THE EFFECTS OF ANKLE BRACING ON LOWER EXTREMITY KINEMATICS AND KINETICS DURING DEEP SQUATS AND JUMP LANDINGS IN FEMALES

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To the Dean of the Graduate School:

I am submitting herewith a dissertation written by Gary Alan Christopher entitled "The Effects of Ankle Bracing on Lower Extremity Kinematics and Kinetics during Deep Squats and Jump Landings in Females." I have examined this dissertation for form and content and recommend that it be accepted in partial fulfillment of the requirements for the degree of Doctor of Philosophy with a major in Kinesiology.

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ABSTRACT

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THE EFFECTS OF ANKLE BRACING ON LOWER EXTREMITY KINEMATICS AND KINETICS DURING DEEP SQUATS AND JUMP LANDINGS IN FEMALES

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Females are at greater risk of sustaining injury to the anterior cruciate ligament (ACL) than males. Numerous risk factors have been identified, including ankle taping. The purpose of this research was to assess the effects of ankle taping and bracing on knee and ankle kinematics during deep squats and knee kinematics and kinetics during reactive drop landings. Eight healthy athletic female participants performed motion trials under three bilateral bracing conditions: untaped, closed basketweave ankle taping, and Webly® ankle orthoses. Participants performed deep squats to determine the effects of bracing on ankle and knee range of motion. Participants performed drop-jumps from their maximum jump height; upon landing, participants moved to the right, left, or stood up. Video and ground reaction force data were collected simultaneously at 60 and 240 Hz, respectively. Segment kinematics were derived from video data; segment kinetics were computed using inverse dynamics. Deep squat dependent variables were ankle and knee range of motion and were analyzed using single-factor (bracing condition) within-subjects repeated measures MANOVA. Drop-jump kinematic variables were knee flexion,

valgus, and rotation angles at foot strike and maximum; kinetic variables were knee extension, varus, and rotation moments at foot strike and maximum and maximum ground reaction force. Drop-jump data were analyzed using two-factor (3×3) (bracing condition and side: upright, contralateral, and ipsilateral) within-subjects repeated measures MANOVA; α was set at .05. Deep squat data indicated significant differences among bracing conditions ($F_{(4,28)} = 3.500$, $p = .019$), with the taped condition providing the greatest restriction of motion at the knee and ankle. Drop-jump data indicated no multivariate kinematic ($F_{(14,18)} = 1.181$, $p = .364$) or kinetic differences ($F_{(14,18)} = .867$, $p = .601$) among bracing conditions. There were, however, multivariate kinematic ($F_{(14,18)} = 3.540$, $p = .007$) and kinetic differences ($F_{(14,18)} = 5.271$, $p = .001$) among side conditions. While ankle bracing and taping affected the range of motion at the knee, it does not appear that bracing and taping affected the kinematics or kinetics of the knee during reactive maneuvers known to be associated with ACL injuries.

Key Words: ankle taping/bracing, ACL injury risk, kinematics, kinetics

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CHAPTER I

INTRODUCTION

Since the enactment of *Title IX of the Education Amendments of 1972 to the Civil Rights Act of 1964* (Title IX), women's participation in organized athletic programs has increased dramatically. In 1982, the first year participation statistics were recorded, women accounted for 27.8% of participants in National Collegiate Athletic Association (NCAA) sponsored sports. By 2001, women's participation was 41.9%, an increase of 231%. During that same period, men's participation increased 124% (NCAA, 2002). Concurrent with the increase in female participation in athletics is an increased incidence of injuries in general and, in particular, injuries to the anterior cruciate ligament (ACL).

Female athletes have a greater risk of sustaining an ACL injury than male athletes; various studies place the risk from 2 to 8 times greater than in males (Arendt, 2001; Arendt & Dick, 1995; Ireland, 2002; Ireland, Gaudette, & Crook, 1997; Medvecky, Bosco, & Sherman, 2000). A six-year study at the United States Naval Academy determined the relative risk of ACL injury in female midshipmen from 2.01 to 10.85 times that of male midshipmen, depending upon the sport or military exercise (Gwinn, Wilckens, McDevitt, Ross, & Kao, 2000). A four-year study at the United States Military Academy revealed female cadets with one or more risk factors for ACL injury had an activity dependent relative risk ranging from about 1.0 (equal to that of males) to 37.7 (Uhorchak et al., 2003).

Uhorchak et al. observed injury rates of 2.1% in males and 6.6% in females, a female-to-male ratio of approximately 3:1.

Most ACL injuries, in both men and women, are non-contact in nature (Agel, Arendt, & Bershadsky, 2005; Arendt & Agel, 1999; Arendt & Dick, 1995; Ireland, 1999; Lloyd, 2001). Non-contact ACL injuries typically occur when landing from a jump, stopping, or changing direction quickly. A combination of hip adduction, hip internal rotation, and knee valgus places high stress on the ACL and can contribute to injury.

Risk factors for sustaining a non-contact ACL injury are generally broken down into extrinsic and intrinsic factors. Intrinsic factors are particular to an individual and may or may not be modifiable. Non-modifiable risk factors include menstrual cycle/hormonal status, intercondylar notch width, and ACL size relative to body weight and height. Modifiable intrinsic risk factors include strength and neuromuscular control of movements. Extrinsic factors can be modified through changes in environment or conditioning; these include playing surface, shoe type, and protective equipment (Arendt, 2001; Griffin et al., 2000; Harmon & Ireland, 2000; Ireland, 1999). Some researchers have suggested ankle bracing or taping as a possible extrinsic risk factor for ACL injury (Ferguson, 1973; Santos, McIntire, Foecking, & Liu, 2004), because a local constraint of motion in the kinetic chain can have remote effects.

The limbs of the human body form a kinetic chain – a link-system that transmits forces to and from the external environment. The links of the system are the limbs and joints of the body. Ground reaction forces are transmitted up the kinetic chain, while muscle forces are transmitted up and down the kinetic chain. This interconnectedness can

lead to cascading effects, with motion in one segment affecting motion in adjacent segments (Dugan, 2005; Leetun, Ireland, Willson, Ballantyne, & Davis, 2004). For example, subtalar pronation during weight bearing leads to calcaneal eversion, talar internal rotation and plantar flexion, and tibial internal rotation (Starkey & Ryan, 1996). Any restriction of this natural pattern may result in the abnormal transmission of forces to the knee. This provides some rationale that ankle bracing may affect the knee and play a role as an ACL injury risk factor.

Purpose of the Study

The purpose of this study was to investigate the effects of ankle bracing on (a) the sagittal plane kinematics of the ankle and knee during deep squats and (b) the kinematics and kinetics of the knee during jump landings. The following ankle-bracing conditions were investigated: untaped, basketweave athletic taping, and Webly® Ankle Orthosis (Hely & Weber, Inc., Santa Paula, CA.).

Research Questions

1. Does ankle bracing or taping affect sagittal plane kinematics (maximum knee flexion and ankle dorsiflexion at maximum knee flexion) during a deep squat (i.e., does ankle bracing limit motion at the knee and ankle)?
2. Does ankle bracing or taping affect the kinematics of the knee (valgus, rotation, & flexion angles, and anterior tibial translation) during jump landings at touchdown and at maximum?

3. Does ankle bracing or taping affect the kinetics of the knee (sagittal, frontal, and transverse plane moments, and ground reaction force) during jump landings at touchdown and at maximum?

Null Hypotheses

$H_0 1$: There will be no differences in sagittal plane kinematics of the knee or ankle during a deep squat among bracing conditions.

$H_0 2$: There will be no differences in knee kinematics during jump landings among bracing conditions.

$H_0 3$: There will be no differences in knee kinetics during jump landings among bracing conditions.

Definitions

Anterior cruciate ligament (ACL) – major ligament of the knee that serves to limit anterior translation of the tibia with respect to the femur. Additionally, the ACL is crucial to the rotary stability of the knee, controlling external rotation of the femur on the tibia (internal tibial torsion) in a weight-bearing situation.

Arthrokinematics – description of the small magnitude motions that occur at bone surfaces during movement; descriptors are roll, glide, and spin.

Body segment parameters (BSP) – body segment masses, locations of centers of mass, and moments of inertia. Sources of BSP include cadaver studies, mathematical models, and gamma scanning of living subjects.

Center of gravity – imaginary point about which a body or body segment is balanced; often used interchangeably with center of mass.

Center of mass – imaginary point at which all mass of a body or body segment is considered to be located; location at which gravity is assumed to act.

Center of pressure – point of application of ground reaction force.

Closed kinetic chain (closed chain) – situation in which the distal segment is fixed, resulting in movement of the proximal segment with respect to the distal segment.

Countermovement jump – jump technique employed to maximize jump height or distance; characterized by a preparatory squat featuring hip and knee flexion and ankle dorsiflexion, followed immediately by coordinated rapid hip and knee extension and ankle plantarflexion (Feltner, Fraschetti, & Crisp, 1999).

Drop jump – jump technique in which the participant steps off an elevated platform (without jumping up or stepping down) and lands on either one (same or opposite) or both feet. Following the landing, the participant may perform a variety of tasks, including "sticking" the landing, maximum vertical jump, maximum lateral jump, side-step, or pivot-and-run.

Glide – arthrokinematic descriptor of motion, in which a single point of a moving surface contacts multiple points on a relatively stationary surface.

Ground reaction force – the force exerted by the ground in response (reaction) to a force applied to the ground. Ground reaction forces were measured using dual strain gauge force platforms (Advanced Mechanical Technology, Inc., Watertown, MA).

Ground reaction moment – the moment exerted by the ground in response to an applied moment; measured with the same equipment as ground reaction force.

Inertia – property of an object that reflects its resistance to linear acceleration.

Inverse dynamics – indirect method of determining joint resultant forces and moments based on body segment kinematics, body segment parameters, and ground reaction forces and moments.

Kinematics – area of study that examines the description of motion, including body and joint positions, angles, velocities, and accelerations.

Kinetics – area of study that examines the causes of motion, including joint resultant forces and moments.

Kinetic chain – a link-system that transmits forces and moments to and from the environment. The links of the system are the limbs and joints of the body.

Moment – rotational effect of a force applied at a distance from an axis of rotation; also known as torque.

Moment of inertia – property of an object that reflects its resistance to angular acceleration.

Open kinetic chain (open chain) – situation in which the distal segment is free to move, resulting in movement of the distal segment with respect to the proximal segment.

Osteokinematics – description of the large magnitude motions that occur in adjacent segments (bones) during movement; descriptors include flexion, extension, abduction, adduction, rotation, etc.

Posterior cruciate ligament (PCL) – major ligament of the knee that serves to limit posterior translation of the tibia with respect to the femur.

Roll – arthrokinematic descriptor of motion, in which multiple points of a moving surface contact multiple points on a relatively stationary surface.

Spin – arthrokinematic descriptor of motion, in which a single point of a moving surface contacts a single point on a relatively stationary surface.

Stop jump – jump technique in which the participant runs forward, pushes off the ground with one foot and lands on both feet. Following the landing, the participant may perform a variety of tasks, including "sticking" the landing, maximum vertical jump, maximum lateral jump, maximum backward jump, side-step, or pivot-and-run.

Strain – deformation of a material under load; computed as the change of length divided by the original length of the material ($\sigma = \Delta L/L_0$).

Stress – applied load on a material; computed as the applied force divided by the cross sectional area of the material ($s = F/a$).

Stride jump – jump technique in which the participant pushes off the ground with one foot and lands on one foot (same or opposite as takeoff foot). Following the landing, the participant may perform a variety of tasks, including "sticking" the landing, maximum vertical jump, maximum lateral jump, side-step, or pivot-and-run.

Assumptions

1. The body is modeled as a rigid link system with frictionless pin joints connecting body segments.
2. Body segment parameters can be satisfactorily estimated using adjustments to Zatsiorsky-Seluyanov's segment inertia parameters (de Leva, 1996).

Delimitations

1. Participants were females recruited from the university community.
2. Participants had body mass index (BMI) ≤ 27.

Limitations

1. Knee joint and ankle joint centers were defined as the midpoint of the line connecting the medial & lateral femoral epicondyles and medial & lateral malleoli, respectively. While the actual joint centers change with motion, the assumption of a fixed location is typical in biomechanics and is a recognized limitation of the inverse dynamics process.
2. The hip joint center was defined using the so-called Andriacchi-Tylkowski hybrid method (Bell, Pedersen, & Brand, 1990). While precise location of the actual hip joint center is difficult, this method provides acceptable estimates. Additionally, the inverse dynamics process will focus on the knee, and more accurate means of hip joint center location are not necessary, as the hip joint is not included in the inverse dynamics equations for the knee.
3. Body segment parameters (BSP) were drawn from de Leva (1996), with no further individualization. Actual segment properties of participants may not conform exactly to those reported, nor are segments perfectly rigid. However, the BSP provided in de Leva are based on data from females of approximately the same age, mass, height, and body composition of the target participant group. In addition, segment rigidity assumptions are used extensively in biomechanical research and are considered acceptable estimates of properties of the human body.
4. Participants wore their own shoes. Shoe type, age, orthotic devices, and individual wear patterns can affect the transmission of ground reaction forces to the body during jump landings (Tillman et al., 2003).

Significance of Study

This study is expected to be applicable to athletes of all ages and skill levels, as all are susceptible (in varying degrees) to knee injuries, and should be concerned with the effects of ankle bracing or taping on the potential for such injuries. There is some debate as to the effect of ankle bracing on the potential for knee injury (Bieze, 2002). The research done to date is inconclusive, and these questions remain unanswered.

CHAPTER II

REVIEW OF LITERATURE

Introduction

Before 1965, no mention was found in the medical literature of anterior cruciate ligament injuries. Prior to that time, it is assumed that ACL injuries were classified under the broad category of knee sprains. A computerized literature search was conducted, using the EBSCO Host databases CINAHL, Pre-CINAHL, MEDLINE, and SportDiscus, the keywords "anterior cruciate ligament" and "knee injury" and limited to articles published in 1970 and before. This search yielded 28 unique papers, the earliest of which were published in 1965. Since 1970, there has been a literal explosion in the publication of ACL injury related research. The same keyword search for the period 1971 to present yielded over 10,000 publications. This review of literature will focus on the mechanisms and risk factors for ACL injury, and the role of bracing as a potential risk factor.

Mechanisms of ACL injury

The mechanism for a contact ACL injury is a combination of a planted foot, internal or external tibial torsion, extended knee, and the application of an external valgus force at the knee (Arnheim & Prentice, 1997; Starkey & Ryan, 1996). In the vast majority of cases, however, there is no external contact force (Harmon & Ireland, 2000; Ireland et al., 1997; Kirkendall & Garrett, 2000; Olsen, Myklebust, Engebretsen, & Bahr, 2004), even in contact sports (Lloyd, 2001). In most situations, the ACL injury occurs when

suddenly stopping or changing directions (Besier, Lloyd, Cochrane, & Ackland, 2001; Ford, Myer, Toms, & Hewett, 2005), or landing from a jump (Chappell, Yu, Kirkendall, & Garrett, 2002; Fagenbaum & Darling, 2003; Ford, Myer, & Hewett, 2003; Paul et al., 2003). The mechanism of the "classic" non-contact ACL injury places the upper body forward of the center of mass, the lower body in femoral internal rotation and adduction, tibial external rotation and subtalar pronation. Upon landing in this "point of no return" position (Ireland et al., 1997), ground reaction forces impart valgus and extension moments that place excessive stress on the ACL and can result in its failure (Lephart, Ferris, & Fu, 2002).

Risk Factors for ACL Injury

That females are at greater risk for ACL injuries than men is undisputed. Epidemiological studies have identified numerous risk factors for ACL injury. A typical system classifies these risk factors as intrinsic or extrinsic. Intrinsic risk factors refer to qualities that are internal to an individual, such as anatomy or physiology. Extrinsic risk factors concern possible external causes of injury. The risk of actually sustaining an ACL injury is a complex interplay between intrinsic risk factors, extrinsic risk factors, and precipitating conditions (Bahr & Krosshaug, 2005).

Intrinsic risk factors may or may not be modifiable. Fixed factors include joint laxity, lower extremity alignment, knee arthrokinematics, femoral notch width, relative ACL size, and hormonal influences (Bahr & Krosshaug, 2005; Griffin et al., 2006). Intrinsic risk factors that are modifiable, at least to some extent, include neuromuscular coordination and strength/conditioning status.

Non-Modifiable Intrinsic Risk Factors

Joint laxity has been implicated as a potential risk factor for ACL injury (Ahmad et al., 2006; Hewett, 2000; Ramesh, Von Arx, Azzopardi, & Schranz, 2005). Joints that are lax can move through a greater arc of motion before encountering natural restraint forces (e.g., ligaments, muscle and joint flexibility, eccentric muscle contractions, bony prominences). In traveling through a greater arc, attached segments can gain more momentum and provide greater opposition to the motion limiting forces of anatomical restraints. This may subject an individual with lax joints to greater risk of injury.

In an investigation of 169 patients who underwent ACL reconstruction surgery, Ramesh et al. (2005) noted that, of patients who ruptured their ACL's, 42.6% had generalized joint laxity and 78.7% exhibited genu recurvatum. Patients were compared to a non-injured control group in which 21.5 and 37% of subjects exhibited joint laxity and genu recurvatum, respectively. The researchers concluded that ACL injuries were more common in individuals with generalized joint laxity, particularly in those with genu recurvatum. These correlations suggest a greater risk of ACL injury with general joint laxity and knee hyperextension. There were only 32 females in the study and the researchers did not breakdown their findings by gender, so correlations between laxity and gender cannot be made from their research.

Other researchers have noted greater joint laxity in females (Ahmad et al., 2006; Huston & Wojtys, 1996; Rosene & Fogarty, 1999). In a study comparing NCAA Division I athletes (40 female, 60 male) and active healthy adult controls (14 female, 26 male), Huston and Wojtys found that athletes exhibited less joint laxity, as measured by anterior

tibial translation stress testing, than control subjects, and that women exhibited more joint laxity than men. Female athletes were found to have less joint laxity than male controls, but more than male athletes.

In a study of 123 adolescent and pre-adolescent boys (n = 70) and girls (n = 53), Ahmad et al. (2006) noted decreased joint laxity, as measured by a KT-1000 arthrometer, with maturation in boys but not in girls. Rosene and Fogarty (1999) measured anterior knee laxity in 60 collegiate athletes (22 male, 38 female) using a KT-1000 arthrometer and found significantly greater anterior tibial translation in the women.

Rosene and Fogarty (1999) suggested that joint laxity may be tied to increased rates of ACL injuries in females, but this speculation was rooted in other factors, such as hormonal, anatomical, and physiological differences between the sexes, that may contribute to increased laxity. Neither Huston & Wojtys (1996) nor Ahmad et al. (2006) was able to draw any correlations between knee laxity and ACL injury risk. Both groups cited references from researchers with findings on both sides of the issue. In essence, there is conflicting evidence as to whether knee laxity or generalized laxity contributes to ACL injury (Griffin et al., 2006).

Lower extremity alignment has also been implicated as an ACL risk factor. Included in this category are Q angle, genu valgum, genu recurvatumn, and subtalar pronation. The Q angle approximates the quadriceps muscle group line of action, and is measured as the angle between 1) the line joining the anterior superior iliac spine (ASIS) with the center of the patella, and 2) the line joining the tibial tuberosity with the center of the patella (Starkey & Ryan, 1996). Females tend to have larger Q-angles than males

(Hamill & Knutzen, 1995), presumably due to a wider pelvis. A wider pelvis, in turn, may result in genu valgum, which has been implicated as a risk factor for ACL injury (Griffin et al., 2006). In addition, landing with the knee in a valgus position has been cited as a direct cause of ACL injury (part of the so-called "position of no return") (Ireland, 1999; Ireland et al., 1997).

Excessive subtalar pronation may also contribute to the risk of ACL injury. When the subtalar joint pronates while the foot is bearing weight, the calcaneus everts, the talus internally rotates and plantar flexes, and the tibia internally rotates (Starkey & Ryan, 1996). This may predispose the ACL to greater stress, as internal tibial rotation results in tightening of the ACL (Gabriel, Wong, Woo, Yagi, & Debski, 2004; Kennedy, Weinberg, & Wilson, 1974). Other researchers determined that internal tibial rotation places more strain on the ACL than external rotation (Amis & Dawkins, 1991). This contradicts the "classic" mechanism of non-contact ACL injury, which implicates external tibial rotation as a causative factor for ACL injury (Ireland, 1999; Ireland et al., 1997). Thus, it is unclear as to whether excessive subtalar pronation increases the risk of ACL injury.

Another factor that may account for the higher incidence of ACL injury in women is altered knee arthrokinematics. Arthrokinematic descriptors of knee motion are glide and roll. In a pure gliding situation, the femur simply slides across the tibia, while in pure roll the femur behaves like a ball rolling across the floor (Figure 1). Due to anatomical restraints, however, roll and glide never occur in isolation. During flexion and extension of the knee, the adjacent joint surfaces roll and glide with respect to each other.

During closed chain knee flexion, the convex femur rolls posteriorly and glides anteriorly on the [relatively stationary] concave tibia. In extension, the femur rolls anteriorly and glides posteriorly. Non-contact ACL injuries occur when landing from a jump, changing directions quickly, or rapidly decelerating, all closed kinetic chain situations. If the normal roll/glide pattern of the joint is altered, ACL strain can be affected (Hollman, Deusinger, Van Dillen, & Matava, 2003).

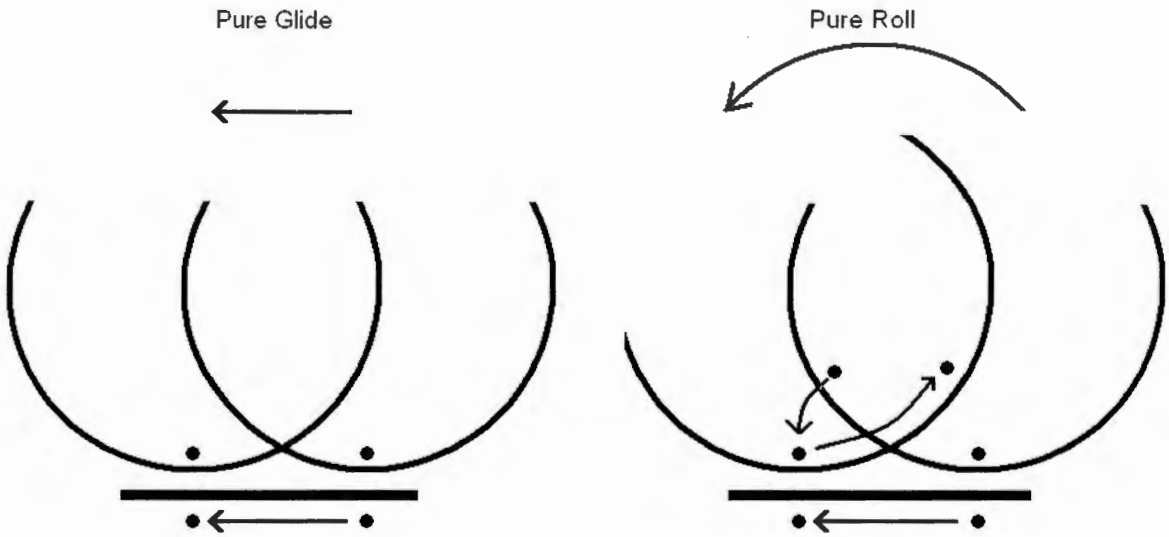

Figure 1. Comparison of arthrokinematic glide and roll. Curved lines represent femoral condyles, while flat lines represent the tibial plateau.

Hollman et al. (2003) compared sagittal plane knee arthrokinematics between young adult men and women during open and closed chain knee extension. The open chain movement was a seated knee extension exercise while the closed chain movement was a sit-to-stand exercise. Inferences of roll and glide were made through analysis of the instant center of rotation (ICR) of the knee through the range of motion. Decreased rolling (increased gliding) was evidenced by the ICR moving farther from the articular

surface, while the ICR moving closer to the articular surface was evidence of increased rolling (decreased gliding).

Hollman et al. (2003) found no differences in percent rolling between open and closed chain conditions in women, while the men exhibited significantly greater rolling during the closed chain condition, particularly at full knee extension. They observed that, in the closed chain exercise, men tended to roll into full extension while women glided into full extension. From this observation they suggested greater closed chain anterior tibial translation in women than men at terminal knee extension. This allowed them to theorize that women experience greater ACL strain than men as the knee approaches full extension. Their conclusions support the supposition of Ramesh et al. (2005), who theorized that the final pathway to ACL rupture is the knee sliding into hyperextension and the ACL "guillotining" itself in the femoral notch. Limitations of this study include a small sample of convenience and the lack of actual ACL strain measurements. Still, the evidence supports the notion that altered arthrokinematics may, in part, explain the increased incidence of ACL injury in females.

Others have suggested differences in intercondylar notch width and ACL size between men and women as potential explanations of the female ACL injury disparity (Anderson, Dome, Gautam, Awh, & Rennirt, 2001; Charlton, St. John, Ciccotti, Harrison, & Schweitzer, 2002; Shelbourne, Davis, & Klootwyk, 1998). In a prospective study, Shelbourne et al. made pre-surgical radiographic measurements of intercondylar notch width in 714 patients with ACL tears. Additionally, the surgeon measured actual notch width during reconstruction surgery; the two measurements correlated moderately

($r = .72, p < .01$). They also determined that women had narrower notches than men of equal height. All patients received 10-mm autogenous patellar tendon graft ACL reconstructions. Twenty-seven patients subsequently tore their contralateral ACL, while 19 tore their reconstructed ACL graft. There was a five-fold higher tear rate of the contralateral ACL in patients with narrower intercondylar notches ($p < .01$); there were no gender differences in tear rates among patients with equal notch widths. Of those who tore their grafts there were no differences by gender or notch width. Shelbourne et al suggest that neither notch width nor shape was a causative factor in ACL injury but, rather, a predictor of ACL size, theorizing that a patient with a smaller intercondylar notch would have a smaller ACL. They stopped short of hypothesizing that a smaller ACL would have less tensile strength than a larger ACL, but from the standpoint of fiber (rope, cord, string, thread, etc.) mechanics such a hypothesis would be logical.

