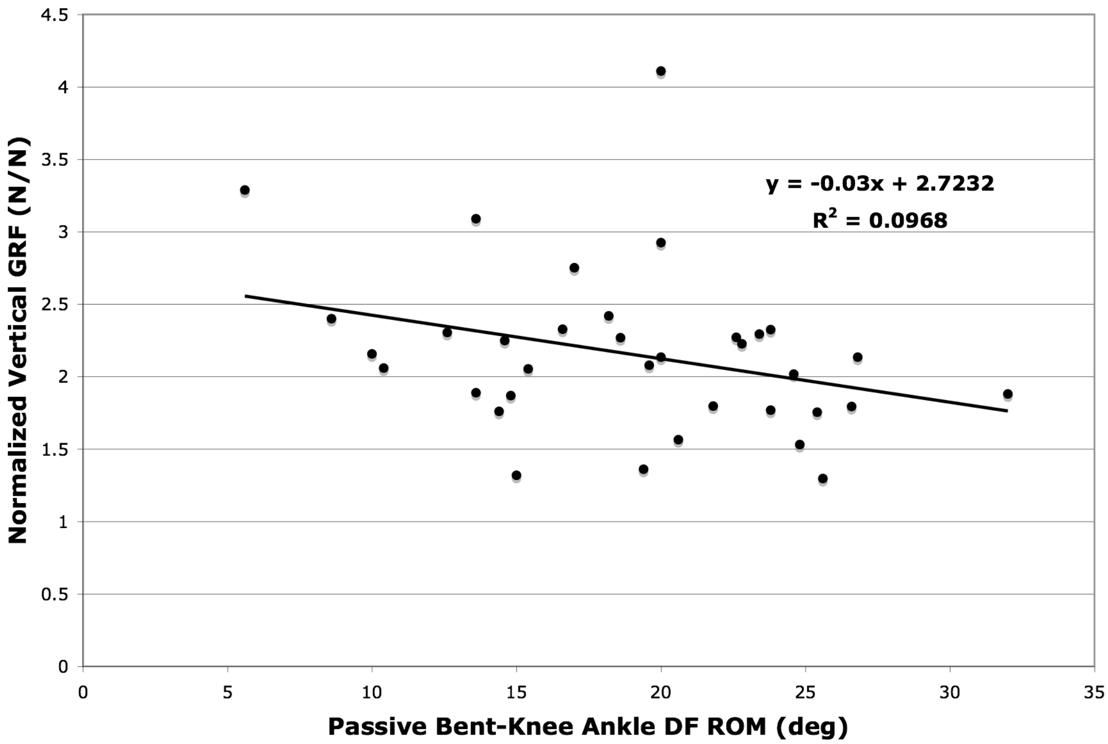
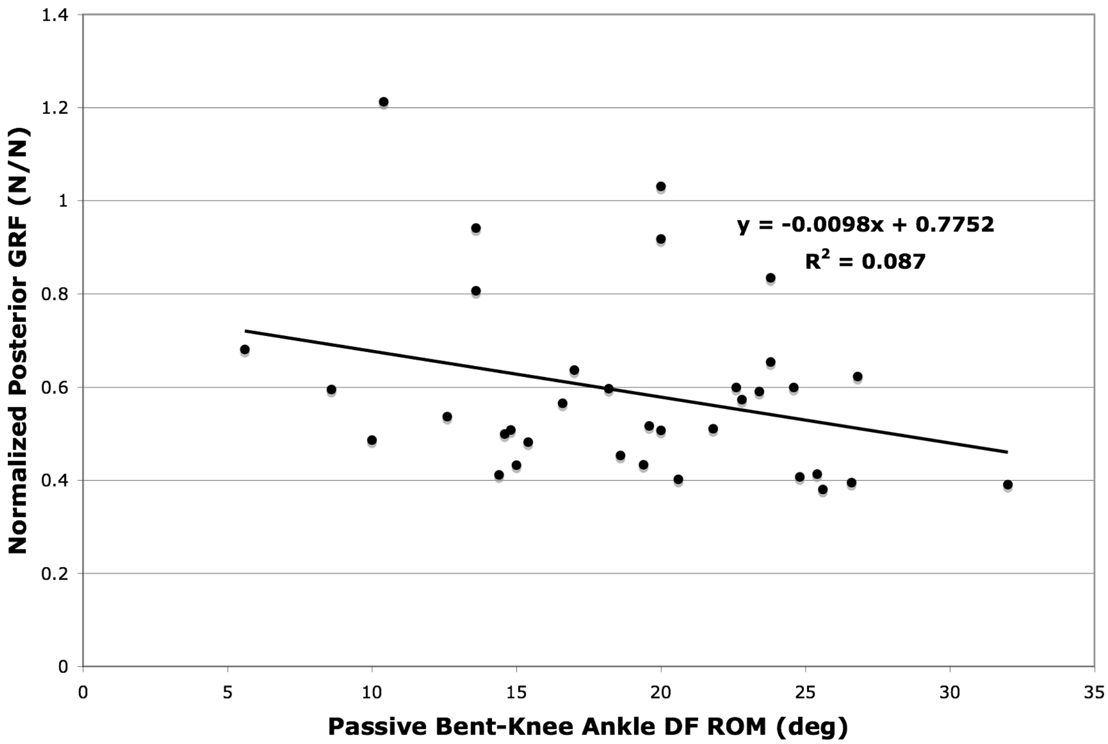


Figure 14: Scatter Plot and Trendline of Relationship Between Passive Bent-Knee Ankle DF ROM and Normalized Posterior GRF



REFERENCES

Agel, J., Arendt, E. A., & Bershadsky, B. (2005). Anterior cruciate ligament injury in national collegiate athletic association basketball and soccer: a 13-year review. *Am J Sports Med, 33*(4), 524-530.

Arendt, E. A., Agel, J., & Dick, R. (1999). Anterior Cruciate Ligament Injury Patterns Among Collegiate Men and Women. *J Athl Train, 34*(2), 86-92.

Bell, D. R., & Padua, D. A. (in press). Medial knee displacement characteristics muscle strength and flexibility characteristics of individuals displaying excessive medial knee displacement.

Bell, D. R., Padua, D. A., & Clark, M. A. (2008). Muscle strength and flexibility characteristics of people displaying excessive medial knee displacement. *Arch Phys Med Rehabil, 89*(7), 1323-1328.

Berns, G. S., Hull, M. L., & Patterson, H. A. (1992). Strain in the anteromedial bundle of the anterior cruciate ligament under combination loading. *J Orthop Res, 10*(2), 167-176.

Blackburn, J. T., & Padua, D. A. (2008). Influence of trunk flexion on hip and knee joint kinematics during a controlled drop landing. *Clin Biomech (Bristol, Avon), 23*(3), 313-319.

Blackburn, J. T., & Padua, D. A. (2009). Sagittal-plane trunk position, landing forces, and quadriceps electromyographic activity. *J Athl Train, 44*(2), 174-179.

Blackburn, J. T., Padua, D. A., Riemann, B. L., & Guskiewicz, K. M. (2004b). The relationships between active extensibility, and passive and active stiffness of the knee flexors. *J Electromyogr Kinesiol, 14*(6), 683-691.

Blackburn, J. T., Padua, D. A., Weinhold, P. S., & Guskiewicz, K. M. (2006). Comparison of triceps surae structural stiffness and material modulus across sex. *Clin Biomech (Bristol, Avon), 21*(2), 159-167.

Blackburn, J. T., Riemann, B. L., Padua, D. A., & Guskiewicz, K. M. (2004a). Sex comparison of extensibility, passive, and active stiffness of the knee flexors. *Clin Biomech (Bristol, Avon), 19*(1), 36-43.

Butler, D. L., Noyes, F. R., & Grood, E. S. (1980). Ligamentous restraints to anterior-posterior drawer in the human knee. A biomechanical study. *J Bone Joint Surg Am, 62*(2), 259-270.

Butler, R. J., Crowell, H. P., 3rd, & Davis, I. M. (2003). Lower extremity stiffness: implications for performance and injury. *Clin Biomech (Bristol, Avon), 18*(6), 511-517.

Chappell, J. D., Herman, D. C., Knight, B. S., Kirkendall, D. T., Garrett, W. E., & Yu, B. (2005). Effect of fatigue on knee kinetics and kinematics in stop-jump tasks. *Am J Sports Med, 33*(7), 1022-1029.

Chappell, J. D., Yu, B., Kirkendall, D. T., & Garrett, W. E. (2002). A comparison of knee kinetics between male and female recreational athletes in stop-jump tasks. *Am J Sports Med, 30*(2), 261-267.

Dempster, W. T., Gabel, W. C., & Felts, W. J. (1959). The anthropometry of the manual work space for the seated subject. *Am J Phys Anthropol, 17*, 289-317.

Devita, P., & Skelly, W. A. (1992). Effect of landing stiffness on joint kinetics and energetics in the lower extremity. *Med Sci Sports Exerc, 24*(1), 108-115.

Ford, K. R., Myer, G. D., Toms, H. E., & Hewett, T. E. (2005). Gender differences in the kinematics of unanticipated cutting in young athletes. *Med Sci Sports Exerc, 37*(1), 124-129.

Gleim, G. W., & McHugh, M. P. (1997). Flexibility and its effects on sports injury and performance. *Sports Med, 24*(5), 289-299.

Granata, K. P., Padua, D. A., & Wilson, S. E. (2002). Gender differences in active musculoskeletal stiffness. Part II. Quantification of leg stiffness during functional hopping tasks. *J Electromyogr Kinesiol, 12*(2), 127-135.

Hagins, M., Pappas, E., Kremenic, I., Orishimo, K. F., & Rundle, A. (2007). The effect of an inclined landing surface on biomechanical variables during a jumping task. *Clin Biomech (Bristol, Avon), 22*(9), 1030-1036.

Hewett, T. E., Myer, G. D., Ford, K. R., Heidt, R. S., Jr., Colosimo, A. J., McLean, S. G., et al. (2005). Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes: a prospective study. *Am J Sports Med, 33*(4), 492-501.