Anderson et al. (2001) made bilateral measurements of knees in high school varsity male ($n = 50$) and female ($n = 50$) basketball players (mean age $\approx$ 16). Using magnetic resonance imaging (MRI) they determined the ACL width in both sagittal and transverse planes, then used those measurements to compute ACL cross-sectional area at the outlet of the intercondylar notch. They also measured notch width, femoral bicondylar width, and lateral condyle width. From the notch width and bicondylar width they computed the notch width index (NWI). Females had significantly smaller ACL's, notches, femurs, and lateral condyles than males; ACL size was smaller in females even after body weight correction. There was no correlation between notch width and ACL

area (males: $r = .177, p = .22$; females: $r = .225, p = .07$), suggesting that notch width is not a good predictor of ACL size, contrary to the supposition of Shelbourne et al (1998).

Charlton et al (2002) measured notch width and volume, ACL area and volume, and femoral bicondylar width using MRI in healthy adults (39 knees of 20 females, mean age 27.8 years; 52 knees of 28 males, mean age 26.3 years). Women had significantly smaller ACL's and notch widths than men. Interestingly, these findings were not attributed to gender, but to anthropometrics. In essence, the women had smaller ACL's and notch widths because they were, on average, smaller than the men. They also noted a significant correlation between ACL volume and intercondylar notch volume – a change in ACL volume of ~100 mm^3 for every 190 mm^3 change in notch volume ($p < .001$) – contradicting Anderson et al. (2001) and providing some support to the supposition of Shelbourne et al. (1998) that smaller notches contained smaller ACL's.

In addition to the above findings, Shelbourne et al. (1998) questioned the value of using the NWI to standardize notch width measurements between individuals of different height (Anderson et al., 2001; Souryal, Moore, & Evans, 1988). The value of the NWI lies in the presumption that notch width and femoral bicondylar width increase proportionately with increasing height. Shelbourne et al. found that, while bicondylar width increases proportionately with height, notch width does not. As such, two individuals with equal notch widths but different heights (and thus, different bicondylar widths) would have different NWI and, presumably, different risk for ACL injury, even if their ACL's were the same size. Shelbourne et al. argued that absolute notch width was a more useful measure. Charlton et al. (2002) made similar findings and echoed the

position of Shelbourne et al. concerning NWI. Anderson et al. (2001) found no differences in NWI based on gender. They acknowledge the findings of Shelbourne et al., but argue that absolute notch width does not vary proportionately with ACL size, does not standardize equally between people of different size or gender, and is not a good measure of classifying ACL injury risk either.

When clinicians first noted the increased incidence of ACL injuries in females, one of the first areas to receive research attention was the female menstrual cycle. As the major physiological difference between the sexes, it was natural to assume that fluctuating hormone levels associated with the female menstrual cycle might be, at least partially, responsible for ACL injury. The menstrual cycle is typically divided into follicular, ovulatory, and luteal phases, with variable hormone levels during different phases (Figure 2). Various researchers have found different correlations between ACL injury incidence and menstrual cycle phase (Beynnon et al., 2006; Hewett, Zazulak, & Myer, 2007; Wojtys, Huston, Boynton, Spindler, & Lindenfeld, 2002). Wojtys et al. found an increased incidence of ACL injury during the ovulatory phase of the menstrual cycle. Beynnon et al, however, found increased incidence of ACL injury during the preovulatory phase. Hewett et al., in a review of seven studies, also found evidence for increased ACL injuries during the preovulatory phase.

Depending upon the method of menstrual cycle phase assessment (serum hormone concentrations, urine hormone metabolite concentrations, or recall), the phase of the cycle when an ACL injury occurs may vary significantly. Wojtys et al. (2002) assessed menstrual cycle phase with a questionnaire and via a urine sample taken within

24 hours of ACL injury. They found only fair agreement between the two methods ($\kappa = .59$). Beynnon et al. (2006) reported that 73.9% of ACL injuries in alpine skiers occurred during the pre-ovulatory phase when phase was assessed by serum hormone concentrations. When phase was assessed by a menstrual history questionnaire, only 57% of ACL injuries occurred in the preovulatory phase. Both groups recommended that future ACL injury research in women document menstrual cycle phase via the more reliable methods of serum or urine assays, instead of the highly subjective measure of a recall questionnaire.

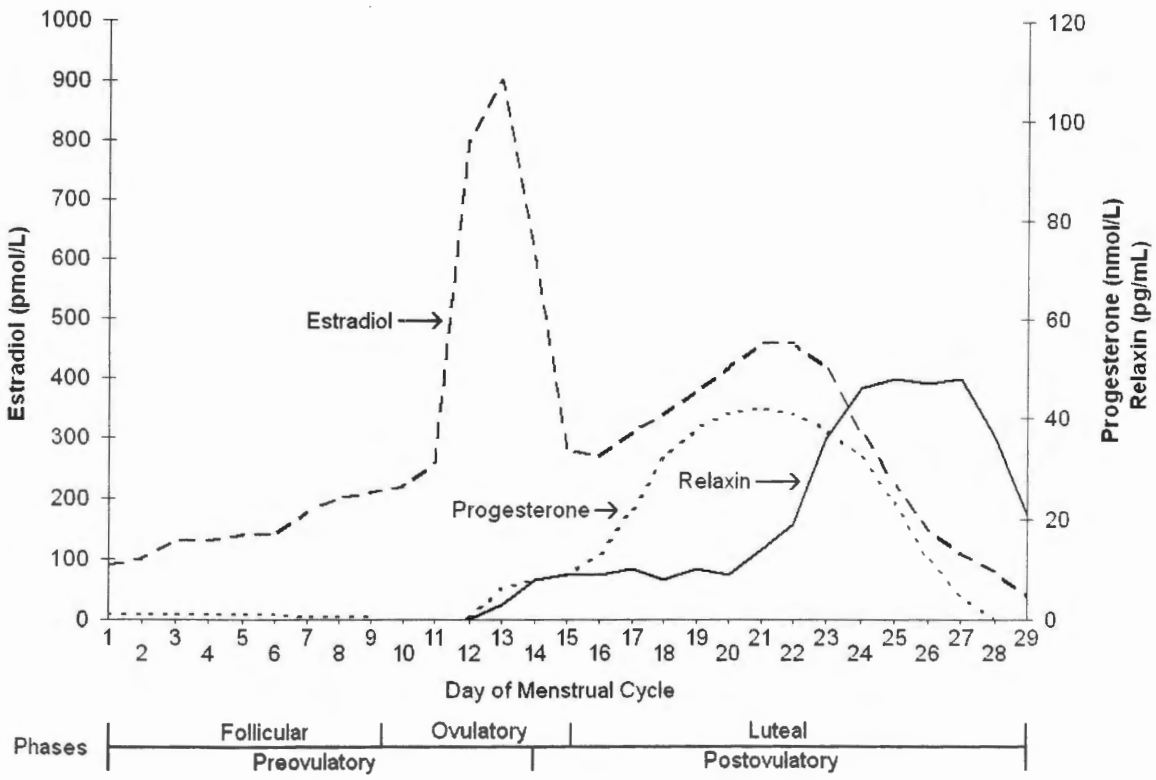

Figure 2. Serum hormone concentrations vs. day of menstrual cycle. Adapted from Harmon & Ireland (2000), Copyright Elsevier (2000), used by permission.

While researchers have documented differing risk of ACL injury across the menstrual cycle, there is debate over specific changes and effects on ligament structure. Some have suggested that estrogen and/or estrogen receptors on the female ACL alter the ligament's mechanical properties. Recent research in rats, however, has found this not to be the case (Warden, Saxon, Castillo, & Turner, 2006). Another hormone that has been implicated is relaxin, an insulin-like hormone linked to increased secretion of collegenase and decreased collagen synthesis and content (Dragoo, Lee, Benhaim, Finerman, & Hame, 2003). Relaxin was once thought to be present in significant amounts only during pregnancy, but has been found in some non-pregnant women during the luteal phase of the menstrual cycle (Figure 2). Dragoo et al. found evidence of relaxin receptors on human female ACL's, but not on male ACL's. The presence of this collagenolytic hormone in pregnant and some non-pregnant women, along with relaxin receptors on female ACL's, may have implications for ACL injury risk, as collagen is the major load bearing protein in the ACL structure (Dragoo et al.). Additionally, the increase in relaxin and estrogen during the luteal phase is accompanied by an increase in progesterone, which is secreted from the corpus luteum after ovulation. There is the possibility of a synergistic effect between relaxin and estrogen (Dragoo et al.), and possibly progesterone, as no one has been able to specifically link any of these individual hormones to a change in ligament mechanical properties.

The above intrinsic risk factors are considered non-modifiable. In the opinion of some, research time and funds should not be expended on issues that cannot be changed, but rather on actual ACL injury prevention through education and conditioning programs

(Teresa Stadler, MD, Orthopædic Surgeon, Texas A&M Medical School, personal communication, February 27, 2004, referring specifically to the female menstrual cycle; Warden et al, 2006). The following ACL injury risk factors (both intrinsic and extrinsic) are considered modifiable and present the most promising areas of research in ACL injury prevention, both in women and men.

Modifiable Intrinsic Risk Factors

Many researchers have documented differences in the manner of performance of activities between males and females (Chappell et al., 2002; Decker, Torry, Wyland, Sterett, & Steadman, 2003; Fagenbaum & Darling, 2003; Ford et al., 2005; Hollman et al., 2003; Huston & Wojtys, 1996; McLean, Lipfert, & van den Bogert, 2004; Zeller, McCrory, Kibler, & Uhl, 2003), leading to speculation about differences in neuromuscular coordination between the sexes and the possible effects of such differences on the risk of ACL injury.

Huston & Wojtys (1996) conducted a comprehensive evaluation of NCAA Division I athletes and age & sex matched control subjects. They measured knee joint laxity (arthrometer), anterior tibial translation (stress test), isokinetic strength and endurance of hamstrings and quadriceps (isokinetic dynamometer), neuromuscular efficiency (isokinetic dynamometer), muscle reaction time (EMG during stress test), and muscle recruitment order (EMG during stress test). In general, non-athletes had looser knees than athletes, and women had looser knees than men. Men demonstrated greater (body weight normalized) isokinetic strength and endurance in both the hamstrings and quadriceps than women. Female athletes took significantly longer to reach peak knee

flexion torque at both 60 and 240 deg/s. Female athletes also took longer to generate peak hamstring torque than peak quadriceps torque at both speeds tested.

During the anterior tibial translation stress test (Huston & Wojtys, 1996), female athletes exhibited a quadriceps-hamstrings-gastrocnemius muscle recruitment order, while the other groups recruited muscles in the hamstrings-quadriceps-gastrocnemius order. This muscle recruitment order of the female athletes may indicate an increased risk for ACL injury. Interestingly, the five strongest female athletes recruited muscles in the hamstrings-quadriceps-gastrocnemius order in response to anterior tibial translation. This suggests that conditioning may have an effect on muscle recruitment order and that women may be able to alter muscle recruitment patterns through a "muscle rebalancing" conditioning program (Huston & Wojtys, 1996).

Chappell et al. (2002) compared knee kinetics between men and women during backward, forward, and vertical stop-jump tasks. Women experienced greater ground reaction forces (as a percentage of body weight) during landing and takeoff than men. Women also had greater extension and valgus moments than men during the landing phase of all stop jump tasks. Anterior tibial shear force was greater in women during the landing phase, while it was greater in men during the takeoff phase. Chappell et al. theorized that women might have different motor control strategies than men, and that these differences may result in women assuming positions that expose the ACL to greater risk of injury.

Decker et al. (2003) investigated differences between men and women in the performance of landing from a 60 cm drop-jump. They found that women landed in a

more upright body position, with less knee flexion and more ankle plantar flexion than men. The women traveled through greater ankle and knee ranges of motion (ROM) than the men, and had greater joint angular velocities, indicating faster loading of the knee and ankle. There were no differences between men and women in vertical ground reaction forces (as a percent of body weight) or in the timing of those forces during landing. Women and men both utilized the knee as the primary energy absorber during the impact phase (first 100 ms after touchdown), but women relied, to a far greater extent than men, on the knee and ankle joints for energy absorption (Table 1). Such reliance on smaller joints (with smaller muscles) may place those joints at greater risk for injury. One potential drawback to this study is the assumption of landing symmetry. Decker et al. measured GRF on one leg only, and assumed leg loading symmetry on landing. Symmetry of leg loading during two-footed jump landings has previously been investigated and found to be an unreliable assumption (Schot, Bates, & Dufek, 1994).

During their arthrokinematics research, Hollman et al. (2003) made electromyographic measurements of the semitendinosis and vastus lateralis muscles during the exercises to determine relative co-contraction activities and muscle firing patterns between the hamstring and quadriceps muscle groups. Although there were no relative differences in quadriceps or hamstring activity between men and women, the authors noted a significant interaction between gender and movement condition, with women exhibiting a greater quadriceps-hamstring ratio during the closed chain activity ($p < .01$). This study provides additional evidence for quadriceps dominance among women which may partially explain the increased risk of ACL injury.

Table 1

Kinetic Variables from Decker et al. (2003) (Mean (SD))

	Male	Female
Negative joint work (%BW×Ht) (indicates energy absorption)		
Hip	-4.86 (1.45)	-3.20 (1.92)*
Knee	-6.59 (2.07)	-8.58 (2.15)*
Ankle	-4.74 (1.40)	-6.42 (1.71)*
Joint contribution to energy absorption during landing		
Hip	30%	18%
Knee	41%	47%
Ankle	29%	35%

*Significantly different from males, $p < .05$

Zeller et al. (2003) evaluated kinematics and electromyography during single leg squats on the dominant leg. Kinematic variables analyzed were joint ranges of motion at the ankle, knee, and hip, as well as trunk flexion and lateral flexion. Muscles monitored included the rectus femoris, vastus lateralis, medial gastrocnemius, biceps femoris, gluteus maximus, and gluteus medius. Kinematic analysis showed that women performed the single legged squat with the foot pronated throughout, more knee valgus, less knee varus, and greater hip adduction than men. Men showed less pronation and no knee valgus during the squat. Neither group abducted the hip during the squat. Women had significantly greater activation of the rectus femoris than men ($p < .05$).

The kinematic results of Zeller et al. (2003) indicate ligament dominance among women. Women adopted hip, knee, and foot positions that placed them in the classic stance during which non-contact ACL injuries typically occur: hip adduction, knee valgus, and sub-talar pronation. This is the typical ligament dominant position described by Hewett et al. (2002). The far greater activation of the quadriceps musculature by women suggests quadriceps dominance for knee stabilization. Both quadriceps and ligament dominance have been suggested as risk factors for ACL injury in women (Hewett et al.).

McLean et al. (2004) studied the effect of gender and a simulated opponent on the kinematics and kinetics of the stance leg during side-step cutting maneuvers. They determined that women initiated the cutting maneuver with less hip and knee flexion, more knee valgus, more subtalar pronation, and less hip abduction than the males. This description is typical of the ligament dominant pattern observed by Hewett et al. (2002), and places the ACL at greater risk of injury. McLean et al. made use of a single force plate but, unlike Decker et al. (2003) who assumed landing symmetry during two-footed jump landings, McLean et al. were investigating the kinematics and kinetics of the stance leg during a run-and-cut maneuver. Therefore, landing symmetry is not an issue with this study.

Ford et al. (2005) investigated the kinematics of an unanticipated jump-stop-cut maneuver in adolescent basketball players (54 male, 72 female). They found that the girls landed with greater knee valgus (abduction) angles than the boys ($p = .033$); girls also displayed greater maximum ankle eversion during the stance phase ($p < .001$). There were no significant differences in maximum knee valgus, knee flexion at ground contact,

or maximum knee flexion. Greater valgus positioning of the knee is indicative of ligament dominance (Hewett, Paterno, & Myer, 2002), in which the athlete relies on knee ligaments (passive structures) rather than musculature (active structures) to absorb forces during landing. Ligament dominance can impart high knee valgus moments that place excessive stress on the ACL and may increase the risk of injury.

Fagenbaum & Darling (2003) found evidence for *decreased* ACL injury risk in women as compared to men. They measured kinematics and electromyography during one-legged drop landings from 25.4 and 50.8 cm (10 and 20 in.); measurements were made under fatigued and unfatigued conditions. Women landed with greater knee flexion than men, and exhibited similar muscle recruitment patterns as the men. These characteristics are indicative of greater ACL protection and less risk of injury. Fagenbaum & Darling drew their study participants from NCAA Division I men's and women's basketball teams at the same university. They considered their participants to be elite athletes, which may partially explain their findings, as elite athletes tend to be better conditioned and better coached than recreational or lower division athletes. Perhaps some factors of the men's and women's basketball programs at that university were duplicated that might account for the similarity of muscle recruitment patterns. Furthermore, the researchers only analyzed knee flexion angles and quadriceps, hamstring, and gastrocnemius electromyographic data. There may be other factors that may have indicated an increased ACL injury risk in their female participants, but such factors were neither measured nor accounted for.

Other researchers have documented differences in the manner of activity performance with maturation (Ahmad et al., 2006; Hass et al., 2003; Hass et al., 2005; Hewett, Myer, & Ford, 2004), with similar speculation about neuromuscular coordination and ACL injury risks as those advanced with the gender issue. Ahmad et al. (2006) investigated changes in knee laxity (KT-1000 arthrometer) and hamstring and quadriceps strength (handheld dynamometer) with maturity in boys and girls attending a city soccer camp. They divided the participants into four groups based on age (boys) or menstrual cycle status (girls). Boys 13 and younger, and premenarchal girls comprised the "immature" groups, while older boys and girls 2 or more years post menarche comprised the "mature" groups. Mature boys showed significantly less knee laxity than the other groups ($p < .05$). Boys and girls showed significant strength increases with maturity; boys increased their hamstring strength much greater than their quadriceps strength, while the opposite was true for the girls. Immature boys and girls had similar quadriceps-to-hamstring (Q:H) strength ratio. Mature boys had lower Q:H ratios than immature boys, while mature girls had significantly higher Q:H ratios than the other groups ($p < .05$), indicating quadriceps dominance. Quadriceps dominance has been identified as an ACL injury risk factor, and may be a maturation issue, a coaching or training issue, or a combination of these and/or other effects.

Hass et al. (2003, 2005) investigated the effects of maturation on the kinematics and kinetics of stride-jump and single-leg drop-jump landings in mature and immature female athletes. Sixteen prepubescent girls (8-11 years) and 16 women (18-25 years), at least 6 years post menarche with a regular menstrual cycle, participated in these studies.

Description of methods and anthropometric characteristics of the study participants lead me to believe that the same athletes participated in both studies at the same time. They discovered similar kinematics in both studies, noting that the girls landed with more knee flexion than the women. Kinetics, however, were different between the two studies. In stride-jump landings, girls produced less ground reaction force than the women, while in the drop-jump landings, women produced less ground reaction force than the girls. This suggests the mature females were more adept at anticipating the impact of the drop-jump than their immature counterparts, even though they landed in a position of less knee flexion, placing the knee at greater risk of injury.

Hewett et al. (2004) measured kinematics and kinetics of the knee during drop-jump landings among children at various maturational stages (pre-pubertal, early pubertal, and late or post-pubertal). They found that, as girls matured, they landed with greater knee valgus positioning, achieved greater maximum valgus positions during the landing, and tended to absorb more of the impact of landing with their dominant leg. Their results suggest a maturational component to the increased risk of ACL injury in females.

The greatest drawback in most of the aforementioned neuromuscular research is small sample size (Table 2). Only 4 of the 12 studies made use of relatively large samples (Ahmad et al., 2006; Ford et al., 2005; Hewett et al., 2004; Huston & Wojtys, 1996). The other studies incorporated much smaller samples, limiting their statistical power and ability to draw inferences to other populations. While small samples of convenience are

common in biomechanics research, interventional studies typically require many more participants to demonstrate sufficient statistical power to reveal treatment differences.

As numerous researchers have identified neuromuscular deficiencies in females, attention has shifted from reconstruction and rehabilitation to efforts aimed at prevention. A number of clinicians have successfully decreased the risk and occurrence of ACL injuries in females through neuromuscular and proprioceptive training programs (Caraffa, Cerulli, Projetti, Aisa, & Rizzo, 1996; Heidt, Sweeterman, Carlonas, Traub, & Tekulve, 2000; Hewett, Lindenfeld, Riccobene, & Noyes, 1999; Hewett et al., 2002; Holm et al., 2004; Myer, Ford, & Hewett, 2004; Myer, Ford, Palumbo, & Hewett, 2005). All of these programs are multi-faceted comprehensive training and conditioning programs designed to teach or re-teach athletes (particularly females) how to run, jump, land, cut, and stop in a manner that reduces stress on the ACL. All have met with great success in the programs/groups in which they were implemented. Female ACL injury rates, however, remain higher than male rates, suggesting that neuromuscular training programs are not implemented universally, are implemented improperly, or only provide a partial solution to the female ACL injury question.

Table 2

Comparisons Among Various Neuromuscular Performance Studies

	Adult		Minor			
Research Group	F	M	F	M	Sport	Skill Level
Ahmad et al. (2006)			53	70	soccer	various, ages 10–18
Chappell (2002)	10	10			various	recreationally active
Decker et al. (2003)	9	12			volleyball, basketball	intramural
Fagenbaum & Darling (2003)	8	6			basketball	NCAA Div. I varsity
Ford et al. (2005)			72	54	basketball	middle/high school
Hass et al. (2003, 2005)	16[a]		16[b]		various	recreationally active
Hewett et al. (2004)			100	81	soccer, basketball	various, ages 10–18
Hollman et al. (2003)	5	6			not specified	healthy university students
Huston & Wojtys (1996)	40[c]	60[c]			various	NCAA Div. I varsity
McLean et al. (2004)	8	8			not specified	not specified
Zeller et al. (2003)	9	9			Various	NAIA varsity

[a]Age 18-25. [b]Age 8-11. [c]Also examined 14 F & 26 M healthy age matched control subjects

Extrinsic Risk Factors

Extrinsic risk factors concern possible external causes of injury, are generally modifiable, and include shoe type, playing surface, and protective equipment (Bahr & Krosshaug, 2005; Griffin et al., 2006). Much of the research on the effects of shoe type and playing surface on injuries has been directed at the shoe-surface interface. The nature of the materials at the shoe-surface interface determines the friction coefficient. For a given surface, the higher the coefficient of friction, the less likely the shoe is to slip on that surface or, in other words, the greater the traction. Depending upon the activity engaged in, an athlete may want high or low friction. In addition to the shoe and surface materials, environmental factors such as temperature and moisture can affect the coefficient of friction by altering material properties or affecting surface lubrication. If, for a given activity, friction is too high, the shoe may become fixated to the surface, resulting in the transfer of excessive forces and/or moments to the skeleton that can cause injury (Lambson, Barnhill, & Higgins, 1996; Torg, Stilwell, & Rogers, 1996). If, however, friction is too low, slippage and falls become a problem (Lambson et al.).

Since AstroTurf (Southwest Recreation Industries, Leander, TX) was introduced in 1966, anecdotal reports of increased musculoskeletal injuries have plagued the artificial turf industry. Numerous studies have found conflicting evidence concerning injury rate on artificial turf versus natural grass. Research has, however, determined that shoes tend to stick more to artificial turf at higher ambient temperatures (Torg et al., 1996). Other researchers have found evidence of increased injuries under varying

temperature and moisture conditions on both grass and artificial turf (Orchard & Powell, 2003; Orchard, Seward, McGivern, & Hood, 1999).

In an analysis of National Football League games from 1989-1998, Orchard and Powell (2003) found the ACL injury incidence rate to be lower in games played outdoors in cooler weather, regardless of playing surface. This bias did not extend to games played in domes, where temperature is regulated. In a review of ACL injuries in the Australian Football League, where games are played exclusively on grass, Orchard et al. (1999) found evidence of increased ACL injuries when the fields were dry, warm, and hard. They theorized that moister, softer fields had lower traction (friction), and restricted foot motion less than drier, harder fields. Lambson et al. (1996) noted a significantly higher rate of ACL injuries in players who wore shoes with longer cleats (0.017%) versus players who wore shoes with shorter cleats (0.005%) ($p = .0062$). These results, while specific to American and Australian football, have implications for athletes in any sport on grass, turf, or dirt where cleats are worn for traction (e.g., soccer, rugby, softball, baseball), and suggest that shoe selection be carefully considered.

What of the athlete who competes not in cleats on grass, dirt, or turf, but in court shoes on wood, clay, grass, composite, asphalt, or concrete? Friction, or traction, is still a concern. The type of shoe and playing surface still determine the friction coefficient which, if too high or low, can lead to injuries. While research on grass/turf sports abounds, only one study was found that examined the relationship between floor material and ACL injury risk in court sports (Olsen, Myklebust, Engebretsen, Holme, & Bahr, 2003). In this study, Olsen et al. examined ACL injuries in the top three divisions of

Norwegian team handball over a period of seven seasons between 1989 and 2000. During that time there were 174 total ACL injuries; Olsen et al. were able to determine the floor type for only 53 of the injuries. Upon analyzing the injury and floor type data, Olsen et al. concluded that women were at greater risk of ACL injury when playing on artificial floors (Table 3), which had higher friction coefficients than wooden floors. Their results must be interpreted with caution, because floor data was available for only 53 of the 174 injuries, male injury rates were not tracked in the 1998-2000 seasons, individual shoe-surface friction coefficients were not measured, only a small percentage of the floors were actually tested for friction coefficient (Table 4), and friction coefficients varied from floor to floor. Still, the above mentioned studies provide rationale for considering the shoe-surface interface as a risk factor for ACL injury, regardless of sport, playing surface, or shoe type.