Hirth, C. J. (2004). Clinical movement analysis to identify muscle imbalances and guide exercise. *Athletic Therapy Today, 12*(4), 10-14.

Jennings, A. G., & Seedhom, B. B. (1998). The measurement of muscle stiffness in anterior cruciate injuries -- an experiment revisited. *Clin Biomech (Bristol, Avon), 13*(2), 138-140.

Kubo, K., Kanehisa, H., & Fukunaga, T. (2003). Gender differences in the viscoelastic properties of tendon structures. *Eur J Appl Physiol, 88*(6), 520-526.

Lloyd, D. G., & Buchanan, T. S. (2001). Strategies of muscular support of varus and valgus isometric loads at the human knee. *J Biomech, 34*(10), 1257-1267.

Majewski, M., Susanne, H., & Klaus, S. (2006). Epidemiology of athletic knee injuries: A 10-year study. *Knee, 13*(3), 184-188.

Markolf, K. L., Burchfield, D. M., Shapiro, M. M., Shepard, M. F., Finerman, G. A., & Slauterbeck, J. L. (1995). Combined knee loading states that generate high anterior cruciate ligament forces. *J Orthop Res, 13*(6), 930-935.

McNair, P. J., Wood, G. A., & Marshall, R. N. (1992). Stiffness of the hamstring muscles and its relationship to function in anterior cruciate ligament deficient individuals. *Clinical Biomechanics, 7*(3), 131-137.

Morgan, D. L. (1977). Separation of active and passive components of short-range stiffness of muscle. *Am J Physiol, 232*(1), C45-49.

Mountcastle, S. B., Posner, M., Kragh, J. F., Jr., & Taylor, D. C. (2007). Gender differences in anterior cruciate ligament injury vary with activity: epidemiology of anterior cruciate ligament injuries in a young, athletic population. *Am J Sports Med, 35*(10), 1635-1642.

Noyes, F. R., Matthews, D. S., Mooar, P. A., & Grood, E. S. (1983). The symptomatic anterior cruciate-deficient knee. Part II: the results of rehabilitation, activity modification, and counseling on functional disability. *J Bone Joint Surg Am, 65*(2), 163-174.

Noyes, F. R., Mooar, L. A., Moorman, C. T., 3rd, & McGinniss, G. H. (1989). Partial tears of the anterior cruciate ligament. Progression to complete ligament deficiency. *J Bone Joint Surg Br, 71*(5), 825-833.

Padua, D. A. (2003). Muscle stiffness and biomechanical stability. *Athletic Therapy Today, 8*(6), 62-64.

Prodromos, C. C., Han, Y., Rogowski, J., Joyce, B., & Shi, K. (2007). A meta-analysis of the incidence of anterior cruciate ligament tears as a function of gender, sport, and a knee injury-reduction regimen. *Arthroscopy, 23*(12), 1320-1325 e1326.

Schmitz, R. J., Kulas, A. S., Perrin, D. H., Riemann, B. L., & Shultz, S. J. (2007). Sex differences in lower extremity biomechanics during single leg landings. *Clin Biomech (Bristol, Avon), 22*(6), 681-688.

Self, B. P., & Paine, D. (2001). Ankle biomechanics during four landing techniques. *Med Sci Sports Exerc, 33*(8), 1338-1344.

Sell, T. C., Ferris, C. M., Abt, J. P., Tsai, Y. S., Myers, J. B., Fu, F. H., et al. (2007). Predictors of proximal tibia anterior shear force during a vertical stop-jump. *J Orthop Res, 25*(12), 1589-1597.

Serway, R. A., & Jewett, J. W. (2004). Physics for Scientists and Engineers. (6th ed.), Brooks/Cole.

Smith, B. A., Livesay, G. A., & Woo, S. L. (1993). Biology and biomechanics of the anterior cruciate ligament. *Clin Sports Med, 12*(4), 637-670.

Vesci, B. J., Padua, D. A., Bell, D. R., Strickland, L. J., Guskiewicz, K. M., & Hirth, C. J. (2007). Influence of hip muscle strength, flexibility of hip and ankle musculature, and hip muscle activation on dynamic knee valgus motion during a double-legged squat. *Journal of Athletic Training, 42*(2), S-83.

Wagner, H., & Blickhan, R. (1999). Stabilizing function of skeletal muscles: an analytical investigation. *J Theor Biol, 199*(2), 163-179.

Yu, B., Lin, C. F., & Garrett, W. E. (2006). Lower extremity biomechanics during the landing of a stop-jump task. *Clin Biomech (Bristol, Avon), 21*(3), 297-305.

Zhang, S. N., Bates, B. T., & Dufek, J. S. (2000). Contributions of lower extremity joints to energy dissipation during landings. *Med Sci Sports Exerc, 32*(4), 812-819.

Zheng, N., Fleisig, G. S., Escamilla, R. F., & Barrentine, S. W. (1998). An analytical model of the knee for estimation of internal forces during exercise. *J Biomech, 31*(10), 963-967.