Table 3

ACL Injuries vs. Floor Type in Norwegian Team Handball, 1989-2000

ACL injuries		Injury exposure (hours)		Injury incidence per 1000 hours exposure (mean(*SD*))	
Women	Men	Women	Men	Women	Men
		Wooden floors			
8	4	19,474	12,502	0.41 (0.09)	0.32 (0.13)
		Artificial floors			
36	5	37,548	24,612	0.96 (0.04)	0.20 (0.12)

Table 4

Friction Coefficients (μ) of Venues Used for Norwegian Team Handball.

Floor type	Venues used for match play	Venues tested by NBI	μ mean (*SE*)	μ Range
Wooden	91	7	0.46 (0.07)	0.37–0.53
Artificial	170	22	0.57 (0.05)	0.53–0.73

Testing conducted by the Norwegian Building Research Institute (NBI), 1990-2000

Ankle Bracing as a Risk Factor

Prophylactic knee and ankle bracing have been advocated as effective in reducing injuries to the lower extremities. Evidence concerning the use of such braces is, however, conflicting. Using data from a study of football players at the U.S. Military Academy, Griffin et al. (2006) estimated that players who did not wear knee braces suffered three times more ACL injuries than those who were braced (16 total ACL injuries, four in braced athletes, 12 in unbraced players). Data from multiple sports in multiple North Carolina high schools over several seasons (Yang et al., 2005), however, implicated ankle and knee bracing with increased risk of injury to those joints.

Ferguson (1973) recommended against the use of prophylactic ankle taping in healthy individuals, arguing that the artificial support of the tape took away from the natural protective functions of the skeleton (particularly the subtalar joint) and knee and ankle musculature. In his article, Dr. Ferguson cited several studies without specific reference to any published results; he also referred to anecdotal evidence from his

practice as a major league baseball team physician. Santos et al. (2004) found evidence that suggested ankle bracing may increase the risk of knee injury during forceful trunk turning while in a single-leg stance. Sitler et al. (1994) did not find any evidence of increased knee injury risk from the use of prophylactic ankle bracing in intramural basketball players. After considering the data and potential confounding factors, Yang et al. (2005) determined more research needed to be done to determine if bracing does indeed increase the risk of injury.

While the idea of ankle bracing as a risk factor for ACL injury is controversial, indirect evidence lends some support to this notion. From the 1970's to the 1990's, lower extremity alpine skiing injuries decreased, but skiing associated ACL injuries increased (Ekeland, 1995; Natri, Beynnon, Ettlinger, Johnson, & Shealy, 1999; St-Onge, Chevalier, Hagemeister, van de Putte, & de Guise, 2004). Some of this increase has been attributed to the ski-boot-binding system.

The ski-boot-binding system leads to three ACL injury mechanisms that are unique to skiing: valgus external rotation, boot induced anterior drawer, and flexion-internal rotation (Ekeland, 1995; Natri et al., 1999). Boot design has been implicated in these mechanisms; prior to the advent of the modern (stiff) ski boot, severe knee sprains (including ACL ruptures) were less common than they are now (Natri et al.). Modern ski boots place the ankle in dorsiflexion and restrict plantarflexion (Natri et al.).

In contrast to ski boots, the primary motion restricted by ankle braces and ankle taping techniques is inversion (Barkoukis, Sykaras, Costa, & Tsorbatzoudis, 2002; Callaghan, 1997; Hume & Gerrard, 1998), although plantarflexion is somewhat restricted

by the manner of tape application or brace construction (Arnheim & Prentice, 1997; Cordova, Ingersoll, & LeBlanc, 2000). Despite this difference, restriction of ankle motion has the potential to affect the knee joint (Dugan, 2005; Leetun et al., 2004; Santos et al., 2004) due to cascading effects in the kinetic chain.

Summary

Since the 1972 enactment of *Title IX*, participation in organized athletic programs by females has increased dramatically. Concurrent with the increase in female participation in athletics is an increased incidence of injuries in general and, in particular, injuries to the ACL. Females have been found to be at greater risk of sustaining an ACL injury than males. In attempting to solve the female ACL injury enigma, numerous risk factors for ACL injury, in both men and women, have been identified. In the past 30 years, research in ACL injuries has progressively moved from treatment and repair methods toward prevention methods, and from identifying risk factors toward determining which risk factors are modifiable. Ankle taping was identified as a potential risk factor in ACL injuries as early as 1973, although only anecdotally (Ferguson, 1973).

Ankle bracing and taping are generally accepted practices to minimize ankle sprains. Most of the studies conducted in relation to ankle bracing and/or taping, however, have focused on the effects of such treatments at the ankle; only two studies were located that examined the effects of ankle bracing on the knee. One study suggested that the use of ankle bracing may be a risk factor for ACL injury (Santos et al., 2004), while the other contended ankle bracing has no effect on the injuries of the knee (Sitler et al., 1994). The need for additional study is evident.

CHAPTER III

METHODS

Participants

Eight healthy athletic females (three gymnasts, three volleyball players, one basketball player, and one soccer player) were recruited from the university student population. Four were university athletes (gymnasts and one volleyball player) while the others had recently competed in high school varsity sports. Participants were pre-screened for orthopædic health history and were excluded on history of knee ligament or meniscus injuries, congenitally absent ACL or PCL, and ankle sprains for which the participant was symptomatic or under physician care. During a pilot study it was determined that placement and tracking of reflective markers became more difficult as body mass index (BMI) increased. Therefore, volunteers with BMI greater than 27 were also excluded. Athletes who wore physician or therapist prescribed ankle braces were also excluded. This investigation was reviewed and approved by the Texas Woman's University Institutional Review Board; all participants granted informed consent prior to health history disclosure and data collection. Participant's age, mass, and height were recorded and body mass index (BMI) computed; data are presented in Table 5.

Table 5

Participant Characteristics (n = 8)

Parameter	Mean (*SD*)
Age (years)	20.3 (1.5)
Height (cm)	167.0 (6.7)
Mass (kg)	64.0 (9.1)
BMI	23.0 (2.8)
Jump height (cm)	39.9 (5.5)

The number of participants was based on pilot data collected from a single female volunteer in the age range of the target population. The volunteer performed four, two-legged squat trials with and without Webly® ankle braces, squatting as low as possible while keeping her heels on the floor. Orientation angles (ankle dorsiflexion and abduction/adduction, and knee valgus, flexion, and internal rotation) at maximum knee flexion were determined and compared using paired t-tests (braced vs. un-braced). The t-test with the largest average difference and standard deviation between conditions was selected as the most conservative case for power calculations. Means, standard deviations, desired α (.05), and desired power (.80) were entered into two different power calculations (Ouwehand, 2004) to determine an appropriate sample size. Based on these calculations, the minimum number of participants was determined to be eight.

Design

For the deep squat trials a single-factor, within-subjects repeated measures design was used. The factor was bracing condition (untaped, athletic taping, and Webly® ankle orthosis). Dependent variables evaluated were maximum knee flexion and maximum ankle dorsiflexion; right and left leg values were averaged for comparison among bracing conditions. Differences in dependent variables among bracing conditions were evaluated using repeated measures multiple analysis of variance (MANOVA) with Bonferroni adjustment for multiple comparisons. All statistical analyses were conducted with SPSS 14.0 (SPSS, Inc., Chicago).

A two-factor (3 × 3), within-subjects repeated measures design was used for the drop-jump trials; factors were bracing condition and side (upright, ipsilateral, and contralateral). The side factor was chosen due to similar motion patterns among movement trials. Data from both legs were pooled as follows: the upright condition combined both right and left leg data of the stand-up condition; the ipsilateral condition combined right leg data from the rightward jump and left leg data from the leftward jump, while the contralateral condition combined left leg data from the rightward jump and right leg data from the leftward jump. Kinematic dependent variables analyzed were knee valgus, rotation, and flexion orientation angles at foot strike and maximum and anterior tibial translation from touchdown to maximum knee flexion. Kinetic dependent variables analyzed were knee moments at foot strike and at maximum in the sagittal (flexion/ extension), frontal (varus/valgus), and transverse (torsion) planes, and maximum ground reaction force during the landing. Knee joint moments were normalized to percent body

weight multiplied by height (%BW×Ht) (Decker et al., 2003). Ground reaction forces were normalized to percent body weight (%BW) (McNitt-Gray, 1993). Differences in dependent variables among bracing conditions and sides were evaluated using repeated measures MANOVA. Significant MANOVA's were followed with univariate tests and a Bonferroni adjustment was made for multiple comparisons. In order to minimize practice effects order of bracing conditions and trials within conditions were randomized using balanced Latin squares.

Calibration and Motion Capture

Eight digital cameras (Panasonic AG-DVC-15/20 with Panasonic AG-LW4307 wide angle lens, Panasonic Broadcast & Television Systems Company, Secaucus, NJ) were utilized to capture calibration, static, and motion trials. Video images were directly captured to a laboratory computer (PC) at 60 Hz using Kwon3D XP motion analysis suite (Visol, Seoul, Korea). Ground reaction force data from two force plates (American Mechanical Technology, Inc., Watertown, MA) were collected at 240 Hz simultaneously with video images. A calibration frame (Visol) of dimensions 3 m × 1 m × 2 m, with 48 reference markers on 8 vertical poles, was used to define the area of motion capture (Figure 3). Additional markers (2 cm diameter) were placed 7.3 cm above the posterior corners of the right side force plate and above the posterior left corner of the left side force plate. The additional markers were used to align the global reference frame with the primary (right side) force plate and to identify force plate locations in motion trial reconstructions. A camera calibration trial was captured prior to each data collection session. Prior to digitizing static and motion trials, the appropriate calibration trial was

digitized and a calibration was conducted using a non-linear least-squares reconstruction algorithm, also known as the direct solution method (DSM) (Christopher, Yoon, & Kwon, 2008; Kwon, 2005). Digitizing error across all eight cameras and seven calibrations was 0.06 ± 0.01 pixels. Reconstruction error of calibration frame reference points was 0.24 ± 0.04 cm.

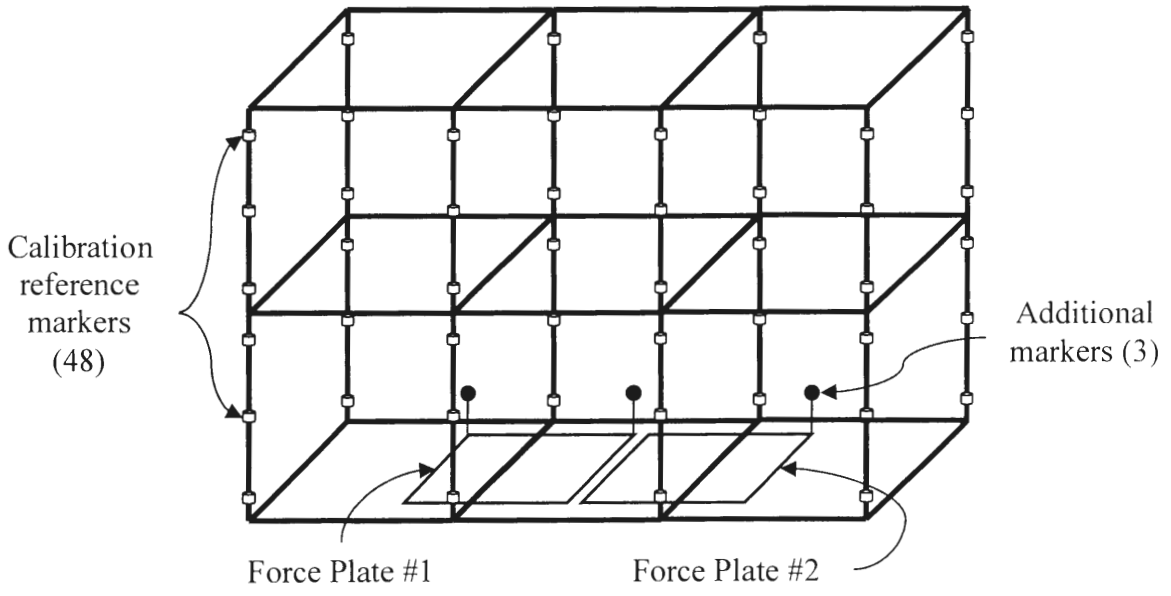

Figure 3. Video calibration frame, force plates, and additional markers.

Procedures

Participants changed into athletic attire and performed a stretching/warm-up routine to which they were accustomed. After the warm-up, standing reach was measured as follows: participants stood with shoulders fully flexed, and elbows, wrists, and fingers fully extended; the lowest slat on a Vertec™ jump height trainer (Sports Imports, Columbus, OH) was set to the height of the participant's extended fingers. Participants then performed three two-footed maximum counter-movement jumps, swatting the

Vertec™ slats at the peak of each jump. The highest slat moved indicated the height jumped. Jump height was determined as the average of the three jumps.

Following jump height measurement, retro-reflective markers (1 cm diameter) were affixed over the following bony landmarks: anterior superior iliac spine (ASIS), sacrum, greater trochanter (GT), lateral and medial femoral epicondyle, and lateral and medial malleolus. Additional markers were placed on the anterior thigh, lateral thigh, anterior shank (superior and inferior), lateral shank, and exterior of participant's shoes at the calcaneus and the base of the second metatarsal. Markers were affixed to the participant's skin where possible. Except for the sacrum, all markers were placed bilaterally (Figure 4).

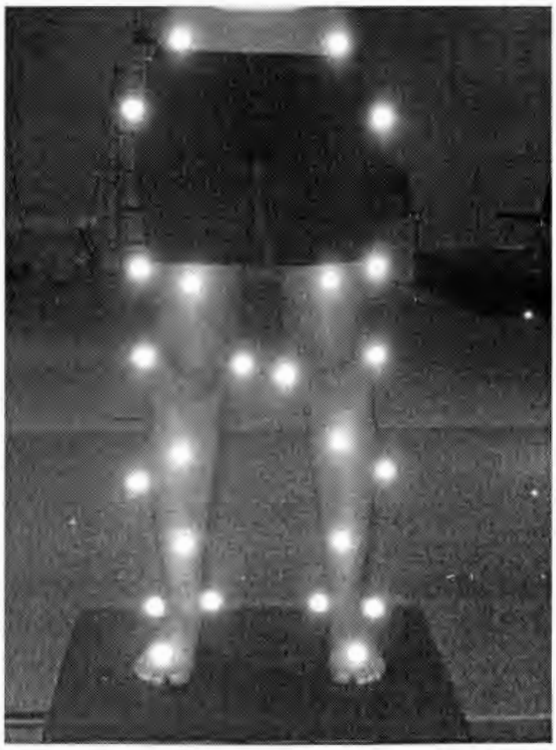

Figure 4. Lower body marker set (anterior view). For motion trials, medial markers were removed and participants wore their own shoes.

A static trial was captured with medial markers affixed to aid in defining locations of joint centers. Following the static trial, each participant performed a deep squat, squatting as low as possible while keeping their heels in contact with the floor. Squat trials were performed to determine if the different bracing conditions affected knee and ankle joint range of motion. Squats were performed in three bilateral ankle-bracing conditions: untaped; closed basketweave athletic taping with three medial-to-lateral stirrups pulling the ankle into eversion, three heel locks and three figure-eights; and Webly® Ankle Orthosis (Hely & Weber, Inc., Santa Paula, CA) pulling the ankle into eversion (Figure 5). Medial ankle and knee markers were removed following the static trial. Participants then performed drop-jump trials in the three bracing conditions. Taping and bracing were applied by a certified athletic trainer.

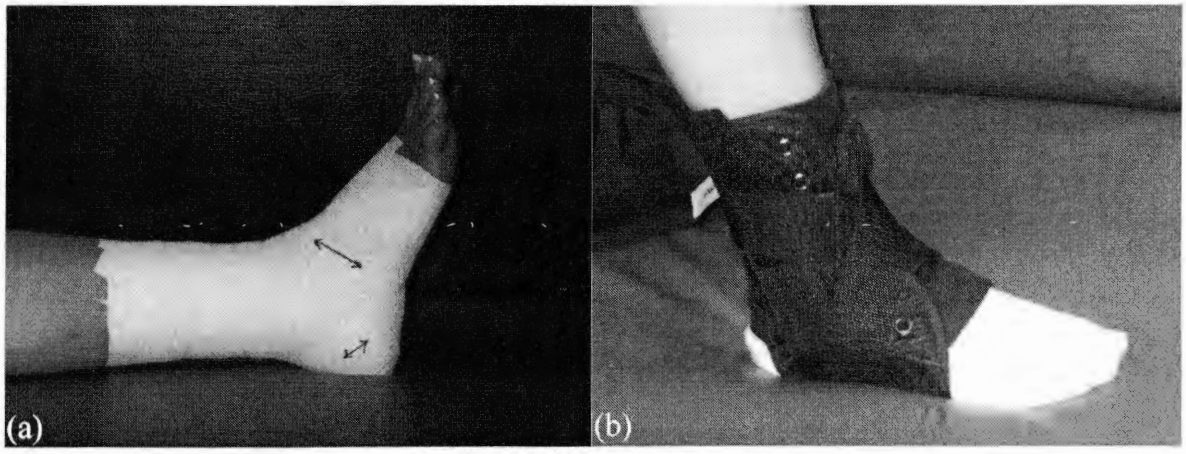

Figure 5. (a) Closed basketweave taping showing figure-eights (upper arrow) and heel locks (lower arrow). (b) Webly® Ankle Orthosis.

In order to simulate game/practice conditions under which an athlete leaps as high as possible (e.g., to block a volleyball spike or retrieve a basketball rebound) and then drops back to the floor, a platform was set to each participant's jump height. Participants

mounted the platform and, upon direction, stepped off and dropped from the platform, landing with each foot on separate laboratory force plates (AMTI). At the time of the drop, a visual cueing device (Visol) indicated a direction to move (Figure 6). Upon landing, participants stood-up straight or pivoted and stepped to the right or left, depending on the visual cue received. Two valid trials in each direction under each bracing condition were captured.

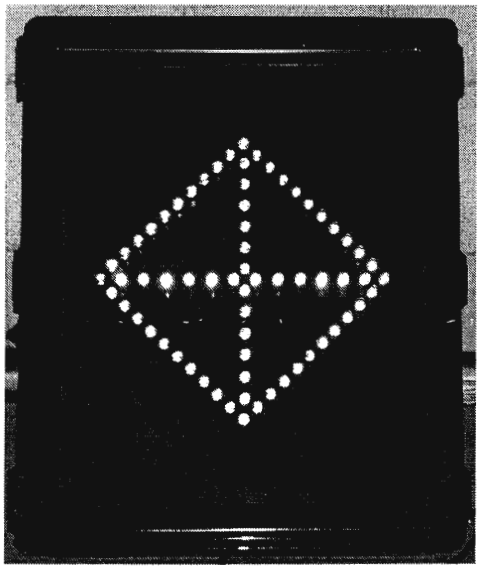

Figure 6. Visual cueing device.

Joint Centers, Reference Frames, and Orientation Angles

Each static, squat, and drop-jump trial was digitized using Kwon3D XP to determine 3-dimensional (3D) coordinates of the reflective markers. Joint center locations were determined based on 3D marker coordinates. Hip joint centers were defined using the Andriacchi-Tylkowski hybrid method (Bell et al., 1990). In this method, the anterior-posterior position of the hip joint center was determined by the

position of the greater trochanter marker. The mediolateral position was defined as a point 14% of the inter-ASIS distance (IAD) medial to the ASIS and the superior-inferior position was defined by a point 30% IAD inferior to the inter-ASIS line (Figure 7).

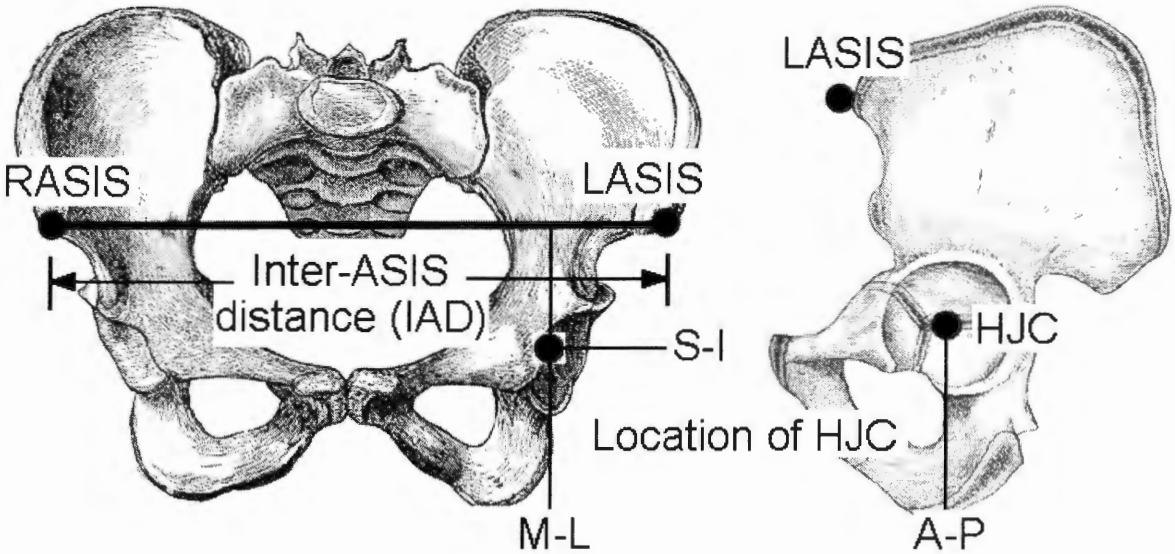

Figure 7. Andriacchi-Tylkowski hybrid method of hip joint center (HJC) location. R = right, L = left, ASIS = anterior superior iliac spine, M-L = mediolateral, S-I = superior-inferior, A-P = anterior-posterior. The A-P location of the HJC was defined by the position of the greater trochanter marker; the M-L and S-I locations of the HJC were 14% IAD medial and 30% IAD inferior, respectively, from the ipsilateral ASIS. Unedited images are in the public domain and were retrieved November 9, 2008, from http://en.wikipedia.org/wiki/Image:Gray242.png and Image:Gray237.png.

Knee joint centers were defined as the midpoint of a line connecting the medial and lateral femoral epicondyles. Ankle joint centers were defined as the midpoint of a line connecting the medial and lateral malleoli. The relative location of the knee or ankle joint center to the lateral thigh or superior anterior shank marker, respectively, was computed in the static trial, allowing location of joint centers in motion trials without the medial markers. Locations of segment centers of mass were determined from body

segment parameters (de Leva, 1996) and marker coordinates. Linear velocities and accelerations of joint centers and segment centers of mass were determined as the first- and second-time derivatives of position, respectively.

Anatomical reference frames (Figure 8) were defined using unit vectors based on 3D marker coordinates. The **i**, **j**, and **k** unit vectors in the direction of the *x*-, *y*-, and *z*-axes, respectively, were defined as follows: The *x*-axis of the pelvis was defined by the unit vector pointing from the left ASIS to the right ASIS (Equation 1). A temporary *y* unit vector was defined pointing from the left ASIS to the sacrum (Equation 2). The *z*-axis was determined as the cross-product of the temporary and *x*-axis vectors (Equation 3). The *y*-axis was then determined as the cross-product of the *z*- and *x*-axes vectors (Equation 4). The origin of the pelvis reference frame was located midway between the right and left ASIS markers.

$$\mathbf{i}_P = \frac{\mathbf{r}_{LASIS} - \mathbf{r}_{RASIS}}{\left|\mathbf{r}_{LASIS} - \mathbf{r}_{RASIS}\right|} \tag{1}$$

$$\mathbf{a} = \frac{\mathbf{r}_{LASIS} - \mathbf{r}_{Sacrum}}{\left|\mathbf{r}_{LASIS} - \mathbf{r}_{Sacrum}\right|} \tag{2}$$

$$\mathbf{k}_P = \frac{\mathbf{a} \times \mathbf{i}_P}{\left|\mathbf{a} \times \mathbf{i}_P\right|} \tag{3}$$

$$\mathbf{j}_P = \mathbf{k}_P \times \mathbf{i}_P \tag{4}$$

The right thigh reference frame was defined using the body vector of the thigh and the position vectors of the right hip joint center and right lateral thigh. The thigh body vector defined the negative *z*-axis of the thigh; when rotated 180° and unitized, the resultant vector defined the right thigh *z*-axis (Equation 5). A temporary *x* unit vector was

defined from the hip joint center to the lateral thigh marker (Equation 6). The y-axis was determined as the cross-product of the z-axis and temporary vectors (Equation 7). The x-axis was computed as the cross product of the y- and z-axes vectors (Equation 8). The left thigh reference frame was defined in the same manner as the right thigh frame, except that the temporary x-vector was oriented in the negative direction, necessitating a reversal of the cross-product operation to yield an anteriorly oriented y-axis (Equations 9 – 12). The origin of the thigh reference frame was located at the center of mass of the thigh.

$$\mathbf{k}_{RT} = -\frac{\mathbf{r}_{ThighBody}}{\left|\mathbf{r}_{ThighBody}\right|} \tag{5}$$

$$\mathbf{b} = \frac{\mathbf{r}_{RHJC} - \mathbf{r}_{RLateralThigh}}{\left|\mathbf{r}_{RHJC} - \mathbf{r}_{RLateralThigh}\right|} \tag{6}$$

$$\mathbf{j}_{RT} = \frac{\mathbf{k}_{RT} \times \mathbf{b}}{\left|\mathbf{k}_{RT} \times \mathbf{b}\right|} \tag{7}$$

$$\mathbf{i}_{RT} = \mathbf{j}_{RT} \times \mathbf{k}_{RT} \tag{8}$$

$$\mathbf{k}_{LT} = -\frac{\mathbf{r}_{ThighBody}}{\left|\mathbf{r}_{ThighBody}\right|} \tag{9}$$

$$\mathbf{c} = \frac{\mathbf{r}_{LHJC} - \mathbf{r}_{LLateralThigh}}{\left|\mathbf{r}_{LHJC} - \mathbf{r}_{LLateralThigh}\right|} \tag{10}$$

$$\mathbf{j}_{LT} = \frac{\mathbf{c} \times \mathbf{k}_{LT}}{\left|\mathbf{c} \times \mathbf{k}_{LT}\right|} \tag{11}$$

$$\mathbf{i}_{LT} = \mathbf{j}_{LT} \times \mathbf{k}_{LT} \tag{12}$$

The right shank reference frame was defined using the shank body vector and the position vectors of the knee joint center and the superior anterior tibia. The shank body vector defined the negative *z*-axis of the shank; when rotated 180° and unitized, the resultant vector defined the shank *z*-axis (Equation 13). A temporary *y* unit vector was defined from the knee joint center to the superior anterior tibia marker (Equation 14). The *x*-axis was determined as the cross-product of the temporary and *z*-axis vectors (Equation 15). The *y*-axis was computed as the cross product of the *z*- and *x*-axes vectors (Equation 16). The left shank reference frame was defined in the same manner (Equations 17 – 20). The origin of the shank reference frame was located at the center of mass of the shank.

$$\mathbf{k}_{RS} = -\frac{\mathbf{r}_{ShankBody}}{\left|\mathbf{r}_{ShankBody}\right|} \tag{13}$$

$$\mathbf{d} = \frac{\mathbf{r}_{RKJC} - \mathbf{r}_{RSupAntTib}}{\left|\mathbf{r}_{RKJC} - \mathbf{r}_{RSupAntTib}\right|} \tag{14}$$

$$\mathbf{i}_{RS} = \frac{\mathbf{d} \times \mathbf{k}_{RS}}{\left|\mathbf{d} \times \mathbf{k}_{RS}\right|} \tag{15}$$

$$\mathbf{j}_{RS} = \mathbf{k}_{RS} \times \mathbf{i}_{RS} \tag{16}$$

$$\mathbf{k}_{LS} = -\frac{\mathbf{r}_{ShankBody}}{\left|\mathbf{r}_{ShankBody}\right|} \tag{17}$$

$$\mathbf{e} = \frac{\mathbf{r}_{LKJC} - \mathbf{r}_{LSupAntTib}}{\left|\mathbf{r}_{LKJC} - \mathbf{r}_{LSupAntTib}\right|} \tag{18}$$

$$\mathbf{i}_{LS} = \frac{\mathbf{e} \times \mathbf{k}_{LS}}{\left|\mathbf{e} \times \mathbf{k}_{LS}\right|} \tag{19}$$

$$\mathbf{j}_{LS} = \mathbf{k}_{LS} \times \mathbf{i}_{LS} \tag{20}$$

The right foot reference frame was defined using the foot body vector and the position vectors of the ankle joint center and the calcaneus. The foot body vector defined the negative z-axis of the foot; when rotated 180° and unitized, the resultant vector defined the foot z-axis (Equation 21). A temporary y unit vector was defined from the heel marker to the ankle joint center (Equation 22). The x-axis was determined as the cross-product of the temporary and z-axis vectors (Equation 23). The y-axis was computed as the cross product of the z- and x-axes vectors (Equation 24). The left foot reference frame was defined in the same manner (Equations 25 – 28). The origin of the foot reference frame was located at the center of mass of the foot.

$$\mathbf{k}_{RF} = -\frac{\mathbf{r}_{FootBody}}{\left|\mathbf{r}_{FootBody}\right|} \tag{21}$$

$$\mathbf{f} = \frac{\mathbf{r}_{RCalcaneus} - \mathbf{r}_{RAJC}}{\left|\mathbf{r}_{RCalcaneus} - \mathbf{r}_{RAJC}\right|} \tag{22}$$

$$\mathbf{i}_{RF} = \frac{\mathbf{f} \times \mathbf{k}_{RF}}{\left|\mathbf{f} \times \mathbf{k}_{RF}\right|} \tag{23}$$

$$\mathbf{j}_{RF} = \mathbf{k}_{RF} \times \mathbf{i}_{RF} \tag{24}$$

$$\mathbf{k}_{LF} = -\frac{\mathbf{r}_{FootBody}}{\left|\mathbf{r}_{FootBody}\right|} \tag{25}$$

$$\mathbf{g} = \frac{\mathbf{r}_{LCalcaneus} - \mathbf{r}_{LAJC}}{\left|\mathbf{r}_{LCalcaneus} - \mathbf{r}_{LAJC}\right|} \tag{26}$$

$$\mathbf{i}_{LF} = \frac{\mathbf{g} \times \mathbf{k}_{LF}}{\left|\mathbf{g} \times \mathbf{k}_{LF}\right|} \tag{27}$$

$$\mathbf{j}_{LF} = \mathbf{k}_{LF} \times \mathbf{i}_{LF} \tag{28}$$

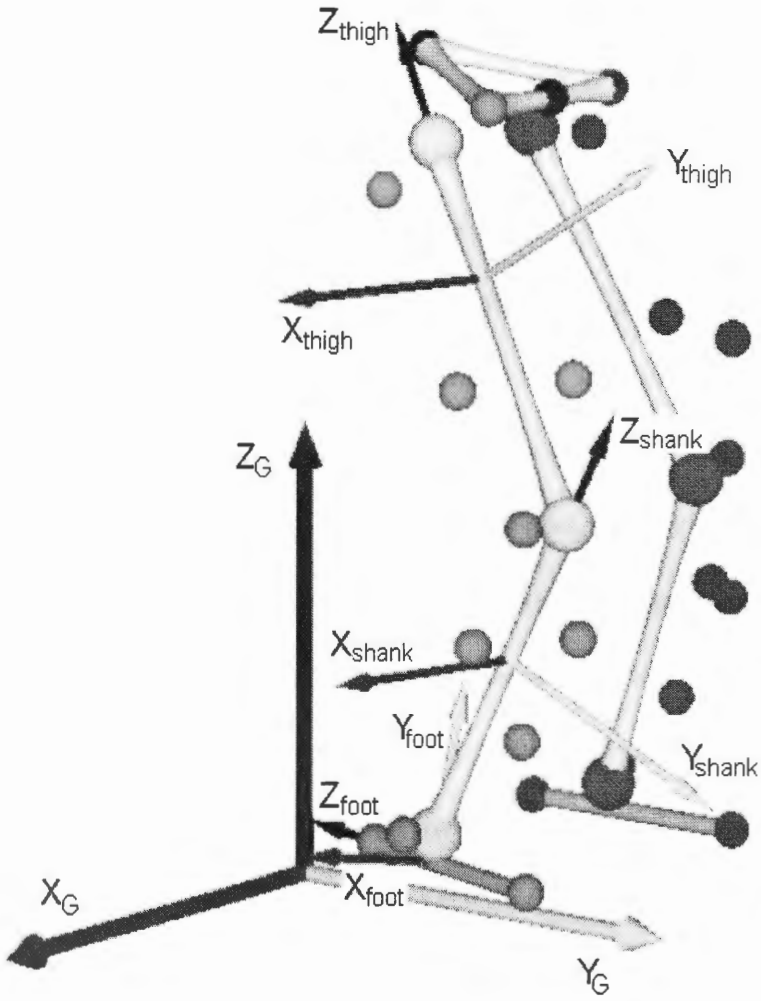

Figure 8. Global ($X_G Y_G Z_G$) and right leg anatomical reference frames ($X_{thigh} Y_{thigh} Z_{thigh}$, $X_{shank} Y_{shank} Z_{shank}$, $X_{foot} Y_{foot} Z_{foot}$). Anatomical reference frames were embedded at segment centers of mass, with X-axes directed to the right, Z-axes directed superiorly along the long axis of the segment, and Y-axes perpendicular to X- and Z-axes. Left leg reference frames were defined in the same manner. Large spheres represent internal computed joint centers; small spheres represent external markers affixed to participants' skin or clothing.

In order to express the position of a given segment in different reference frames, transformation matrices were constructed from the x-, y-, and z-components of the unit vectors for each segment. Orientation angles of each segment in each plane were then

computed from the transformation matrices. An XYZ Cardanian rotation sequence was adopted to provide anatomical relevance to the angles. In this schema, rotations about the X axis described joint flexion/extension; rotations about the Y axis joint abduction/ adduction; rotations about the Z axis described internal/external rotation (Kwon, 1998).

Resultant Joint Forces and Moments

Linear and angular momentum of each segment were computed from body segment parameters (de Leva, 1996) and linear and angular velocities of the segments in each frame. Segment linear momenta were computed as the product of segment mass and linear velocity. Angular momenta were computed using the procedures outlined by Dapena (1978). Changes of momentum from frame-to-frame were determined to compute net joint forces and moments. Net joint forces and moments at the ankle (Figure 9) and the knee (Figure 10) during the period of foot contact with the force plates were computed using inverse dynamics (McNitt-Gray, 1993).

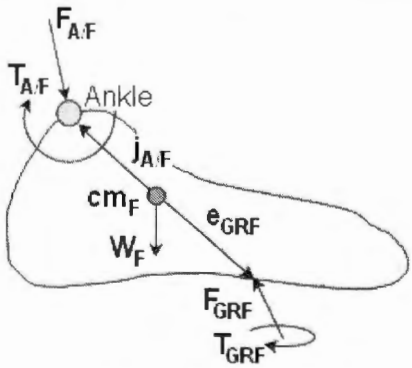

Figure 9. Free body diagram of the foot, where cm_F = foot center of mass, $\mathbf{W}_F$ = weight of foot, $\mathbf{e}_{GRF}$ = position vector from cm_F to ground reaction force center of pressure, $\mathbf{j}_{A/F}$ = position vector from cm_F to ankle joint center, $\mathbf{F}_{GRF}$ = ground reaction force, $\mathbf{T}_{GRF}$ = moment due to ground reaction force, $\mathbf{F}_{A/F}$ = net joint force acting on the foot at the ankle, and $\mathbf{T}_{A/F}$ = net joint moment acting at the ankle due to muscle forces.

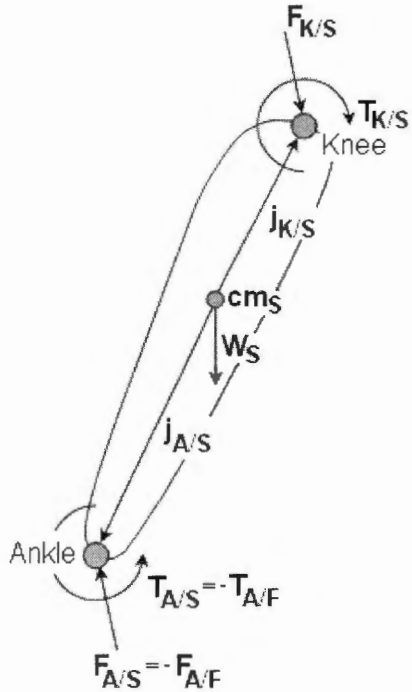

Figure 10. Free body diagram of the shank, where cm_S = shank center of mass, $\mathbf{W}_S$ = weight of shank, $\mathbf{j}_{A/S}$ = position vector from cm_S to ankle joint center, $\mathbf{F}_{A/S}$ = net joint force acting on the shank at the ankle, $\mathbf{T}_{A/S}$ = net joint moment acting at the ankle due to muscle forces, $\mathbf{j}_{K/S}$ = position vector from cm_S to knee joint center, $\mathbf{F}_{K/S}$ = net joint force acting on the shank at the knee, and $\mathbf{T}_{K/S}$ = net joint moment acting at the knee due to muscle forces.

CHAPTER IV

RESULTS AND DISCUSSION

Deep Squat Kinematics

There was a significant multivariate effect for bracing condition during the deep squat ($F_{(4,28)} = 3.500, p = .019$). One-way ANOVA revealed significant differences among ankle bracing conditions in both knee flexion ($F_{(2, 14)} = 10.501, p = .002$) and ankle dorsiflexion ($F_{(2, 14)} = 11.027, p = .001$). Pairwise comparisons revealed a significant difference in maximum knee flexion between both the taped ($p = .027$) and braced ($p = .035$) conditions compared with the untaped condition and a significant difference in maximum ankle dorsiflexion between the taped and untaped conditions ($p = .013$). There was no difference ($p = .058$) in maximum ankle dorsiflexion between the braced and untaped conditions (Table 6). The taped condition imposed the greatest restriction on range of motion, followed by the braced and untaped conditions. Plots of knee and ankle orientation angles during the deep squat are presented in Figure 11.

Table 6

Univariate Analysis of Deep Squat Kinematics

	Bracing Conditions			Statistics		Pairwise Comparisons		
	Untaped (U)	Taped (T)	Braced (B)	*F*	*p*	U-T	U-B	T-B
Maximum Knee Flexion (°)	110.2 (17.3)	93.5 (20.0)	102.0 (18.3)	10.501	.002	*	*	
Maximum Ankle Dorsiflexion (°)	27.2 (4.3)	19.6 (3.7)	24.1 (5.9)	11.027	.001	*		

Values reported as Mean (*SD*).

*Indicates significance of pairwise comparisons ($p < .05$)

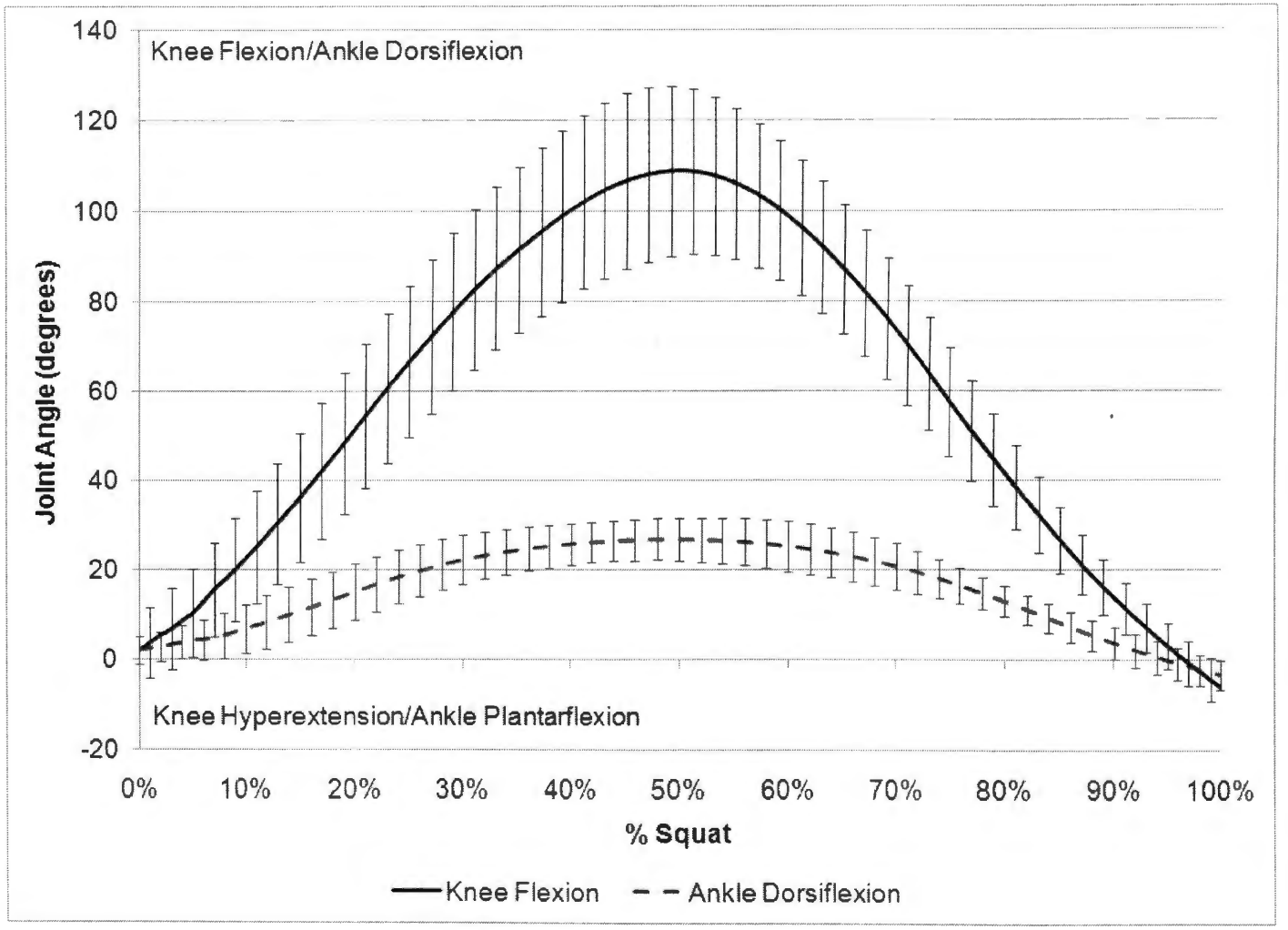

Figure 11. Composite average (±1 SD) joint angles vs. duration of squat during the deep squat in the untaped condition. Patterns of motion were similar in the taped and braced conditions.

Jump Landing Kinematics

There was no multivariate difference in kinematic variables among bracing conditions ($F_{(14,18)} = 1.181, p = .364$). There were also no interactions among bracing and side conditions ($F_{(28, 100)} = .808, p = .735$). There was a significant multivariate difference ($F_{(14,18)} = 3.540, p = .007$) among the side conditions, but this difference did not pertain to the purpose of this study. Plots of knee flexion, varus/valgus, and rotation angles during the drop jump are presented in Figures 12, 13, & 14.

Jump Landing Kinetics

There was no multivariate difference in kinetic variables among bracing conditions ($F_{(14,18)} = .867$, $p = .601$). There were also no interactions among bracing and side conditions ($F_{(28, 100)} = 1.083$, $p = .374$). There was a significant multivariate difference ($F_{(14,18)} = 5.271$, $p = .001$) among the side conditions, but this difference did not pertain to the purpose of this study. Plots of knee extension, varus/valgus, and rotation moments and resultant ground reaction forces are presented in Figures 15 through 18.

Table 7

Jump Landing Kinematics – Descriptive Statistics

Variable	Side	Untaped	Taped	Braced
Knee Flexion at Foot Strike (°)	Upright	26.2 (8.0)	29.0 (4.8)	27.4 (6.7)
	Contralateral	19.7 (3.8)	23.2 (4.2)	23.0 (5.1)
	Ipsilateral	33.2 (8.9)	31.0 (5.6)	31.2 (7.6)
Knee Valgus[a] at Foot Strike (°)	Upright	-0.1 (3.7)	-1.7 (1.9)	-0.8 (1.9)
	Contralateral	-3.9 (4.3)	-4.8 (3.6)	-4.8 (2.9)
	Ipsilateral	-1.0 (4.4)	-0.5 (3.4)	0.1 (2.8)
Knee Rotation[b] at Foot Strike (°)	Upright	-3.5 (7.8)	-3.6 (7.4)	-4.2 (7.9)
	Contralateral	-1.1 (11.4)	-3.5 (7.8)	-0.5 (7.6)
	Ipsilateral	-3.3 (10.3)	-3.2 (7.1)	-1.6 (11.2)
Maximum Knee Flexion (°)	Upright	75.5 (12.0)	76.8 (13.2)	78.1 (11.9)
	Contralateral	58.1 (9.8)	59.2 (6.4)	59.8 (7.9)
	Ipsilateral	78.1 (9.6)	76.0 (8.6)	77.7 (10.2)

Table 7 (continued)

Jump Landing Kinematics – Descriptive Statistics

Variable	Side	Untaped	Taped	Braced
Maximum Knee Valgus[a] (°)	Upright	-11.1 (9.6)	-16.2 (5.0)	-13.8 (9.0)
	Contralateral	-7.3 (10.1)	-8.5 (4.3)	-8.9 (6.3)
	Ipsilateral	-10.0 (14.6)	-11.1 (12.1)	-12.6 (14.2)
Maximum Knee Rotation[b](°)	Upright	0.6 (15.3)	1.5 (16.5)	2.4 (13.1)
	Contralateral	1.3 (19.6)	-4.7 (12.1)	3.0 (14.9)
	Ipsilateral	0.7 (12.8)	-2.4 (13.3).	5.3 (14.6)
Tibial Translation (cm)	Upright	-0.0044 (0.0078)	-0.0028 (0.0109)	-0.0046 (0.0114)
	Contralateral	0.0032 (0.0068)	0.0039 (0.0077)	0.0039 (0.0039)
	Ipsilateral	-0.0059 (0.0079)	-0.0066 (0.0069)	-0.0069 (0.0090)

Values reported as Mean (*SD*).

[a]Positive values are varus angles, negative values are valgus angles. [b]Positive values are internal tibial rotation, negative values are external tibial rotation.

Table 8

Jump Landing Kinetics – Descriptive Statistics

Variable	Side	Untaped	Taped	Braced
Knee extension moment at foot strike (%BW×Ht)	Upright	1.7 (1.2)	2.9 (0.9)	2.2 (1.4)
	Contralateral	2.0 (0.5)	2.4 (1.2)	2.4 (1.1)
	Ipsilateral	2.2 (0.8)	2.9 (1.0)	2.7 (0.7)
Knee varus moment at foot strike (%BW×Ht)	Upright	0.6 (0.4)	0.7 (0.4)	0.6 (0.4)
	Contralateral	1.2 (0.7)	1.5 (0.7)	1.3 (0.4)
	Ipsilateral	0.6 (0.7)	0.5 (0.8)	0.3 (0.9)
Knee rotation moment at foot strike[a] (%BW×Ht)	Upright	-0.2 (0.2)	-0.2 (0.2)	-0.2 (0.2)
	Contralateral	-0.3 (0.2)	-0.3 (0.3)	-0.3 (0.1)
	Ipsilateral	-0.4 (0.3)	-0.2 (0.3)	-0.3 (0.3)
Maximum knee extension moment (%BW×Ht)	Upright	9.5 (1.5)	9.9 (1.3)	9.3 (1.6)
	Contralateral	11.8 (2.0)	12.3 (1.7)	12.5 (2.4)
	Ipsilateral	7.7 (1.9)	8.2 (2.1)	7.8 (2.9)

Table 8 (continued)

Jump Landing Kinetics – Descriptive Statistics

Variable	Side	Untaped	Taped	Braced
Maximum knee varus moment (%BW×Ht)	Upright	2.6 (1.8)	2.5 (2.7)	2.8 (1.5)
	Contralateral	2.3 (2.6)	2.5 (1.4)	2.2 (2.2)
	Ipsilateral	1.7 (3.4)	2.6 (2.7)	2.3 (3.0)
Maximum knee rotation moment (%BW×Ht)	Upright	-0.4 (1.0)	-0.3 (1.0)	-0.6 (0.7)
	Contralateral	-0.1 (1.1)	0.3 (1.2)	0.1 (1.3)
	Ipsilateral	-0.6 (1.8)	-0.4 (1.3)	-0.6 (1.2)
Maximum ground reaction force (%BW)	Upright	205.0 (48.7)	218.5 (36.8)	210.9 (41.7)
	Contralateral	321.2 (69.6)	324.3 (77.6)	309.9 (64.9)
	Ipsilateral	145.0 (25.1)	158.4 (21.9)	163.5 (31.3)

Values reported as Mean (*SD*).

[a]Positive values are internal tibial rotation moments, negative values are external tibial rotation moments.

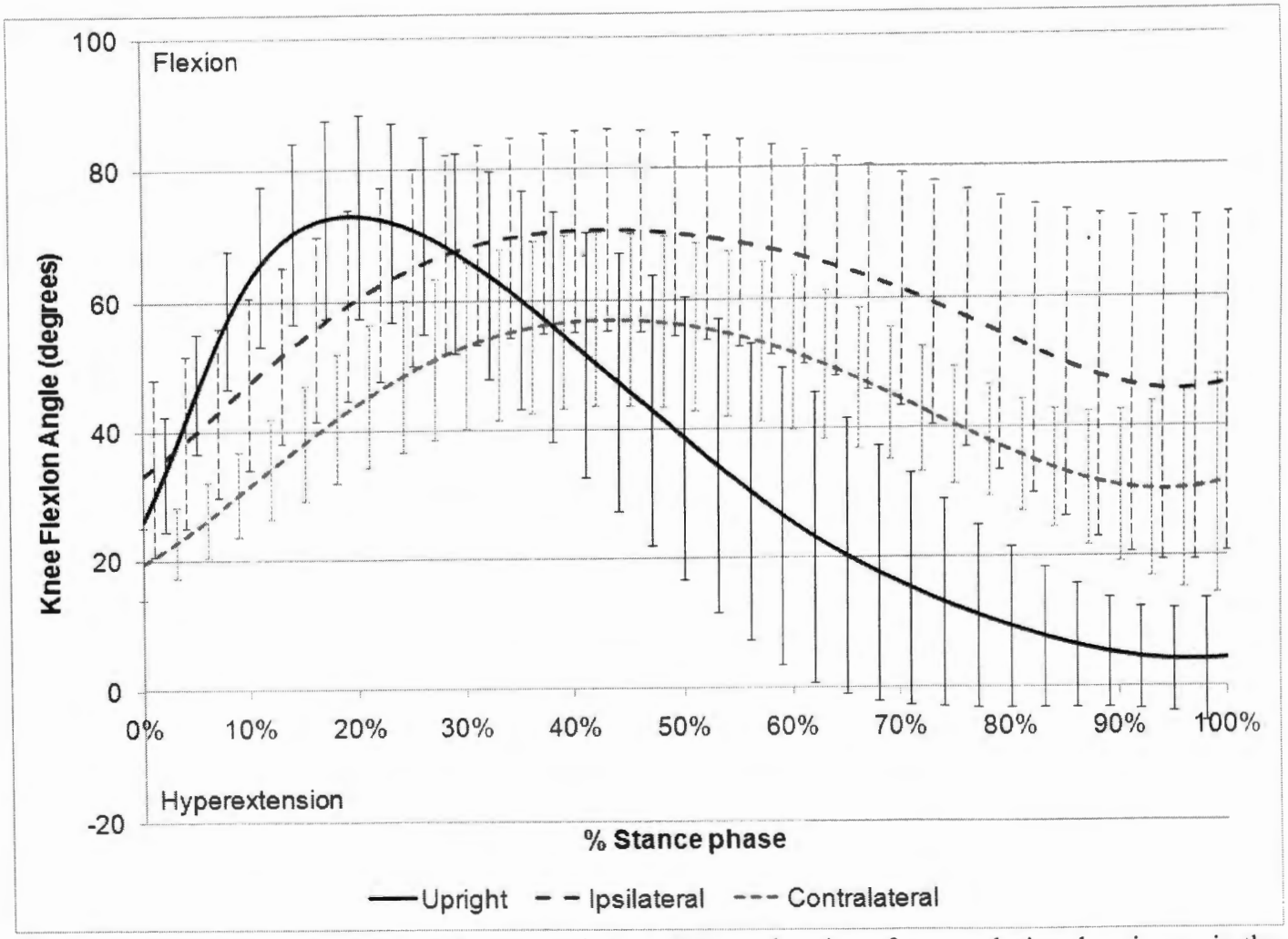

Figure 12. Composite average knee flexion angle (± 1 SD) vs. duration of stance during drop jumps in the untaped condition. Patterns of motion were similar in the taped and braced conditions.

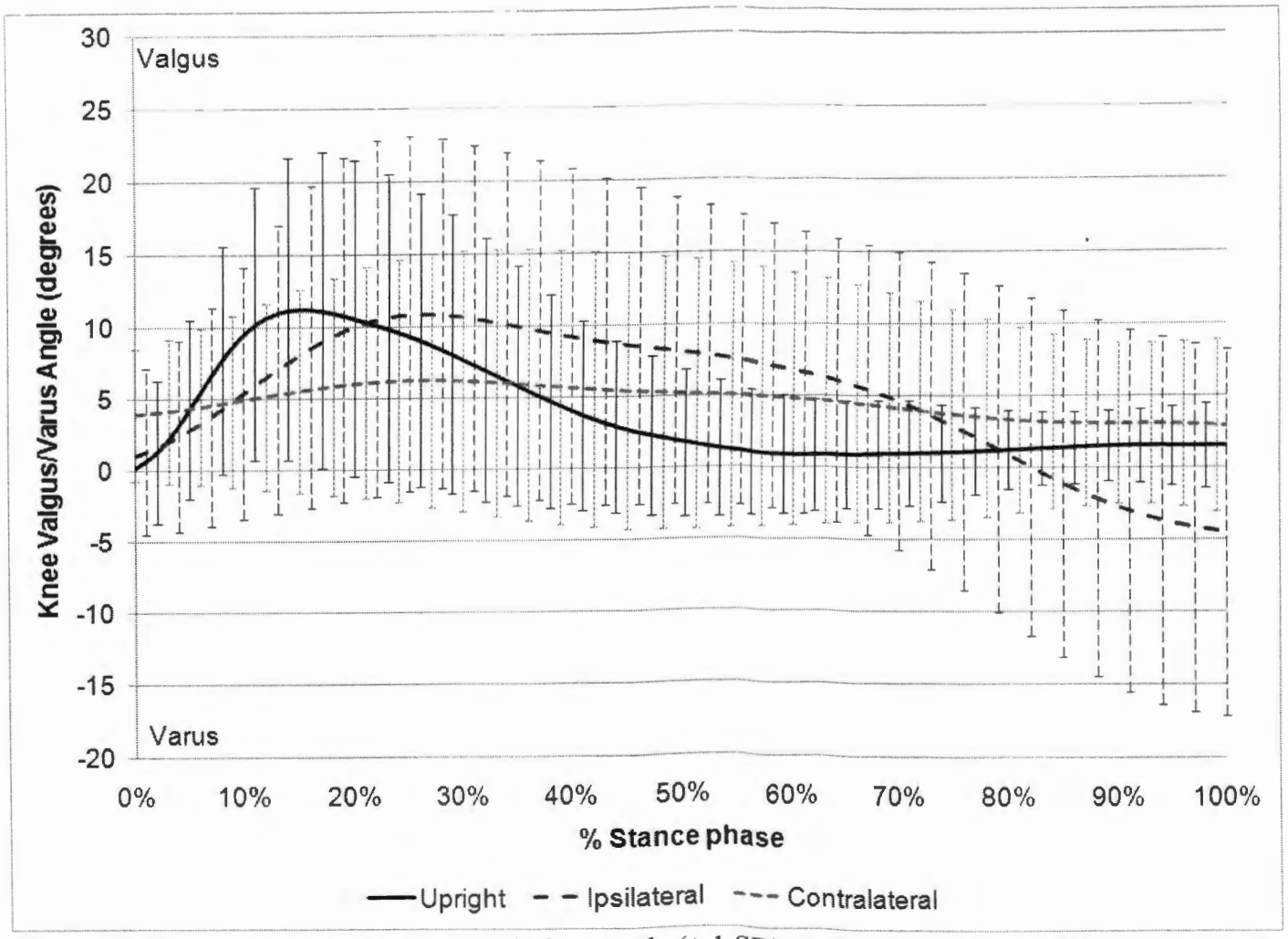

Figure 13. Composite average knee varus/valgus angle (± 1 SD) vs. duration of stance during drop jumps in the untaped condition. Patterns of motion were similar in the taped and braced conditions.

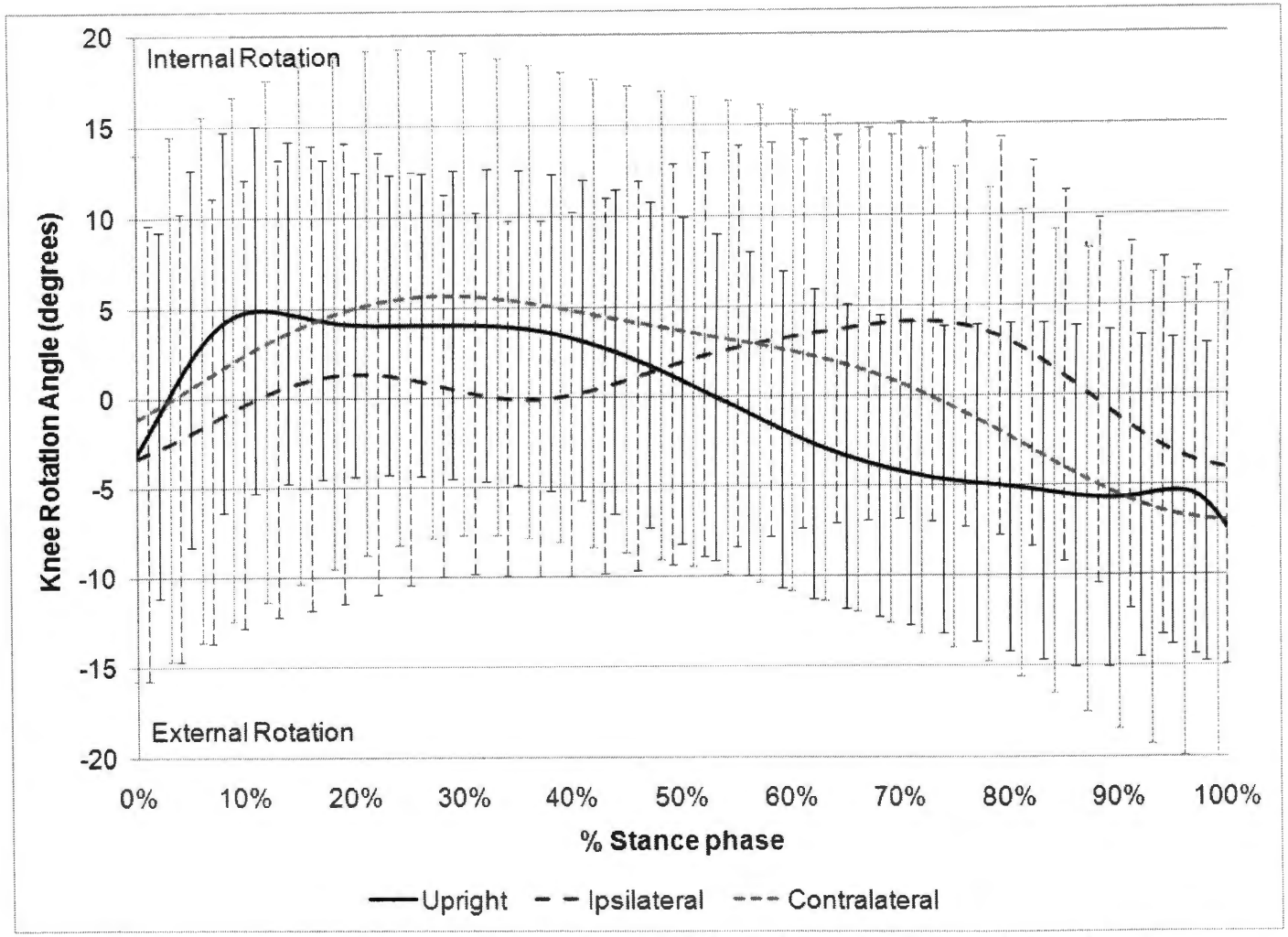

Figure 14. Composite average knee rotation angle (± 1 SD) vs. duration of stance during drop jumps in the untaped condition. Patterns of motion were similar in the taped and braced conditions

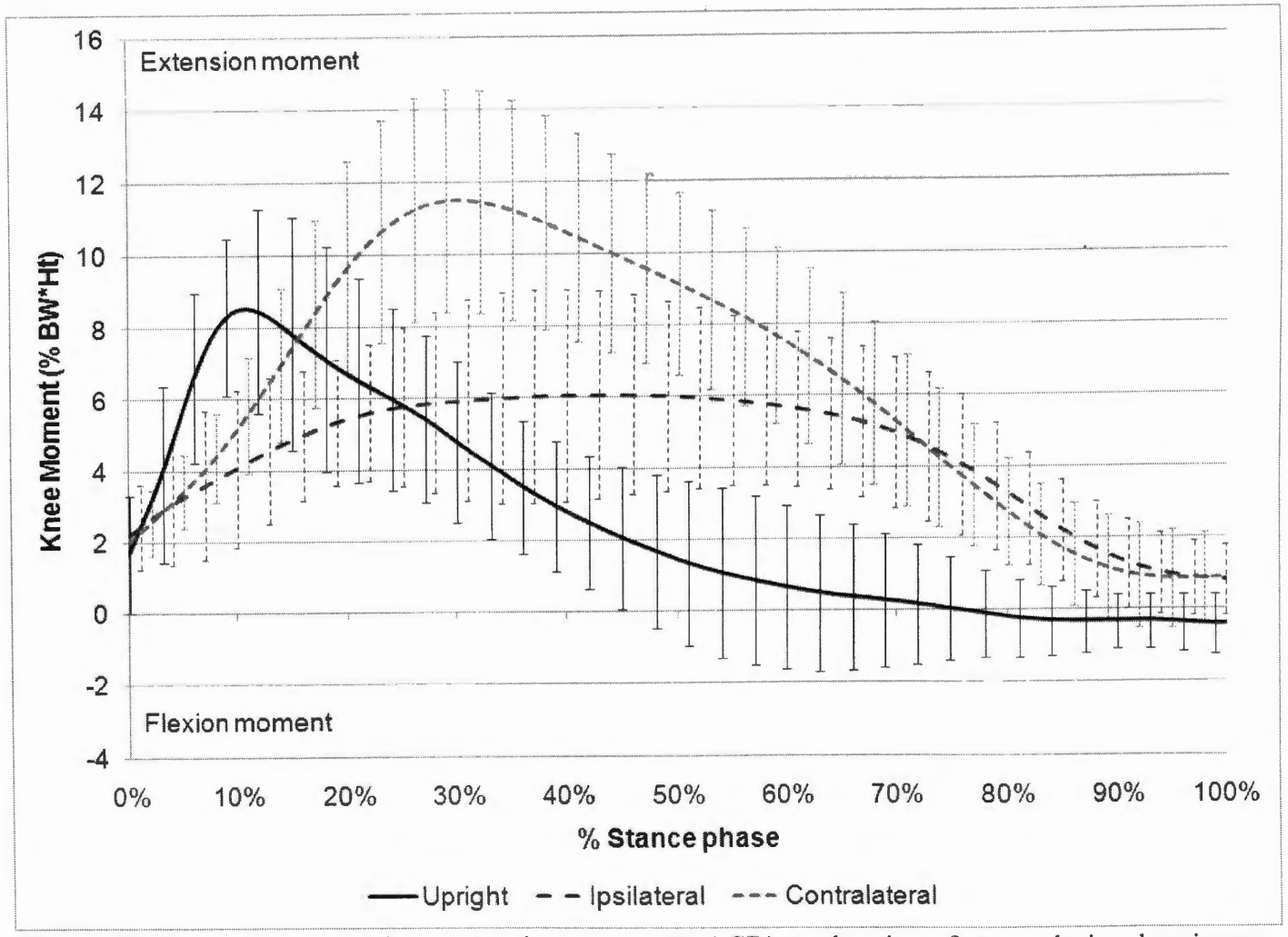

Figure 15. Composite average knee extension moment (± 1 SD) vs. duration of stance during drop jumps in the untaped condition. Moment patterns were similar in the taped and braced conditions.

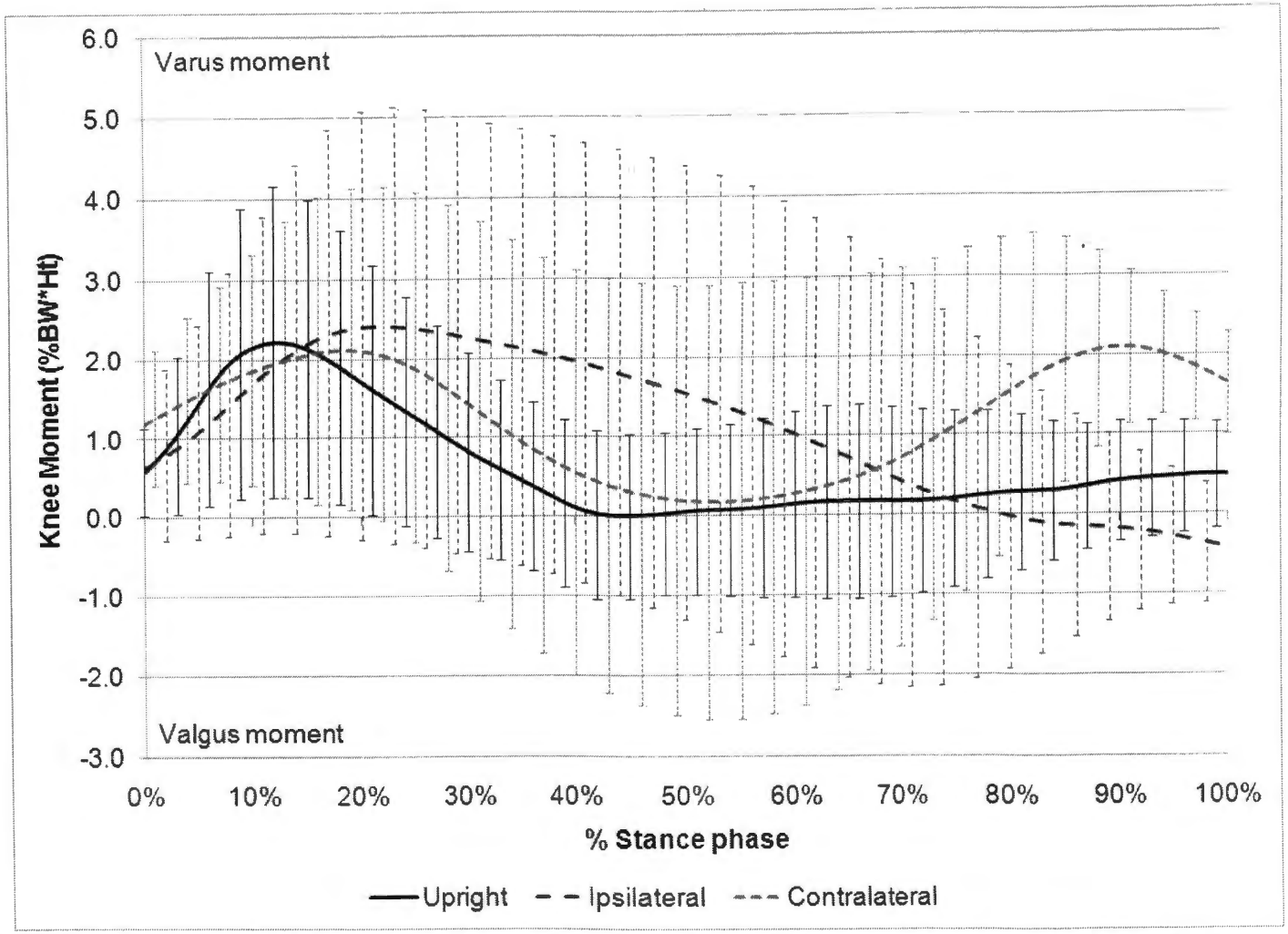

Figure 16. Composite average knee varus/valgus moment (± 1 SD) vs. duration of stance during drop jumps in the untaped condition. Moment patterns were similar in the taped and braced conditions.

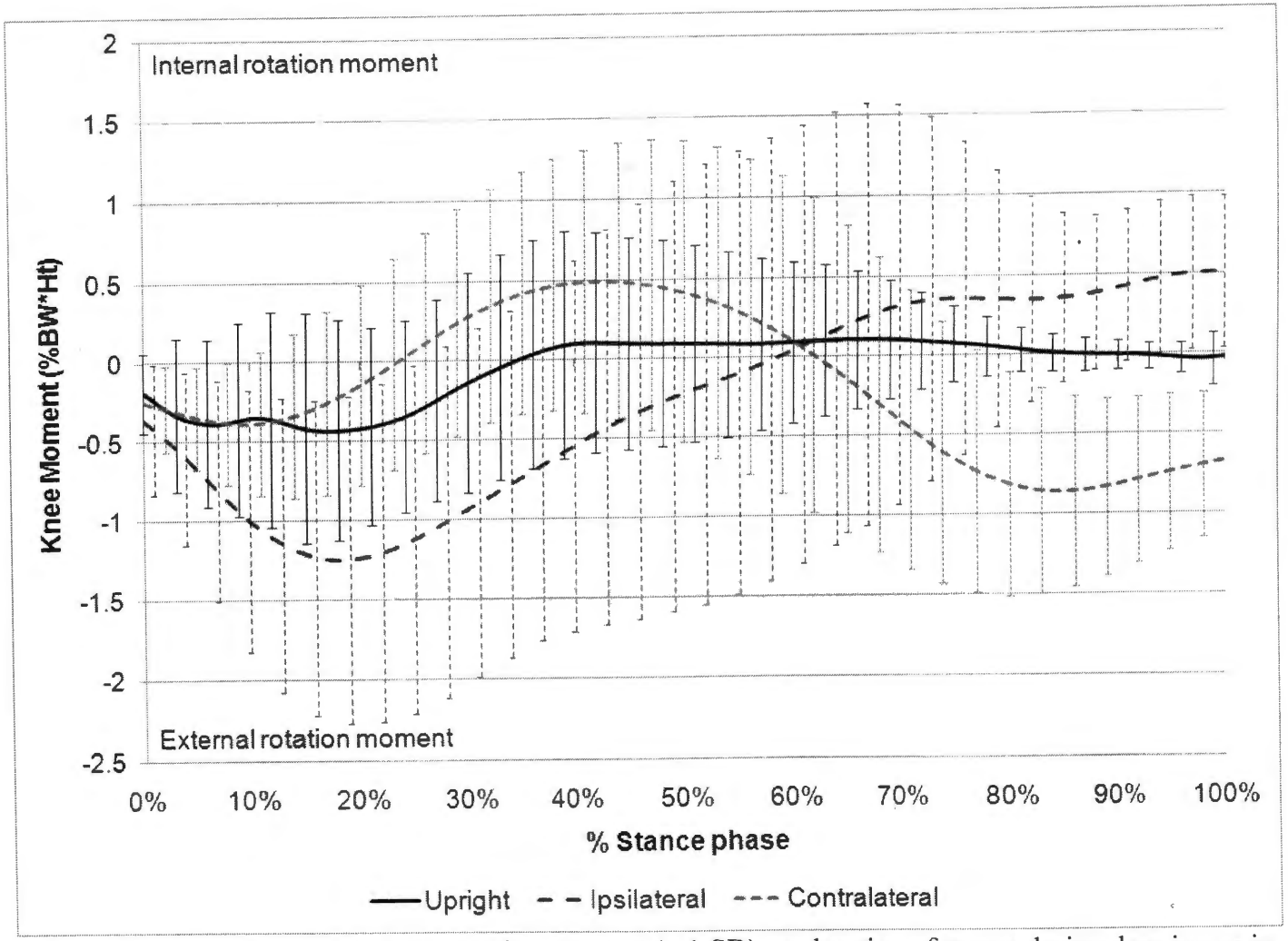

Figure 17. Composite average knee rotation moment (± 1 SD) vs. duration of stance during drop jumps in the untaped condition. Moment patterns were similar in the taped and braced conditions.

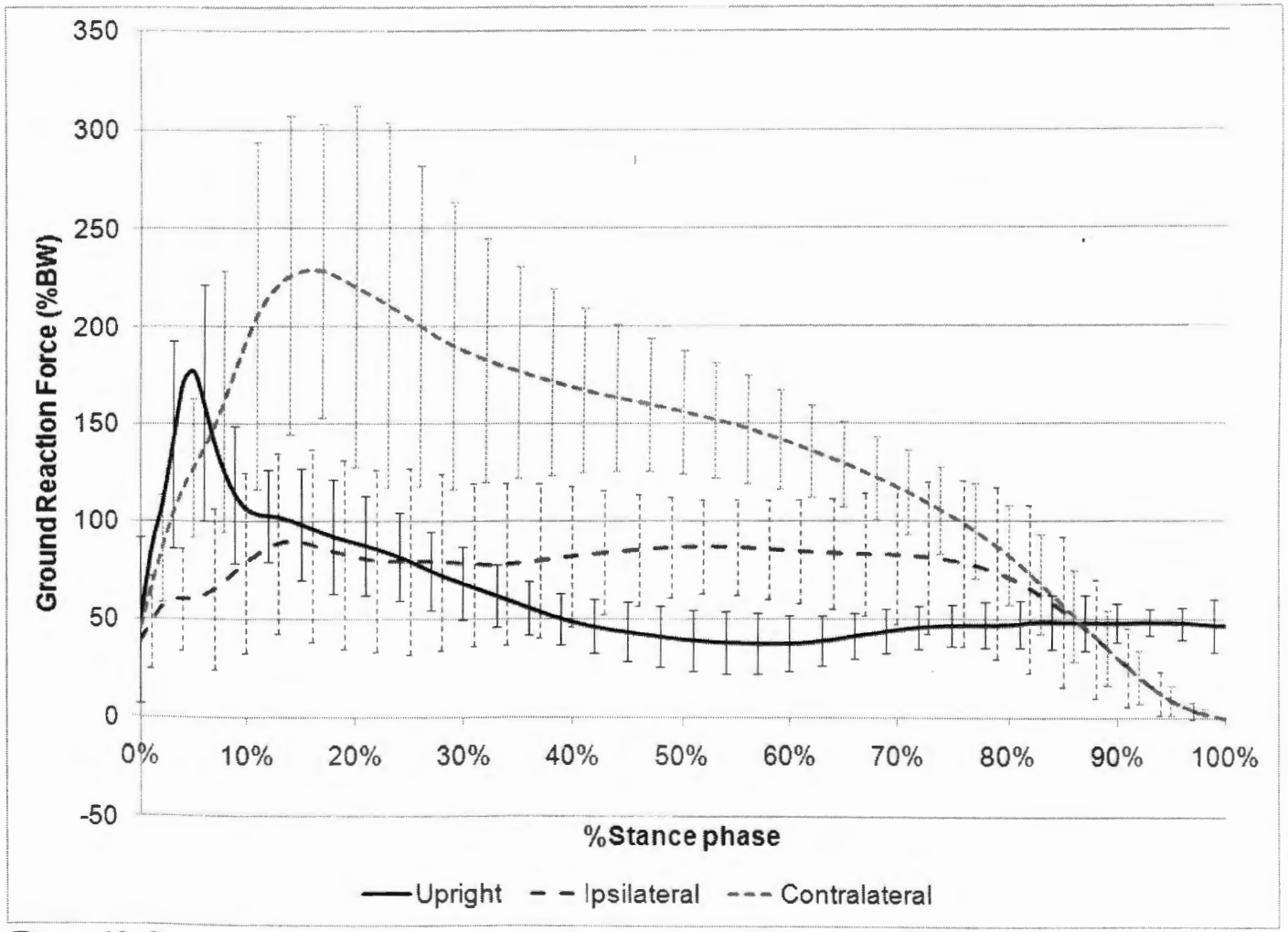

Figure 18. Composite average resultant ground reaction force (± 1 SD) vs. duration of stance during drop jumps in the untaped condition. Force patterns were similar in the taped and braced conditions.

Discussion

The purposes of this study were to determine if ankle bracing had an effect on knee motion during a deep squat and during reactive maneuvers that are associated with increased risk of ACL injury. Deep squat data indicated that ankle bracing/taping affected the sagittal plane range of motion of both the ankle and the knee. The greatest restriction of range of motion due to taping and bracing during the deep squat occurred at the ankle, as expected. There was a lesser effect on range of motion at the knee. The taped condition restricted range of motion by 15.2% and 27.9% at the knee and ankle, respectively, as compared to the untaped condition. The braced condition restricted motion by 7.4% and 11.4% at the knee and ankle, respectively, as compared to the untaped condition.

Data from drop jumps, which are more ballistic in nature than the deep squats, suggest ankle bracing does not affect sagittal plane knee kinematics during reactive maneuvers that are known to stress the ACL. During the drop landings knee flexion angles did not closely approach the range of motion limitations imposed by the bracing/taping. In the most restrictive bracing condition (taped), there was a 23° difference between the average maximum knee flexion angles during the deep squat and that experienced during the drop jumps. One potential explanation for these findings is that, when the range of motion of a particular activity is less than the range of motion restriction imposed by bracing or taping, the bracing or taping does not alter the kinematics of the activity.

No differences in frontal or transverse plane kinematics at the knee were observed among bracing conditions. In contrast to the sagittal plane kinematics, frontal and

transverse plane measures were much more variable. Due to their small ranges of motion, transverse and frontal plane knee angles are much more sensitive to marker placement and skin motion artifact errors (Cappozzo, Catani, Leardini, Benedetti, & Della Croce, 1996; Lucchetti, Cappozzo, Cappello, & Della Croce, 1998; Ramsey & Wretenberg, 1999; Schache, Baker, & Lamoreux, 2006), and typically require a corrective procedure or optimization process to minimize variability. No such procedures were incorporated for this study.

Only three kinematic studies were located which closely approximated the protocol used in this study (Decker et al., 2003; Fagenbaum & Darling, 2003; Ford et al., 2003). Decker et al. and Fagenbaum & Darling only evaluated sagittal plane kinematics, while Ford et al. only evaluated frontal plane kinematics. Additionally, Fagenbaum & Darling investigated single-leg landing strategies, while this and the other studies examined two-leg landings. Data from only the unbraced-upright condition combination from this study were compared, as these were the only similar conditions between the present and the other studies. Comparisons of kinematic data from these and the present study are detailed in Table 9.

Table 9

Comparison of Kinematic Variables Among Several Studies

Variable	Present study[a]	Ford et al. (2003)	Decker et al. (2003)	Fagenbaum & Darling (2003)[b, c]	
Female participants (n)	8	47	9	8	
Landing height (cm)	39.9 (5.5)	31[d]	60[d]	25.4[d]	50.8[d]
Knee flexion at foot strike (°)	26.2 (9.6)	NR	22.8 (8.0)	~35	~29
Maximum knee flexion (°)	75.5 (12.2)	NR	98.4 (10.6)	~52	~62
Knee valgus at foot strike (°)	0.1 (4.7)	5.9[e]	NR	NR	NR
Maximum knee valgus (°)	11.1 (12.4)	27.6[e]	NR	NR	NR

Values reported as Mean (*SD*). NR = Not reported.

[a]Only data from the untaped, upright condition in this study were compared to the other studies. [b]Single dominant leg landings, non-fatigued condition. [c]Knee angles estimated from graphs in literature. [d]Landing height was the same for all participants. [e]*SD* not reported.

Although there was no main effect of bracing on kinetic variables, a notable but non-significant change due to bracing was observed in the knee extension moment at foot strike. Knee extension moments at foot strike from all trials in a given bracing condition were averaged and compared. The mean knee extension moment at foot strike in the taped condition was 42.3% greater than in the untaped condition and 10.5% greater than in the braced condition. The braced condition extension moment was 27.4% greater than

in the untaped condition. This is surprising, in that the kinematic differences among bracing conditions were insignificant. This also suggests that, at least in the sagittal plane, while restriction of ankle motion has no kinematic effect, there can be a pronounced kinetic effect of ankle bracing at the knee. A likely explanation is that bracing and taping restricted subtalar joint motion so the normal shock absorption occurring at that joint was restricted or eliminated. With shock absorption attenuated at the ankle, the next available joint at which shock absorption can occur is at the knee. The greatest knee extension moment would be expected to occur in the bracing condition with the least amount of subtalar motion, as seen in this study. The participants performed drop jump trials immediately after deep squats with no other warm-up, so the kinetic effect due to loosening of taping/bracing over time was not assessed. This effect would be expected to decrease with time after application.

Only two kinetic studies were located which closely approximated the protocol used in this study (Decker et al., 2003; Schot et al., 1994). These studies only reported sagittal plane kinetics, so no comparisons of frontal or transverse plane kinetics can be made. The drop height used by Decker et al. and Schot et al. was 60 cm. For comparison with the present study, knee moments reported in those studies were scaled to a drop height of 39.9 cm by computing the drop time for 60 cm and 39.9 cm and multiplying by the drop-time ratio. Decker et al. reported knee moments in percent body weight × body height (%BW×Ht). Schot et al. reported moments in Nm/kg body mass; for comparison purposes, their values were converted to %BW×Ht using reported participant mass and height data. The knee extension moments determined in this study are similar to, but

smaller than, the scaled moments from Decker et al. and Schot et al., and are considered reasonable values for the untaped-upright condition. A comparison of maximum knee extension moment data from these studies are presented in Table 10.

Table 10

Comparison of Maximum Knee Extension Moments Among Several Studies

Variable	Present study[a]	Decker et al. (2003)	Schot et al. (1994)
Female participants (n)	8	9	5
Landing height (cm)	39.9 (5.5)	60[b]	60[b]
Maximum knee extension moment (%BW×Ht)	9.54 (1.86)	11.90 (1.64)[c]	10.85 (3.95)[c, d]

Values reported as Mean (*SD*).

[a]Only data from the untaped, upright condition in this study were compared to the other studies. [b]Landing height was the same for all participants. [c]Values reported in literature were scaled to 39.9 cm drop height. [d]Values were estimated from graphs and converted to %BW×Ht.

During analysis of drop-jump trials, the only kinetic differences noted were in the side condition. While analysis of the side condition was not an original focus of this research, the differences noted were determined to be interesting and bear discussion. When movement was to the right or left, the contralateral limb experienced a 35.3% greater maximum knee extension moment and a 51.1% greater maximum resultant

ground reaction force than the ipsilateral limb. The pattern was expected, as the contralateral limb functions to push-off from the ground in the direction of intended movement. The magnitude of the differences, however, is surprising and supports the noted pattern in ACL injuries occurring as a result of an abrupt direction change. In addition, the contralateral knee experienced a 64.7% greater varus moment at foot strike than the ipsilateral limb. Also noted were kinematic differences, with the contralateral knee having 45.2% less flexion at foot strike, 89.1% greater valgus at foot strike, and 31% less maximum knee flexion than the ipsilateral knee. The differences in varus moment, flexion, and valgus angles at foot strike seem to indicate some anticipation of the direction of motion.

The direction of intended movement was given at the beginning of the drop jump trials along with a "go" signal. It was not the investigator's intention to conceal the direction of intended movement until right before the landing. In practice, the command to move and the receipt of the visual cue were practically simultaneous. The imposition of a time delay in presenting the visual cue would eliminate the participant's ability to know the direction of movement before jumping, making the task truly reactive.

Also observed during data collection was the tendency of some participants to walk through the motion during drop jump trials with motion to the side. This was countered by having them repeat the movement in a more dynamic fashion. A suggested improvement in the study design would be to impose a maximum effort vertical or lateral jump following the landing of the drop jump.

In order to improve the accuracy of frontal and transverse plane kinematics, computation of helical axes of the knee during a squat trial and the use of the average helical axis to define/align the anatomical axes of the thigh and shank should be implemented. Alternatively, an optimization or correction process could be implemented to minimize variability in the kinematics. This would, in turn, reduce variability in the kinetics computed through inverse dynamics.

CHAPTER V

CONCLUSIONS

Study Purposes

The purposes of this study were to investigate the effects of ankle bracing on (a) the sagittal plane kinematics of the ankle and knee during deep squats and (b) the kinematics and kinetics of the knee during jump landings. The ankle-bracing conditions that were investigated were: untaped, basketweave athletic taping, and Webly® Ankle Orthosis.

Null Hypotheses

H_01: *There will be no differences in sagittal plane kinematics of the knee or ankle during a deep squat among bracing conditions.* There was a significant multivariate main effect for bracing condition during the deep squat ($F_{(4,28)} = 3.500, p = .019$). One-way ANOVA revealed significant differences among ankle bracing conditions in both knee flexion ($F_{(2, 14)} = 10.501, p = .002$) and ankle dorsiflexion ($F_{(2, 14)} = 11.027, p = .001$). Pairwise comparisons revealed a significant difference in maximum knee flexion between both the taped ($p = .027$) and braced ($p = .035$) conditions compared with the untaped condition and a significant difference in maximum ankle dorsiflexion between the taped and untaped conditions ($p = .013$). There was no difference ($p = .058$) in maximum ankle dorsiflexion between the braced and untaped conditions. The taped

condition provided the greatest restriction on range of motion at both joints, followed by the braced and untaped conditions. *This hypothesis is rejected.*

$H_0$2: *There will be no differences in knee kinematics during jump landings among bracing conditions.* There were no differences in kinematic dependent variables among bracing conditions ($F_{(14,18)} = 1.181$, $p = .364$). *This hypothesis is accepted.*

$H_0$3: *There will be no differences in knee kinetics during jump landings among bracing conditions.* There were no differences in kinetic dependent variables among bracing conditions ($F_{(14,18)} = .867$, $p = .601$). *This hypothesis is accepted.*

Conclusions

Evidence from the deep squats indicates that ankle bracing and taping affect the sagittal plane range of motion of the knee as well as the ankle. This restriction could possibly affect the injury potential of the knee. Drop jump data, however, suggest ankle bracing does not have kinematic or kinetic effects at the knee during reactive maneuvers that are known to stress the ACL. This is most likely because, during the drop landings, the knee flexion angles did not approach the range of motion limitations imposed by the bracing/taping. Bracing may affect sagittal plane knee kinetics at foot strike, although the effect is not significant. Factors affecting the inability to detect differences include small sample size, lack of time delay for the visual cue, and high variability in valgus and rotation angles of the knee.

Suggestions for Future Research

While there were no appreciable differences in the effects of ankle bracing on knee motion in this study, more tightly controlled studies with more participants may be able to reveal differences. Changes that should be implemented include:

1. The computation of helical axes of the knee during a squat trial and the use of the average helical axis to define/align the anatomical axes of the thigh and shank.
2. Test more types of ankle braces under conditions similar to this study.
3. Impose a "maximum effort" movement (vertical and lateral) task upon landing from the drop jump.
4. Impose a time delay in the presentation of the movement signal in order to make the post-drop movement truly reactive.
5. Perform the study with more participants to improve statistical power.
6. Standardize shoes for all participants.

Additional studies that could be conducted include:

1. A controlled prospective randomized study to track ACL injury occurrence under different ankle bracing conditions across several seasons in one or more sports.
2. A long-term prospective study to follow athletes from pre- to post-puberty in manner of performance of the jump landings under different bracing conditions.

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APPENDIX A

Institutional Review Board Documents

TEXAS WOMAN'S UNIVERSITY
DENTON DALLAS HOUSTON

Institutional Review Board
Office of Research and Sponsored Programs
P.O. Box 425619, Denton, TX 76204-5619
940-898-3378 Fax 940-898-3416
e-mail: IRB@twu.edu

May 12, 2006

Mr. Gary Christopher
Department of Kinesiology - PH 208

Dear Mr. Christopher:

Re: Effects of Ankle Bracing on Knee Kinematics and Kinetics During Jump Landings

The above referenced study has been reviewed by the TWU Institutional Review Board (IRB) and appears to meet our requirements for the protection of individuals' rights.

If applicable, agency approval letters must be submitted to the IRB upon receipt PRIOR to any data collection at that agency. A copy of the approved consent form with the IRB approval stamp and a copy of the annual/final report are enclosed. Please use the consent form with the most recent approval date stamp when obtaining consent from your participants. The signed consent forms and final report must be filed with the Institutional Review Board at the completion of the study.

This approval is valid one year from May 5, 2006. According to regulations from the Department of Health and Human Services, another review by the IRB is required if your project changes in any way, and the IRB must be notified immediately regarding any adverse events. If you have any questions, feel free to call the TWU Institutional Review Board.

Sincerely,

Dr. David Nichols, Chair
Institutional Review Board - Denton

enc.

cc. Dr. Charlotte Sanborn, Department of Kinesiology
Dr. Young Hoo Kwon, Department of Kinesiology
Graduate School

TEXAS WOMAN'S UNIVERSITY
CONSENT TO PARTICIPATE IN RESEARCH

Title: Effects of ankle bracing on knee kinematics and kinetics during jump landings

Investigator: Gary A. Christopher, MS, ATC, (940) 898-2498, garvalan59@mail.twu.edu

Advisor: Young-Hoo Kwon, Ph.D. (940) 898-2598, ykwon@mail.twu.edu

Explanation and Purpose of the Research

Female athletes are at greater risk of sustaining an injury to the anterior cruciate ligament (ACL) than male athletes. Risk factors for sustaining a non-contact ACL injury are generally broken down into extrinsic and intrinsic factors. Intrinsic factors are particular to an athlete and cannot be modified. Extrinsic factors are those that can be modified through changes in environment or conditioning. Some researchers have suggested ankle bracing or taping as a possible extrinsic risk factor for ACL injury.

The purpose of this study is to compare the motion and forces at the knee during jump landings with three ankle-bracing conditions: un-braced, standard athletic tape, and lace-up brace.

Research Procedures

We ask that you report to the Biomechanics lab (PH 124) and change into your own Lycra/spandex shorts and a t-shirt, tank top, or sports bra and perform a stretching/warm-up routine to which you are accustomed. We will then measure your two-handed standing-reach. You will then perform three two-footed maximum jumps to determine your jump height; a 30-second rest will be allowed between trials. Your vertical jump height will be computed as the difference between the average of your three jumps and your standing reach.

Following jump height measurement, you will perform motion trials in three ankle-bracing conditions: un-braced, athletic taping, and lace-up ankle brace. Order of trials will be randomized and rest will be allowed at your request. Ankle taping will be done by a Texas Licensed Athletic Trainer.

For motion trials, reflective markers will be affixed as shown in the attached diagrams. Where possible, markers will be affixed to your skin. Where markers are placed on skin, your skin will be cleaned with an alcohol swab prior to marker placement. Markers at your feet will be placed on your shoes.

A platform will be set to your maximum jump height; you will drop from the platform, landing with each foot on separate laboratory force plates. At the time of the drop, a lighted arrow will indicate a direction to move. Upon landing, you will stand-up straight, or pivot and step to the right or left, depending on the visual cue received. One valid trial in each of the three directions will be recorded, for nine trials (three bracing conditions × three trials per bracing condition).

Potential Risks

Release of confidential information: *Confidentiality will be protected to the extent that is allowed by law.* You will be identified by an alphanumeric code. All computer files and paperwork associated with you will be identified solely by this code. Computerized video images will be stored on DVD-ROM. Access to computerized video images and will be restricted to the principal

Parent/Guardian or Participant Initials_____

investigator, his advisor, and graduate assistants involved in video data collection and analysis. DVD's will be stored in a locked desk in the principal investigator's office (PH 211). DVD's will be destroyed in three years or upon publication of research results, whichever occurs first.

Coercion: If you are a student of the principal investigator, you will be recruited by a graduate assistant. The principal investigator will recruit all other participants. If you are a student of the principal investigator, no extra credit in any course will be offered. Your standing on your team or grade in a class taught by the principal investigator will not be affected by participation or non-participation.

Injury: You will be screened for history of knee, ankle, hip, or spinal injuries or abnormalities; such history will be disqualifying. Disqualifying conditions include, but are not limited to: sprain or rupture of any of the major ligaments of the knee (Anterior Cruciate Ligament (ACL), Posterior Cruciate Ligament (PCL), Medial Collateral Ligament (MCL), Lateral Collateral Ligament (LCL)); congenital absence of ACL or PCL; ankle sprain (for which you are symptomatic or under a doctor's care); flat feet; herniated spinal disk; avulsion of any muscle attaching to pelvis, femur, or tibia. If you wear ankle braces by physician or therapist prescription, you will be ineligible. Ultimately, your risk of injury during this study is no greater than the risk encountered by competitive athletes who voluntarily wear ankle braces or ankle taping. Personnel trained in First Aid will be present during data collection. Ice bags will be available during data collection to apply in the event of an injury. If you are injured, you will be treated according to standard first aid practices and will be advised to visit the Emergency Room, Student Health Clinic (if you are a TWU student), or your family physician. Emergency medical services will be called if deemed necessary by the principal investigator or first aid personnel, or if requested by your parents (if you are under 18).

Embarrassment: Clothing specifications are necessitated by data collection procedures. To ensure accurate and consistent marker placement, the principal investigator will place all markers. At least two investigators will be present during marker placement. You will not be left alone in a room with any investigator. If you are under 18, your parent(s)/guardian(s) will allowed in the room during marker placement. No other participants will be present during testing sessions.

The researchers will try to prevent any problem that could happen because of this research. You should let the researchers know at once if there is a problem and they will help you. However, TWU does not provide medical services or financial assistance for injuries that might happen because you are taking part in this research.

Participation and Benefits

Your involvement in this research study is completely voluntary and you may discontinue your participation in the study at any time without penalty. Direct benefits of this study to you are: a copy of the results will be mailed to you upon request.*

Approved by the
Texas Woman's University
Institutional Review Board
Approved: May 5, 2008
Revised: July 11, 2008

Parent/Guardian or Participant Initials_____

Questions Regarding the Study

If you have any questions about the research study you may ask the researcher or his advisor; their phone numbers are at the top of this form. If you have questions about your rights as a participant in this research or the way this study has been conducted, you may contact the Texas Woman's University Office of Research and Grants at 940-898-3378 or via e-mail at IRB@twu.edu. You will be given a copy of this consent form to keep.

_____ _____

Signature of Participant Date

_____ _____

Signature of Parent/Guardian Date

***If you would like to receive a summary of the results of this study, please provide an address to which this summary should be sent:**

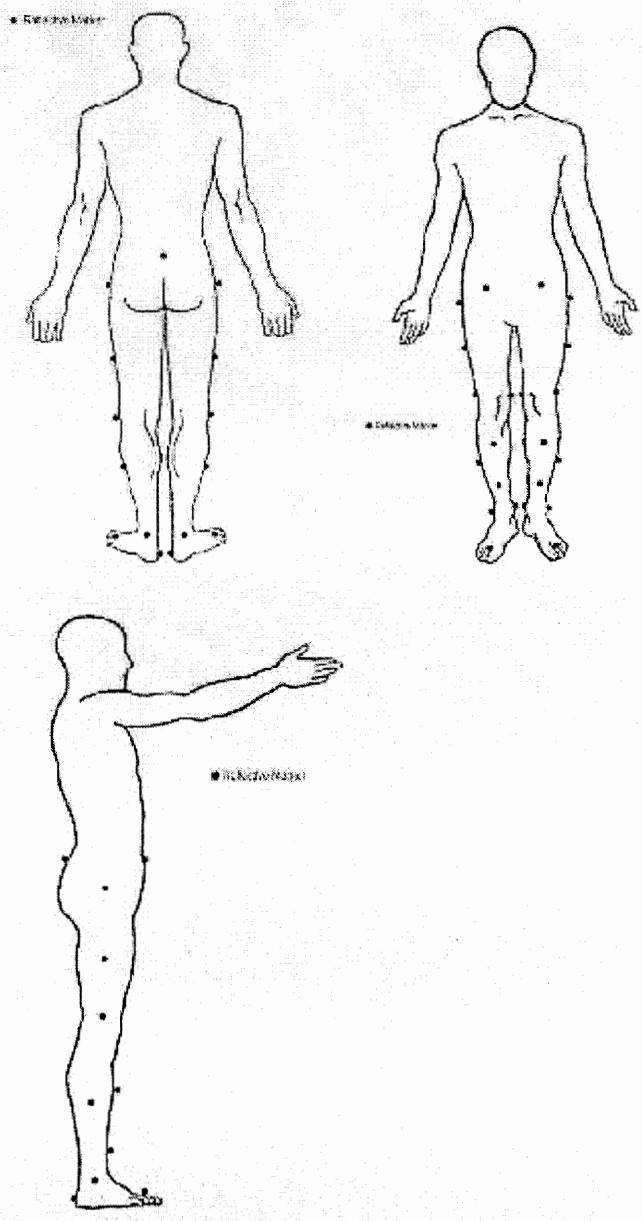
_____

_____

_____

Approved by the
Texas Woman's University
Institutional Review Board
Approved: May 6, 2008
Revised: July 11, 2006

Marker Placement Diagrams



Approved by the
Texas Woman's University
Institutional Review Board
Approved: May 5, 2005
Revised: July 11, 2006

APPENDIX B

Participant Data

Table 11

Individual Participant Characteristics

Participant	Sport	Age	Height (cm)	Mass (kg)	Weight (N)	Body Mass Index (kg/m^2)	Jump Height (cm)			
							1	2	3	Average[a]
GCD25	Volleyball	23	178.5	84.0	824.0	26.4	33.0	34.3	38.1	35.0
GCD30	Gymnastics	19	158.0	67.8	665.1	27.2	36.8	39.4	38.1	38.0
GCD31	Gymnastics	18	165.0	63.8	625.9	23.4	27.9	33.0	38.1	33.0
GCD32	Gymnastics	20	164.0	62.5	613.1	23.2	43.2	44.5	45.7	44.0
GCD35	Volleyball	21	171.0	54.5	534.6	18.6	45.7	45.7	44.5	45.0
GCD36	Volleyball	20	167.5	61.1	599.4	21.8	45.7	47.0	52.1	48.0
GCD37	Basketball	21	172.0	62.0	608.2	21.0	39.4	41.9	43.2	41.0
GCD38	Soccer	20	160.0	56.6	555.2	22.1	33.0	34.3	36.8	35.0

[a]Average jump height rounded to nearest centimeter.

Table 12

Deep Squat Data

Participant	Bracing Condition	Leg	Maximum Knee Flexion Angle (°)	Maximum Ankle Dorsiflexion Angle (°)
GCD25	Untaped	Right	136.4957	123.4953
		Left	135.0633	122.7161
	Taped	Right	124.8189	112.0247
		Left	123.4781	108.7588
	Braced	Right	132.5824	123.9193
		Left	133.8086	125.0079
GCD30	Untaped	Right	91.6561	114.3051
		Left	93.3004	115.3249
	Taped	Right	96.9674	113.4113
		Left	95.3173	113.1878
	Braced	Right	89.5132	114.6259
		Left	85.8176	113.6977
GCD31	Untaped	Right	116.8261	120.3441
		Left	110.2543	110.7126
	Taped	Right	96.5090	111.9961
		Left	94.8417	107.9281
	Braced	Right	109.1828	116.4021
		Left	103.9517	108.5125

Table 12 (continued)

Deep Squat Data

Participant	Bracing Condition	Leg	Maximum Knee Flexion Angle (°)	Maximum Ankle Dorsiflexion Angle (°)
GCD 32	Untaped	Right	128.1191	117.3475
		Left	127.6378	115.7547
	Taped	Right	114.7176	113.8926
		Left	116.3276	112.9987
	Braced	Right	120.0805	112.3055
		Left	119.8857	114.1447
GCD35	Untaped	Right	87.2386	116.4136
		Left	89.6392	120.1894
	Taped	Right	72.5708	112.1737
		Left	70.9436	114.1504
	Braced	Right	88.4704	116.0354
		Left	90.6362	119.0091
GCD36	Untaped	Right	103.4074	111.9789
		Left	103.4762	110.6439
	Taped	Right	96.3600	107.6416
		Left	93.6270	104.2611
	Braced	Right	93.2317	104.5992
		Left	95.9361	105.2065

Table 12 (continued, 2)

Deep Squat Data

Participant	Bracing Condition	Leg	Maximum Knee Flexion Angle (°)	Maximum Ankle Dorsiflexion Angle (°)
GCD37	Untaped	Right	122.4124	122.2062
		Left	121.5931	124.6756
	Taped	Right	82.4887	105.0232
		Left	85.8577	108.8104
	Braced	Right	105.3268	114.3108
		Left	107.0973	119.8800
GCD38	Untaped	Right	98.1018	114.4540
		Left	97.9815	114.2879
	Taped	Right	64.5494	101.6714
		Left	66.2855	105.8654
	Braced	Right	77.6415	109.2401
		Left	78.0941	108.2833

The following notes apply to Tables 13, 14, and 15:

[a] Prior to data processing, left leg values were multiplied by -1 to align with the right leg.

[b] Frontal plane angles: Right leg: positive = varus, negative = valgus. Left leg: positive = valgus, negative = varus.

[c] Rotation angles: Right leg: positive = internal tibial rotation, negative = external tibial rotation. Left leg: positive = external tibial rotation, negative = internal tibial rotation.

[d] Superior tibial translation (cm) from foot strike to maximum knee flexion; positive = anterior, negative = posterior.

The following notes apply to Tables 16, 17, and 18:

[a] Prior to data processing, moment data were normalized by dividing by the product of participant's body weight and height, then multiplying by 100.

[b] Prior to data processing, left leg values were multiplied by -1 to align with the right leg.

[c] Extension moments: positive = extension moment, negative = flexion moment.

[d] Frontal plane moments: Right leg: positive = varus moment, negative = valgus moment. Left leg: positive = valgus moment, negative = varus moment.

[e] Rotation moments: Right leg: positive = internal tibial rotation moment, negative = external tibial rotation moment. Left leg: positive = external tibial rotation moment, negative = internal tibial rotation moment.

[f] GRF = Maximum resultant ground reaction force, computed as the vector sum of the mediolateral, anterior-posterior, and vertical ground reaction forces. Prior to data processing, GRF data were normalized by dividing by participant's body weight, then multiplying by 100.

Table 13

Kinematic Data – Untaped

			Knee Angle at Foot Strike (°)			Maximum Knee Angle (°)			Tibial Translation[d]
Participant	Trial	Leg[a]	Flexion	Frontal[b]	Rotation[c]	Flexion	Frontal[b]	Rotation[c]	
GCD25	left1	Right	19.2514	-7.9011	-1.8507	51.1021	9.5684	-19.6066	-0.0007
		Left	16.2090	-6.2911	28.1322	75.7794	-10.2789	28.1322	-0.0121
	left2	Right	21.7266	-5.6207	-19.1139	52.0074	9.0757	-24.5226	0.0083
		Left	14.2208	-8.4110	28.8771	79.3547	-12.5707	28.8771	-0.0257
	right1	Right	17.8075	8.7720	-20.3572	72.9089	12.3472	. -20.3572	-0.0265
		Left	24.2877	4.1024	12.8228	52.8783	-7.5573	26.7743	0.0050
	right3	Right	26.6941	6.2796	-17.1314	70.6228	11.3847	-17.1314	-0.0126
		Left	19.0050	8.4912	10.6169	47.0971	-10.0554	28.1322	0.0112
	straight1	Right	27.2556	1.6215	-20.8213	80.0078	4.4519	-25.8461	-0.0141
		Left	24.7059	-0.6360	13.1952	84.5342	6.3598	28.1551	-0.0074
	straight2	Right	23.1761	2.7674	-21.8641	71.1785	6.1020	-22.8152	-0.0119
		Left	23.5658	0.4584	12.7827	76.1461	4.5092	28.7510	-0.0065

Table 13 (continued)

Kinematic Data – Untaped (continued)

			Knee Angle at Foot Strike (°)			Maximum Knee Angle (°)			Tibial Translation[d]
Participant	Trial	Leg[a]	Flexion	Frontal[b]	Rotation[c]	Flexion	Frontal[b]	Rotation[c]	
GCD30	left1	Right	14.7021	-2.7846	-16.2720	77.2519	-6.5088	-26.5222	-0.0052
		Left	31.6215	-6.7036	3.9419	90.7049	-25.3763	-18.8331	-0.0098
	left2	Right	19.8301	2.2861	-18.2430	73.7569	-4.9102	-23.1532	-0.0096
		Left	34.2973	-6.3197	7.4141	88.6538	-22.7636	-20.6952	-0.0033
	right1	Right	24.1731	-3.4206	-12.2326	83.8352	-8.3766	-15.7220	-0.0055
		Left	26.2185	-4.6295	8.2907	70.3019	-15.2693	15.1948	0.0062
	right2	Right	24.2648	-2.0626	-19.3373	93.8791	-11.4649	-19.3373	-0.0096
		Left	27.7655	-6.2796	7.4943	76.6388	-18.9420	18.6555	-0.0050
	straight1	Right	20.2426	0.9397	-13.2525	75.9627	-7.1963	-20.3400	-0.0087
		Left	26.2701	-5.7296	8.9267	75.5903	-16.9424	8.9267	-0.0070
	straight2	Right	26.4707	0.7850	-11.8946	93.1228	-15.5386	-20.1968	-0.0019
		Left	36.6464	-8.8006	7.5745	97.4601	-8.8006	8.4970	0.0009

Table 13 (continued, 2)

Kinematic Data – Untaped (continued)

			Knee Angle at Foot Strike (°)			Maximum Knee Angle (°)			Tibial Translation[d]
Participant	Trial	Leg[a]	Flexion	Frontal[b]	Rotation[c]	Flexion	Frontal[b]	Rotation[c]	
GCD31	left1	Right	19.8014	-2.3491	0.0286	36.1078	-2.6184	-28.8656	0.0064
		Left	28.3614	1.4037	0.4068	66.0792	8.7548	-8.9210	-0.0012
	left2	Right	19.7499	-8.3709	-8.8064	69.5456	-14.2838	-9.0470	0.0056
		Left	22.9985	1.0371	-14.6906	57.9661	4.1654	-16.5986	0.0102
	right2	Right	38.9038	-6.2166	2.5840	56.1384	-14.1807	5.4431	0.0047
		Left	14.6677	5.2254	-13.1780	39.9466	5.2254	-13.1780	0.0228
	right3	Right	39.0585	-6.1937	1.6730	56.6082	-16.4955	-16.7476	0.0000
		Left	15.0974	4.9045	0.3266	47.0112	4.9045	-11.9175	0.0056
	straight1	Right	25.4107	-0.4469	-3.8331	62.3894	-13.7739	-13.4301	0.0000
		Left	24.1330	1.7303	-8.2048	62.7274	7.0073	-12.2097	0.0079
	straight2	Right	24.9409	-7.4714	1.1115	51.1250	-17.4065	13.6364	0.0069
		Left	18.7758	3.6039	-7.1047	51.4917	11.5279	-20.7124	-0.0041

Table 13 (continued, 3)

Kinematic Data – Untaped

			Knee Angle at Foot Strike (°)			Maximum Knee Angle (°)			Tibial Translation[d]
Participant	Trial	Leg[a]	Flexion	Frontal[b]	Rotation[c]	Flexion	Frontal[b]	Rotation[c]	
GCD32	left1	Right	15.4469	-4.5034	-12.3415	60.8023	-15.9913	25.9435	0.0017
		Left	52.8554	6.7895	-0.8709	70.3879	16.4038	13.0634	-0.0014
	left2	Right	14.9886	-0.0516	1.8277	57.7656	-12.0894	16.6100	-0.0042
		Left	40.8347	1.9079	-3.6956	72.2500	18.2143	12.8572	0.0024
	right1	Right	32.9164	-1.8678	-9.3507	84.5399	-31.4497	-9.3507	-0.0026
		Left	13.5848	0.3610	-1.7303	60.8481	5.9588	-9.6773	0.0112
	right2	Right	19.6868	4.2456	-1.8105	80.2828	-17.0340	5.9817	-0.0160
		Left	19.4691	-0.2235	-2.7502	48.8790	3.2200	9.9465	0.0094
	straight1	Right	17.7216	1.1115	-1.2032	61.2606	-10.5825	10.2559	-0.0100
		Left	17.4351	-1.6902	5.9244	57.1239	9.1731	5.9244	-0.0002
	straight2	Right	24.0241	-0.6188	-3.6326	63.3691	-15.6704	13.8312	0.0009
		Left	14.9370	-1.5928	-6.0848	57.5078	8.3537	-10.5825	0.0019

Table 13 (continued, 4)

Kinematic Data – Untaped

			Knee Angle at Foot Strike (°)			Maximum Knee Angle (°)			Tibial Translation[d]
Participant	Trial	Leg[a]	Flexion	Frontal[b]	Rotation[c]	Flexion	Frontal[b]	Rotation[c]	
GCD35	left1	Right	13.9057	-6.4859	5.0420	71.3504	-33.9764	18.4321	0.0135
		Left	55.4852	-1.4553	-5.6379	89.5991	-15.0631	24.1788	0.0075
	left2	Right	8.6975	-6.3598	-2.0397	50.0421	-19.9389	20.1223	0.0061
		Left	58.5105	11.3388	-1.9366	80.0078	19.3201	7.5745	0.0123
	right1	Right	21.1708	-6.8125	8.3881	76.3466	-41.1670	18.8274	-0.0054
		Left	19.9618	0.4927	16.1631	53.4169	-6.0275	16.1631	0.0050
	right2	Right	21.7839	-4.1941	-15.7621	81.4345	-35.9187	22.1047	-0.0097
		Left	23.2965	-1.0084	26.5795	23.2965	-1.0084	26.5795	0.0032
	straight1	Right	20.2999	-6.0046	-10.0038	80.2198	-36.9844	15.8709	-0.0073
		Left	38.0387	-2.0111	-15.9741	84.1217	7.7063	-17.0913	0.0090
	straight2	Right	16.2147	-4.3144	-3.4721	81.4288	-30.4584	12.1181	-0.0131
		Left	28.6880	-9.3793	3.4836	86.0926	-9.3793	16.9997	-0.0074

Table 13 (continued, 5)

Kinematic Data – Untaped

			Knee Angle at Foot Strike (°)			Maximum Knee Angle (°)			Tibial Translation[d]
Participant	Trial	Leg[a]	Flexion	Frontal[b]	Rotation[c]	Flexion	Frontal[b]	Rotation[c]	
GCD36	left1	Right	11.7113	-2.0741	-19.0795	67.4199	-5.0764	-19.1196	-0.0089
		Left	30.8079	2.7158	-1.4782	76.8279	18.5581	-13.4244	-0.0060
	left2	Right	16.6043	-2.2002	-12.4504	70.4051	-6.0676	-18.7873	-0.0079
		Left	35.1223	-0.3266	-9.2991	87.3360	21.8240	-12.3988	-0.0094
	right2	Right	21.9099	3.6326	-18.5237	90.6018	19.6123	-18.5237	-0.0229
		Left	13.5963	-3.3461	-2.4637	76.1117	10.2961	-11.9863	-0.0100
	right3	Right	31.2491	3.6211	-17.7732	89.2897	18.3633	-17.7732	-0.0210
		Left	20.1681	0.0630	-6.9557	69.8034	10.0611	-12.8858	-0.0054
	straight1	Right	17.4236	2.8762	-17.2231	86.0353	-4.8587	-28.0864	-0.0291
		Left	18.6211	-2.3950	-1.5413	83.3023	9.1272	-16.5356	-0.0178
	straight2	Right	13.6479	3.3862	-16.1861	77.6587	6.9041	-30.0402	-0.0183
		Left	12.7369	-3.3346	5.1967	76.8451	7.4943	-10.5768	-0.0163

Table 13 (continued, 6)

Kinematic Data Untaped

			Knee Angle at Foot Strike (°)			Maximum Knee Angle (°)			Tibial Translation[d]
Participant	Trial	Leg[a]	Flexion	Frontal[b]	Rotation[c]	Flexion	Frontal[b]	Rotation[c]	
GCD37	left1	Right	26.3160	-13.9000	12.0207	52.9757	-24.7002	22.3797	0.0155
		Left	21.8354	-0.4698	0.8136	71.8374	23.8465	-12.8457	-0.0118
	left2	Right	31.0772	-15.3839	11.8660	48.9936	-20.7296	13.7510	0.0086
		Left	28.7338	1.5069	-4.3430	70.2045	32.4695	-36.8698	0.0030
	right1	Right	36.9214	-10.0268	2.2116	77.2748	-32.3950	11.1784	0.0064
		Left	17.8476	5.9473	-12.3415	57.4333	11.1498	-12.5993	0.0027
	right2	Right	37.4600	-11.6368	2.7903	71.1155	-27.8916	5.2254	0.0023
		Left	26.1498	7.0187	-8.2850	60.2752	12.1467	-22.8037	0.0175
	straight1	Right	35.7583	-8.6230	-0.5901	69.3451	-23.1246	7.5688	0.0001
		Left	24.8091	4.0279	-6.7838	78.8390	23.9095	-16.6960	0.0134
	straight2	Right	43.7912	-13.9687	1.7762	60.1319	-22.9355	9.9294	0.0020
		Left	26.6425	3.4893	-3.5008	73.6824	22.0933	-17.9393	-0.0012

Table 13 (continued, 7)

Kinematic Data Untaped

			Knee Angle at Foot Strike (°)			Maximum Knee Angle (°)			Tibial Translation[d]
Participant	Trial	Leg[a]	Flexion	Frontal[b]	Rotation[c]	Flexion	Frontal[b]	Rotation[c]	
GCD38	left1	Right	20.8843	-9.0814	33.5467	57.3015	-24.3564	45.1032	0.0074
		Left	48.1686	-0.3209	-2.2403	86.2531	17.3778	15.0058	-0.0093
	left2	Right	33.9420	-12.3587	43.2411	62.9222	-19.4004	43.2411	0.0012
		Left	24.7518	-1.7934	7.3109	87.9261	21.6292	-17.3721	-0.0138
	right1	Right	53.3653	-7.9641	23.6059	78.7932	-11.3961	27.8744	-0.0006
		Left	17.5325	2.6643	-2.5497	59.8855	16.1402	-16.8679	-0.0072
	right2	Right	71.5223	-4.3831	18.6383	84.0586	-18.0654	31.0257	-0.0023
		Left	21.6062	4.2857	1.0199	56.9749	19.8129	-34.0280	-0.0065
	straight1	Right	48.9764	-4.8816	15.6074	81.4517	-20.4317	25.2445	0.0029
		Left	24.8492	-3.4034	10.6742	94.8073	27.6911	-14.3239	0.0025
	straight2	Right	54.7977	-2.3606	16.9596	82.5632	-20.3572	23.5027	-0.0036
		Left	35.8672	-5.6150	5.1337	99.1389	25.0841	-9.6486	-0.0013

Table 14

Kinematic Data Taped

			Knee Angle at Foot Strike (°)			Maximum Knee Angle (°)			Tibial Translation[d]
Participant	Trial	Leg[a]	Flexion	Frontal[b]	Rotation[c]	Flexion	Frontal[b]	Rotation[c]	
GCD25	left1	Right	22.2537	-9.6486	-17.4408	50.9589	-9.6486	-28.6823	0.0085
		Left	24.9981	-11.2357	27.8343	82.5117	-17.7903	27.8343	-0.0177
	left2	Right	29.6105	-17.3950	-1.6100	59.4043	-18.0023	-28.8656	0.0018
		Left	15.7506	-6.5088	26.3331	68.4513	-22.2480	26.3331	-0.0208
	right1	Right	27.9432	-3.4148	-8.9553	80.8042	11.5451	-12.6051	-0.0265
		Left	16.5069	6.6406	17.6471	59.4959	-7.1562	33.6384	-0.0020
	right2	Right	23.5027	4.1826	-35.0249	77.9509	9.1215	-35.0249	-0.0107
		Left	21.2281	8.5142	9.7403	52.1506	-11.1784	22.6777	0.0141
	straight1	Right	32.8419	-7.1849	-8.2219	73.0636	-12.5650	-18.7529	0.0064
		Left	38.1647	4.2342	3.4950	81.3142	18.3346	29.6792	0.0199
	straight2	Right	38.0902	-4.4576	-8.2449	86.2932	-13.1207	-22.9069	0.0068
		Left	35.4145	2.7101	10.2101	88.9230	24.6486	25.6055	0.0284

Table 14 (continued)

Kinematic Data Taped

Participant	Trial	Leg[a]	Knee Angle at Foot Strike (°) Flexion	Frontal[b]	Rotation[c]	Maximum Knee Angle (°) Flexion	Frontal[b]	Rotation[c]	Tibial Translation[d]
GCD30	left1	Right	25.2674	-3.7128	5.7983	70.1816	-13.4473	18.7185	0.0237
		Left	32.6930	-1.0371	0.1375	93.4093	16.6903	5.8614	0.0123
	left2	Right	26.2243	-4.4462	5.7811	66.5548	-8.6574	15.0172	0.0288
		Left	45.2522	3.0481	-9.8721	85.8921	20.7124	-10.1184	0.0150
	right1	Right	34.6353	-2.0913	-0.3953	94.2859	-8.3423	-7.8839	-0.0035
		Left	31.2491	3.6326	-0.5386	80.9589	7.9183	-10.9664	-0.0013
	right2	Right	31.2319	-3.7128	1.9251	79.8875	-8.2563	-2.7788	-0.0031
		Left	28.6078	4.1081	2.0569	64.5953	7.4198	8.5428	0.0016
	straight1	Right	23.6173	-3.7013	-6.1708	92.5900	-26.5279	19.4061	0.0046
		Left	32.4008	7.8495	8.9611	87.5537	16.7877	-14.6620	0.0012
	straight2	Right	29.0604	-0.2636	2.1772	86.6026	-24.6601	18.7243	0.0158
		Left	29.9943	0.9282	2.0741	88.9632	15.2464	-17.0512	-0.0022

Table 14 (continued, 2)

Kinematic Data Taped

			Knee Angle at Foot Strike (°)			Maximum Knee Angle (°)			Tibial Translation[d]
Participant	Trial	Leg[a]	Flexion	Frontal[b]	Rotation[c]	Flexion	Frontal[b]	Rotation[c]	
GCD31	left1	Right	19.2285	-5.2254	11.1555	58.5506	14.2724	-20.1910	0.0095
		Left	36.0333	2.2517	12.8744	70.9952	13.6708	20.3343	-0.0051
	left2	Right	22.4771	-3.1226	8.6689	60.5273	-15.1375	-10.0841	0.0015
		Left	37.8439	2.3262	5.3915	73.7683	9.3106	-10.7430	0.0001
	right1	Right	32.7789	-5.3915	-0.7105	50.1223	-13.7052	10.7315	-0.0058
		Left	21.8068	6.7552	2.0111	45.9168	6.7552	-10.7200	0.0086
	right2	Right	26.8431	-2.2517	0.4125	57.6052	-9.2705	-17.0570	-0.0002
		Left	18.0539	0.7792	27.2327	53.3710	7.6146	27.2327	0.0072
	straight1	Right	29.3412	-1.9481	-5.9473	62.2977	-19.6295	11.9462	-0.0023
		Left	31.8278	2.7215	-8.5256	61.1804	12.1582	-25.3247	0.0103
	straight2	Right	28.6078	-5.3400	0.9740	53.9898	-16.7189	19.6582	-0.0027
		Left	24.5971	0.0172	-1.8736	51.8355	7.5917	-10.9836	0.0030

Table 14 (continued, 3)

Kinematic Data Taped

Participant	Trial	Leg[a]	Knee Angle at Foot Strike (°) Flexion	Frontal[b]	Rotation[c]	Maximum Knee Angle (°) Flexion	Frontal[b]	Rotation[c]	Tibial Translation[d]
GCD32	left1	Right	17.5440	-1.4496	-0.2464	54.7117	-12.7197	5.7067	-0.0029
		Left	40.4623	-8.5371	-3.6956	67.0991	-14.3870	10.4909	-0.0074
	left2	Right	17.4638	-1.4668	-2.8304	61.3982	-15.6819	-11.6425	-0.0054
		Left	45.3725	-1.0428	7.4026	70.1243	11.1097	17.4981	-0.0070
	right1	Right	31.3924	0.5730	-9.0126	75.2408	-21.5146	-15.2865	-0.0146
		Left	18.3919	5.4316	12.7082	48.7014	5.4316	12.7082	0.0029
	right2	Right	30.9913	-0.6474	-8.5027	78.6499	-19.9504	-16.1631	-0.0114
		Left	20.9072	2.0569	14.5818	53.2851	2.0569	14.5818	-0.0045
	straight1	Right	21.2968	-0.4412	-10.2502	69.3852	-15.0172	-10.2502	-0.0145
		Left	22.6490	1.7074	18.2602	62.4467	8.0672	18.2602	-0.0022
	straight2	Right	23.9496	-0.3380	-11.5279	64.3775	-10.8232	-11.7915	-0.0139
		Left	23.4282	1.1058	15.9511	58.9917	4.1024	15.9511	0.0008

Table 14 (continued, 4)

Kinematic Data Taped

Participant	Trial	Leg[a]	Knee Angle at Foot Strike (°) Flexion	Frontal[b]	Rotation[c]	Maximum Knee Angle (°) Flexion	Frontal[b]	Rotation[c]	Tibial Translation[d]
GCD35	left1	Right	13.0577	-5.0936	1.8736	54.0299	-17.7159	21.4114	-0.0054
		Left	49.1999	-6.6062	-0.0859	80.7928	-22.0474	14.5589	0.0054
	left2	Right	16.0485	-2.1601	-6.1536	74.3986	-22.9756	13.1608	-0.0050
		Left	50.0708	-2.6471	5.3858	92.7447	-17.2747	7.1792	0.0040
	right1	Right	22.0073	-4.2227	1.2605	76.5242	-31.5642	-9.5684	-0.0189
		Left	20.0593	-7.6776	7.3167	57.9031	-10.7143	35.0937	-0.0086
	right2	Right	21.5432	-3.2888	-5.9129	83.0445	-37.5459	14.2437	-0.0072
		Left	26.0237	-0.6016	20.4088	65.2198	4.5837	20.4088	0.0012
	straight1	Right	13.8828	-3.7873	-4.9102	101.1672	-32.7445	20.6437	-0.0266
		Left	33.5467	-4.5550	18.7816	99.1446	6.6234	25.4680	-0.0098
	straight2	Right	17.9508	-1.3866	-9.2361	96.7325	-29.5990	-9.2361	-0.0284
		Left	33.8733	-2.7101	14.6276	90.7680	5.9931	24.1674	-0.0044

Table 14 (continued, 5)

Kinematic Data Taped

Participant	Trial	Leg[a]	Knee Angle at Foot Strike (°) Flexion	Frontal[b]	Rotation[c]	Maximum Knee Angle (°) Flexion	Frontal[b]	Rotation[c]	Tibial Translation[d]
GCD36	left1	Right	25.5940	0.2636	-7.3854	73.2698	-5.5520	-11.8144	-0.0053
		Left	25.9951	2.6528	-5.3285	87.9605	22.7292	-12.2384	-0.0076
	left2	Right	22.4542	-7.1505	-7.9928	73.6766	-15.6647	-15.1834	-0.0042
		Left	33.0826	11.0352	-26.4936	84.6488	26.4477	-27.2556	0.0045
	right1	Right	27.6280	-4.8243	-9.5512	66.0506	-17.0226	-10.3476	-0.0073
		Left	15.7105	2.2345	-3.3919	53.1991	8.5714	-14.4844	-0.0053
	right2	Right	24.9924	-6.3770	-9.3106	81.3199	-19.8415	-11.9920	-0.0090
		Left	21.2682	5.8213	-6.2223	62.6930	13.5963	-16.7418	-0.0018
	straight1	Right	24.0184	-0.1261	-12.6624	84.5571	-12.8916	-27.1066	-0.0165
		Left	17.6070	-0.8021	-3.8331	78.2947	9.6944	-13.8484	-0.0160
	straight2	Right	31.6101	-2.8934	-13.2984	83.6117	-8.3824	-30.7678	-0.0160
		Left	24.9752	-0.9855	-9.5512	83.6862	5.7410	-27.5364	-0.0156

Table 14 (continued, 6)

Kinematic Data Taped

Participant	Trial	Leg[a]	Knee Angle at Foot Strike (°) Flexion	Frontal[b]	Rotation[c]	Maximum Knee Angle (°) Flexion	Frontal[b]	Rotation[c]	Tibial Translation[d]
GCD37	left1	Right	29.7594	-10.7143	12.0264	55.5941	-13.4301	12.0264	0.0105
		Left	28.7052	5.2483	-4.8759	59.5131	20.4088	-27.7483	-0.0110
	left2	Right	35.3859	-17.3835	16.7361	55.6743	-19.5722	17.8820	0.0251
		Left	29.9141	-1.1001	-1.7475	66.2740	21.5890	-21.3255	-0.0027
	right1	Right	30.9168	-2.1028	-10.3935	71.0296	-27.1983	-10.3935	-0.0093
		Left	18.7243	4.8358	-8.7605	55.6285	-8.0672	-8.7605	0.0145
	right2	Right	34.6926	-1.2548	0.0000	64.6526	-27.9603	13.3614	-0.0132
		Left	21.9443	5.2426	-3.5581	42.4676	6.0046	6.5947	0.0119
	straight1	Right	37.3626	-5.1853	3.8560	67.5689	-24.3163	7.0417	0.0014
		Left	30.0001	-2.0283	3.1341	75.5789	16.1001	-16.6673	0.0022
	straight2	Right	37.2594	-6.9901	-0.4985	69.7748	-27.1124	13.2296	-0.0019
		Left	30.4699	1.6329	-5.5405	70.5483	24.2132	-25.1013	0.0012

Table 14 (continued, 7)

Kinematic Data – Taped

			Knee Angle at Foot Strike (°)			Maximum Knee Angle (°)			Tibial Translation[d]
Participant	Trial	Leg[a]	Flexion	Frontal[b]	Rotation[c]	Flexion	Frontal[b]	Rotation[c]	
GCD38	left1	Right	19.6983	-5.7468	-52.5116	67.5575	-26.8603	-52.5116	0.0104
		Left	31.0429	-3.1685	-5.8041	80.1969	13.2869	-16.4725	-0.0068
	left2	Right	38.4684	-9.4022	-7.2708	47.2919	-13.9802	-12.3873	-0.0053
		Left	18.0310	-3.0596	4.2571	78.4666	14.9313	-18.3690	-0.0076
	right1	Right	26.6998	-1.0829	-6.4515	75.7164	-15.9110	9.3736	-0.0073
		Left	34.8186	-1.5241	-27.7770	59.0949	12.1181	-34.5837	0.0003
	right2	Right	18.8102	1.4037	-7.0989	77.5728	-19.3774	10.0726	-0.0086
		Left	25.1815	3.3289	-1.9710	57.2843	12.7598	-22.4428	0.0010
	straight1	Right	23.9783	-0.7219	-0.4011	61.0830	-19.7327	11.5336	-0.0018
		Left	33.6040	-4.1654	-18.6211	56.4306	6.1077	-18.7529	-0.0040
	straight2	Right	27.1353	-1.8621	2.9450	79.5838	-24.5398	10.5768	-0.0072
		Left	45.8366	-0.6131	-12.6452	88.9402	19.4118	-14.6104	-0.0061

Table 15

Kinematic Data Braced

Participant	Trial	Leg[a]	Knee Angle at Foot Strike (°) Flexion	Frontal[b]	Rotation[c]	Maximum Knee Angle (°) Flexion	Frontal[b]	Rotation[c]	Tibial Translation[d]
GCD25	left1	Right	17.7159	-9.6887	-13.6307	60.7622	-9.6887	-25.0497	0.0034
		Left	14.3354	-5.6723	25.9951	82.1507	-18.9592	25.9951	-0.0204
	left2	Right	20.3801	-7.8495	-14.9141	49.2228	-7.8495	-25.3133	0.0040
		Left	12.2097	-2.2689	25.6800	78.2890	-8.2391	25.6800	-0.0151
	right1	Right	17.7159	8.7376	-27.6567	71.4249	10.9034	-27.6567	-0.0242
		Left	21.7437	9.5111	8.0501	55.3076	9.5111	25.6914	0.0038
	right2	Right	15.8652	6.9614	-29.8225	77.0113	9.4595	-29.8225	-0.0322
		Left	21.5146	3.5351	12.5077	50.4031	-3.5466	30.1777	0.0069
	straight1	Right	26.2472	1.1918	-19.6410	87.8860	4.7269	-20.8098	-0.0271
		Left	19.5207	-1.2089	18.3919	91.4154	-5.1509	24.4882	-0.0191
	straight2	Right	13.7624	1.5355	-20.6952	79.9391	12.0092	-24.5684	-0.0463
		Left	25.0612	-0.0401	24.5455	82.1049	4.7842	26.9061	-0.0125

Table 15 (continued)

Kinematic Data Braced

			Knee Angle at Foot Strike (°)			Maximum Knee Angle (°)			Tibial Translation[d]
Participant	Trial	Leg[a]	Flexion	Frontal[b]	Rotation[c]	Flexion	Frontal[b]	Rotation[c]	
GCD30	left1	Right	19.5035	-4.8931	-4.0623	55.9837	-5.3801	-0.3380	0.0085
		Left	44.6907	-0.9282	-9.2762	80.4318	-12.7197	-8.9324	-0.0042
	left2	Right	22.9928	-3.3747	-3.9305	73.4704	-9.4882	-13.3442	0.0025
		Left	31.3465	-6.4630	2.3205	90.3383	-21.5661	-12.7369	-0.0041
	right1	Right	31.0486	-1.7762	-2.9794	77.5842	-5.7811	-5.8270	-0.0104
		Left	19.7384	0.0401	9.4882	46.9825	-8.0615	11.9462	0.0018
	right2	Right	35.6781	-5.1795	-4.9733	90.0919	-12.6681	-9.9351	-0.0061
		Left	28.2124	-4.0164	4.9274	71.0640	-16.4439	-8.0214	0.0044
	straight1	Right	31.5127	-3.1398	-6.9156	88.4819	-19.4920	-6.9156	0.0078
		Left	36.3026	-4.4347	2.1257	89.9085	8.7376	-7.5229	-0.0006
	straight2	Right	28.3557	-4.5951	4.8472	76.2320	-24.3507	13.7567	0.0061
		Left	31.6731	-2.6986	-3.5008	81.2282	12.7884	-10.3763	0.0087

Table 15 (continued, 2)

Kinematic Data Braced

Participant	Trial	Leg[a]	Knee Angle at Foot Strike (°) Flexion	Frontal[b]	Rotation[c]	Maximum Knee Angle (°) Flexion	Frontal[b]	Rotation[c]	Tibial Translation[d]
GCD31	left1	Right	22.7120	-8.7433	-21.3541	52.4829	-10.8518	-21.3541	0.0049
		Left	35.6552	-0.4412	-5.5749	57.7943	6.5432	-21.3484	0.0048
	left2	Right	18.7930	-4.6753	-13.1322	52.9127	-14.9026	-13.1322	0.0083
		Left	27.7541	0.9568	-4.6524	65.8844	8.7147	-22.3339	0.0007
	right1	Right	38.9898	-7.4427	-7.2021	53.7320	-12.3014	-11.0409	0.0041
		Left	18.0711	4.9045	-14.7365	40.2102	4.9045	-14.9599	0.0113
	right2	Right	39.3106	-4.5550	1.3866	53.3481	-6.5776	-3.1513	0.0009
		Left	21.7839	9.2590	-10.8633	43.8828	11.6081	-18.0711	0.0169
	straight1	Right	21.8698	-4.2342	-17.2976	66.7037	-16.1975	-17.2976	-0.0006
		Left	29.8396	4.6352	-15.6589	62.9050	10.5940	-17.2460	-0.0046
	straight2	Right	27.5879	-5.2540	-2.0111	54.1846	-14.9198	11.2414	-0.0049
		Left	22.2651	2.1199	7.1906	54.9696	10.6799	7.1906	0.0046

Table 15 (continued, 3)

Kinematic Data Braced

Participant	Trial	Leg[a]	Knee Angle at Foot Strike (°) Flexion	Frontal[b]	Rotation[c]	Maximum Knee Angle (°) Flexion	Frontal[b]	Rotation[c]	Tibial Translation[d]
GCD32	left1	Right	18.0883	-1.1975	-6.5489	62.5269	-6.0103	6.6406	-0.0022
		Left	42.9088	-0.5042	-12.9431	75.6247	16.2720	13.1494	-0.0014
	left2	Right	19.7556	-1.4782	0.2636	53.1476	-6.0734	14.0203	0.0026
		Left	44.8053	4.8931	-18.8961	69.9524	23.0272	-19.0050	0.0014
	right1	Right	29.4157	2.8820	-0.7620	75.7450	-25.3935	14.3698	-0.0102
		Left	18.4378	4.5779	5.0592	53.8294	6.1478	-17.8763	0.0074
	right2	Right	32.0684	-0.7162	3.5065	76.8852	-25.8117	14.6677	-0.0091
		Left	19.4806	0.6360	-10.4278	56.5739	6.8468	-18.1857	0.0069
	straight1	Right	18.8618	3.3232	-9.2074	61.8221	-11.6139	-11.6654	-0.0161
		Left	25.7029	0.0172	6.9786	56.6082	16.1803	-8.8923	-0.0136
	straight2	Right	23.2449	0.4985	-6.5546	71.3848	-17.1486	8.2048	-0.0089
		Left	27.8744	-0.0859	3.2544	66.2511	14.3469	7.7235	-0.0082

Table 15 (continued, 4)

Kinematic Data Braced

Participant	Trial	Leg[a]	Knee Angle at Foot Strike (°) Flexion	Frontal[b]	Rotation[c]	Maximum Knee Angle (°) Flexion	Frontal[b]	Rotation[c]	Tibial Translation[d]
GCD35	left1	Right	14.6276	-6.7151	4.3831	66.9100	-26.9691	21.8755	0.0073
		Left	40.5654	-4.6009	-8.7949	88.0292	-18.1857	-11.6368	0.0038
	left2	Right	18.5696	-8.7834	7.0703	58.3500	-23.4855	20.5978	0.0046
		Left	50.3859	0.8365	-8.0214	83.6977	15.6074	-13.0061	0.0114
	right1	Right	26.6425	-1.8449	-4.9618	90.4700	-30.3668	8.9152	-0.0017
		Left	26.6139	-2.2746	-5.5061	73.3386	9.3679	-15.7392	0.0058
	right2	Right	21.8412	-1.6501	0.9969	91.6561	-30.9340	13.5848	0.0076
		Left	27.6509	0.2292	25.1815	77.7618	9.8663	25.1815	-0.0018
	straight1	Right	18.9248	-4.1310	-3.9706	87.7141	-26.7399	10.3705	-0.0009
		Left	30.6418	-4.1310	11.0982	87.2672	7.5057	11.0982	0.0042
	straight2	Right	17.8133	-5.2082	-9.6028	96.2626	-29.4500	45.2236	0.0073
		Left	28.1036	0.3724	2.5210	91.1347	9.3163	-18.0825	0.0054

Table 15 (continued, 5)

Kinematic Data Braced

Participant	Trial	Leg[a]	Knee Angle at Foot Strike (°) Flexion	Frontal[b]	Rotation[c]	Maximum Knee Angle (°) Flexion	Frontal[b]	Rotation[c]	Tibial Translation[d]
GCD36	left1	Right	20.5119	-3.1398	-6.8182	67.3913	-9.1158	-18.9534	-0.0063
		Left	29.9370	5.6666	-10.9721	82.4715	30.4928	-18.1914	-0.0014
	left2	Right	18.8732	-3.6841	-10.8232	66.0678	-7.9985	-14.3641	-0.0004
		Left	27.3817	0.4183	-7.2250	85.6515	25.5081	-17.1200	-0.0063
	right1	Right	29.9829	-0.9855	-6.3999	85.8405	-23.8694	-11.7113	-0.0064
		Left	26.7800	3.3862	-10.0096	71.0238	12.7483	-16.5929	-0.0009
	right2	Right	32.0856	-4.2055	-10.5825	80.0021	-18.7128	-11.8316	-0.0063
		Left	17.9336	5.2254	-4.2342	68.2450	17.2403	-15.0745	-0.0081
	straight1	Right	24.2304	2.0856	-9.1158	79.6469	3.0481	-18.3404	-0.0142
		Left	21.1593	0.0057	-5.8213	75.0517	9.5970	-12.1295	-0.0091
	straight2	Right	20.5864	0.9167	-11.7399	75.8367	-8.3308	-29.0318	-0.0098
		Left	21.0619	-1.8908	-5.0592	74.8168	12.6853	-15.2407	-0.0119

Table 15 (continued, 6)

Kinematic Data – Braced

Participant	Trial	Leg[a]	Knee Angle at Foot Strike (°) Flexion	Frontal[b]	Rotation[c]	Maximum Knee Angle (°) Flexion	Frontal[b]	Rotation[c]	Tibial Translation[d]
GCD37	left1	Right	30.4814	-13.6249	15.7621	55.7373	-21.3369	20.4832	0.0105
		Left	23.3079	-0.0688	4.3602	68.4856	20.4603	-19.5092	-0.0141
	left2	Right	32.1888	-19.3545	15.7334	52.3225	-24.8492	15.7621	0.0054
		Left	23.3251	-4.0565	3.8445	66.9215	28.7052	-22.3855	-0.0167
	right1	Right	29.8683	-0.4756	-7.2078	76.6732	-31.8049	11.9118	-0.0181
		Left	19.0050	4.9905	3.3002	54.3049	-7.0015	-12.8514	0.0035
	right2	Right	26.1784	-2.9393	-6.0791	67.6950	-31.8966	12.7082	-0.0090
		Left	12.3988	0.1203	1.2433	54.6659	-13.8770	9.2475	-0.0017
	straight1	Right	31.8966	-5.3743	-2.8705	70.4967	-21.4344	6.2567	-0.0061
		Left	31.1460	-2.4293	-0.3266	80.3516	22.9756	-24.1903	-0.0069
	straight2	Right	30.2522	-3.6154	-1.2777	68.9211	-20.7067	-9.8377	-0.0061
		Left	27.3817	-0.9225	-2.7903	74.2324	17.2174	-20.2655	0.0029

Table 15 (continued. 7)

Kinematic Data Braced

			Knee Angle at Foot Strike (°)			Maximum Knee Angle (°)			Tibial Translation[d]
Participant	Trial	Leg[a]	Flexion	Frontal[b]	Rotation[c]	Flexion	Frontal[b]	Rotation[c]	
GCD38	left1	Right	42.0093	-11.9462	11.7915	54.3450	-15.0230	15.6647	0.0007
		Left	19.4519	-7.8381	4.4118	89.2726	20.4718	-22.8324	-0.0186
	left2	Right	38.6575	-7.7693	16.0772	63.9822	-14.6448	19.6123	0.0082
		Left	25.2904	-5.7067	1.4496	95.8788	19.9447	-25.3878	-0.0095
	right1	Right	68.4513	-2.9851	2.5210	84.2420	-11.8373	3.4607	-0.0097
		Left	23.8923	0.8193	-0.9511	62.8936	18.1398	-27.3530	0.0000
	right2	Right	30.1261	-7.8724	29.7365	73.0063	-20.8270	30.6532	-0.0009
		Left	36.3255	-4.9274	-22.7063	86.2817	12.9087	-29.3412	0.0057
	straight1	Right	48.6670	-7.6834	7.1562	84.1560	-21.2968	10.6685	0.0091
		Left	35.5406	-2.9335	1.0714	91.2550	23.2220	-15.3553	0.0061
	straight2	Right	46.2492	-8.6746	13.2296	95.6553	-29.0948	14.5188	0.0134
		Left	33.9592	-2.9450	-5.3915	95.0136	25.3247	-21.9328	0.0041

Table 16

Kinetic Data Untaped

Participant	Trial	Leg[b]	Knee Moment[a] at Foot Strike (Nm) Extension[c]	Frontal[d]	Rotation[e]	Maximum Knee Moment[a] (Nm) Extension	Frontal[d]	Rotation[e]	Maximum GRF[f] (N)
GCD25	left1	Right	16.0995	33.0525	-6.6549	143.2074	51.4411	15.0177	2390.3599
		Left	37.6859	4.2597	2.1665	81.9018	10.9862	12.0061	1334.0503
	left2	Right	15.7602	36.9635	-8.2358	139.1813	-53.1554	23.6167	2306.7271
		Left	29.8240	13.4892	-0.8939	80.7609	18.1549	10.5177	1130.3828
	right1	Right	40.1432	-19.1326	0.2229	109.5047	-35.5999	-6.8123	1598.0750
		Left	44.5655	-26.5115	6.2861	141.5662	-40.0684	18.2182	2314.6785
	right3	Right	Knee moments for this trial were not computed by the motion analysis						1658.0441
		Left	program.						2815.3291
	straight1	Right	11.1889	13.1246	-4.4946	126.3563	25.4185	-12.3270	1307.1228
		Left	19.4155	-11.7262	4.0361	126.7735	-33.5894	14.9243	1332.7212
	straight2	Right	23.0939	11.4741	-2.9230	97.3605	23.8397	-5.4856	1421.3885
		Left	33.8368	-15.5849	5.3339	111.9753	-26.4483	12.9570	1440.1167

Table 16 (continued)

Kinetic Data Untaped

Participant	Trial	Leg[b]	Knee Moment[a] at Foot Strike (Nm)			Maximum Knee Moment[a] (Nm)			Maximum GRF[f] (N)
			Extension[c]	Frontal[d]	Rotation[e]	Extension	Frontal[d]	Rotation[e]	
GCD30	left1	Right	22.9508	8.8176	-0.4221	88.0322	23.7928	-10.1975	1369.5480
		Left	25.7847	6.0539	-0.6974	83.1468	13.3689	-10.8084	1031.9612
	left2	Right	28.9981	-3.5510	0.0106	116.2892	22.6838	-4.2060	1400.6300
		Left	33.4852	-1.1761	3.8639	84.7074	15.5332	-15.4173	939.9356
	right1	Right	16.6785	9.1144	-2.2376	82.3703	36.4909	-16.9058	1173.5590
		Left	16.9174	-1.8477	1.4620	98.1055	33.2424	-7.7672	1902.0238
	right2	Right	30.9154	3.9032	-0.2929	92.0012	39.4957	-17.2600	1080.5005
		Left	17.7801	-2.0634	2.3303	83.2588	45.5846	-14.4842	1407.2455
	straight1	Right	26.2927	-0.7871	-0.7224	94.2571	21.7442	-6.7945	1270.9458
		Left	19.8584	3.6653	0.3964	87.4357	14.2953	-3.7225	1315.3204
	straight2	Right	4.8030	7.0028	-2.0779	97.6408	40.3533	-9.0153	1333.6987
		Left	18.6349	7.4018	-1.5333	110.8256	-15.4856	3.8564	1456.8646

Table 16 (continued, 2)

Kinetic Data Untaped

Participant	Trial	Leg[b]	Knee Moment[a] at Foot Strike (Nm) Extension[c]	Frontal[d]	Rotation[e]	Maximum Knee Moment[a] (Nm) Extension	Frontal[d]	Rotation[e]	Maximum GRF[f] (N)
GCD31	left1	Right	41.2873	-5.3086	1.3327	143.0621	-37.4478	17.6208	2637.0391
		Left	19.0986	-4.3432	0.4072	37.0342	38.2945	-13.4369	655.7858
	left2	Right	25.1043	24.6527	-2.8832	112.9882	37.1753	-15.1841	2433.4585
		Left	10.8170	3.2426	-1.3664	75.7543	34.9199	-21.0971	999.3138
	right2	Right	12.3801	8.6264	-2.7788	35.3919	30.0173	-16.6203	805.8865
		Left	8.0003	-0.9139	0.5121	86.9140	25.0167	6.6169	2612.8315
	right3	Right	19.3593	14.7761	-5.8552	45.2686	-28.5605	-17.6980	797.8167
		Left	11.0738	-14.8832	0.4646	74.8003	28.9843	-8.1682	2389.4651
	straight1	Right	39.9095	4.6279	-0.4743	119.1142	30.9042	9.3123	1251.5502
		Left	29.1757	1.5519	-1.8797	113.0666	9.2735	-15.9139	1762.6067
	straight2	Right	15.5934	9.4198	0.0162	73.2063	36.3361	26.2833	1129.0122
		Left	15.4066	0.9930	-1.3648	97.2581	33.5520	-23.2373	1692.0278

Table 16 (continued, 3)

Kinetic Data Untaped

Participant	Trial	Leg[b]	Knee Moment[a] at Foot Strike (Nm)			Maximum Knee Moment[a] (Nm)			Maximum GRF[f] (N)
			Extension[c]	Frontal[d]	Rotation[e]	Extension	Frontal[d]	Rotation[e]	
GCD32	left1	Right	16.9697	17.8939	-2.4484	164.2329	29.9180	13.5002	2785.3105
		Left	20.6880	-13.9566	11.1157	53.1202	-35.1464	30.1014	655.9967
	left2	Right	11.9226	8.0870	-1.0797	157.4689	31.1814	10.9370	2386.7532
		Left	29.9315	-6.4189	2.8309	55.9511	-32.1305	14.0727	699.4007
	right1	Right	23.6597	0.5315	-0.4613	82.0914	54.3252	-9.5260	767.9363
		Left	28.9754	-8.3494	2.1947	142.2388	-34.8755	-15.3811	2392.3604
	right2	Right	27.0021	-0.2157	-0.9683	95.7988	17.0877	17.3491	1756.8785
		Left	17.0504	-3.4259	1.1581	96.2451	27.8484	-15.8711	1456.8778
	straight1	Right	-1.8665	-0.5163	0.2230	126.7828	22.6147	6.7643	2230.1091
		Left	17.4463	3.5119	-0.0345	108.2106	-19.7928	-3.5576	1535.0414
	straight2	Right	38.4687	6.1606	-1.8872	108.8709	32.7932	-3.5435	1503.7728
		Left	11.1252	-8.5216	2.3302	124.7578	-17.9413	-5.5066	2132.2183

Table 16 (continued, 4)

Kinetic Data Untaped

Participant	Trial	Leg[b]	Knee Moment[a] at Foot Strike (Nm) Extension[c]	Frontal[d]	Rotation[e]	Maximum Knee Moment[a] (Nm) Extension	Frontal[d]	Rotation[e]	Maximum GRF[f] (N)
GCD35	left1	Right	24.9545	17.3160	-1.7140	143.8313	102.2984	-19.8698	2039.3351
		Left	16.6791	-9.5813	9.4517	95.8427	-41.2778	31.5662	541.7007
	left2	Right	22.7768	19.8638	0.2844	153.2035	58.7194	-14.3092	2573.7686
		Left	4.9383	-14.7715	14.4088	69.9253	-45.9320	40.9245	597.9131
	right1	Right	5.3306	14.9744	-3.7031	56.6244	84.9441	-31.7114	689.0587
		Left	28.7456	-30.8241	6.4130	139.5348	-44.8276	19.4635	2433.5786
	right2	Right	0.3413	14.6541	-2.3662	51.0103	50.1242	-18.5001	548.5718
		Left	36.4017	-15.3511	4.3980	36.4017	-15.3511	4.3980	2045.4484
	straight1	Right	31.7053	11.4669	1.4752	75.3909	51.7492	-9.9239	1098.9431
		Left	44.0654	-15.1020	7.7320	75.0542	-30.5360	16.6668	1068.4866
	straight2	Right	12.7552	11.7178	-1.1500	88.8926	51.7744	-15.6675	1161.7368
		Left	35.6853	2.5656	3.1593	75.9957	-20.2048	11.4197	983.2354

Table 16 (continued, 5)

Kinetic Data Untaped

Participant	Trial	Leg[b]	Knee Moment[a] at Foot Strike (Nm) Extension[c]	Frontal[d]	Rotation[e]	Maximum Knee Moment[a] (Nm) Extension	Frontal[d]	Rotation[e]	Maximum GRF[f] (N)
GCD36	left1	Right	20.6748	9.6820	-0.0419	118.3650	32.0943	-17.6402	1568.2701
		Left	41.0574	-11.5637	5.3735	72.9702	-30.1599	15.9741	745.1043
	left2	Right	7.0438	17.4771	-3.3083	129.6986	30.8162	-16.2536	1559.0577
		Left	31.1857	-7.4781	5.6744	68.1679	-34.5443	19.6979	643.6389
	right2	Right	2.9093	12.7263	-2.8122	80.6552	23.8669	-21.2377	657.2676
		Left	18.5090	1.5432	0.7832	98.9178	-23.8082	19.5798	1250.8607
	right3	Right	5.6852	7.3505	-3.5794	87.4414	23.5359	-18.0737	686.5885
		Left	18.0191	-3.4866	0.9661	112.1607	-22.4832	12.5561	1857.5297
	straight1	Right	3.7168	11.4866	-2.0940	94.0476	17.2390	-13.0345	885.8764
		Left	5.8833	-9.3270	1.9644	82.5878	-14.8322	13.8525	1006.9300
	straight2	Right	-24.9200	9.0039	0.6732	73.8577	-12.0941	5.9703	1118.4385
		Left	-23.1323	-3.0409	-1.5169	69.0639	-10.8500	9.1563	1294.4899

Table 16 (continued, 6)

Kinetic Data Untaped

Participant	Trial	Leg[b]	Knee Moment[a] at Foot Strike (Nm)			Maximum Knee Moment[a] (Nm)			Maximum GRF[f] (N)
			Extension[c]	Frontal[d]	Rotation[e]	Extension	Frontal[d]	Rotation[e]	
GCD37	left1	Right	7.9221	18.2804	-5.4264	159.8613	104.1751	-9.0876	1982.9856
		Left	40.1525	-5.5495	1.4390	111.8562	-39.7302	10.7667	772.8221
	left2	Right	12.4876	20.5779	-6.0842	131.9769	73.2340	10.8192	1633.2698
		Left	46.1592	-6.4004	3.1398	139.4341	-60.4271	-14.1679	1403.3474
	right1	Right	28.4813	10.8889	-1.4906	142.2681	88.4455	-15.6328	1183.5428
		Left	19.9444	-16.1802	2.1017	132.1805	-37.7511	-18.7021	1597.3427
	right2	Right	32.7083	25.4971	-6.9583	86.1061	53.8178	-13.7660	890.3921
		Left	18.1204	-9.1933	2.2413	161.6121	31.6659	-26.5967	1954.3422
	straight1	Right	2.1059	12.3118	-5.8316	97.1298	57.5225	-14.6944	886.8292
		Left	34.1035	-13.4431	2.0339	137.9902	-54.6052	11.9123	1509.8605
	straight2	Right	-6.8526	4.6633	-3.7657	105.9196	67.3974	-11.6271	1115.9495
		Left	38.6145	-9.0226	2.4381	146.3883	-58.9597	6.5985	1715.9349

Table 16 (continued, 7)

Kinetic Data - Untaped

Participant	Trial	Leg[b]	Knee Moment[a] at Foot Strike (Nm)			Maximum Knee Moment[a] (Nm)			Maximum GRF[f] (N)
			Extension[c]	Frontal[d]	Rotation[e]	Extension	Frontal[d]	Rotation[e]	
GCD38	left1	Right	18.1261	4.1276	-1.4489	123.4598	45.0243	-14.1571	1472.7378
		Left	15.8588	-4.3521	2.9511	55.7150	33.4458	-28.1085	689.4023
	left2	Right	12.7088	15.5162	-9.2587	92.4579	16.4235	13.8666	1537.2037
		Left	25.9553	-1.3941	2.1974	64.7379	-27.6675	-18.6847	759.3351
	right1	Right	16.3620	6.6623	-5.3571	93.4666	-36.4945	33.7695	791.7659
		Left	23.3566	-6.8959	1.3035	107.4337	-42.1273	12.2867	1599.6357
	right2	Right	21.2762	7.3343	-8.5084	81.8755	-37.4277	25.9663	858.3807
		Left	19.7275	-12.3731	3.0381	140.7126	-54.0326	10.2948	1889.5588
	straight1	Right	15.9273	6.4403	-5.9740	81.4533	21.4488	-6.4891	777.7458
		Left	13.1159	-6.7442	2.4617	74.9754	-40.3008	11.1930	823.8690
	straight2	Right	17.7406	4.0058	-4.2718	98.0517	25.8772	-6.3834	770.7550
		Left	25.7266	-3.1440	3.7919	85.4081	-41.9757	11.8624	736.2459

Table 17

Kinetic Data - Taped

Participant	Trial	Leg[b]	Knee Moment[a] at Foot Strike (Nm)			Maximum Knee Moment[a] (Nm)			Maximum GRF[f] (N)
			Extension[c]	Frontal[d]	Rotation[e]	Extension	Frontal[d]	Rotation[e]	
GCD25	left1	Right	31.2550	27.4145	-3.8023	158.5128	-36.6908	21.2042	2576.7544
		Left	49.0744	18.5918	-0.9959	99.8985	26.4641	-16.6703	1436.2584
	left2	Right	33.0134	43.2535	-8.3470	213.3318	77.8214	17.0182	2244.3835
		Left	23.9596	6.9676	0.3490	87.6415	8.5061	11.5925	969.5631
	right1	Right	43.1994	-2.2070	4.5378	119.0465	14.5661	8.5134	1576.0631
		Left	23.9316	-25.2114	3.3712	116.9185	-49.2714	19.8211	2458.2542
	right2	Right	51.9894	-16.7184	3.6517	107.8157	-22.3476	5.7970	1305.3123
		Left	14.2581	-24.7741	7.1726	140.2682	47.1123	21.4845	2636.9343
	straight1	Right	47.6370	12.2966	-1.9036	136.7407	26.2597	7.0326	1539.5162
		Left	79.8115	-13.5420	1.7852	170.4277	-41.0841	-7.0810	2088.3328
	straight2	Right	36.3338	11.9585	-4.0195	137.9577	32.1545	-5.8458	1409.0437
		Left	50.1646	-28.2784	8.7208	132.5616	-39.9267	10.8830	1686.8730

Table 17 (continued)

Kinetic Data - Taped

Participant	Trial	Leg[b]	Knee Moment[a] at Foot Strike (Nm)			Maximum Knee Moment[a] (Nm)			Maximum GRF[f] (N)
			Extension[c]	Frontal[d]	Rotation[e]	Extension	Frontal[d]	Rotation[e]	
GCD30	left1	Right	45.3545	1.9336	1.6690	125.4826	33.7102	9.2783	2143.4189
		Left	31.4421	0.0472	-0.2484	68.2377	-23.2995	-16.8904	1116.1105
	left2	Right	45.3623	5.7426	1.1343	133.2403	25.7140	11.7871	2020.4514
		Left	30.8203	-15.2707	8.0387	77.1959	-34.1295	-11.2110	1017.6863
	right1	Right	21.5929	-0.0313	0.3551	95.7566	30.4584	-18.4312	972.9223
		Left	46.9440	-12.7158	3.7096	115.0124	-13.5409	10.1270	2030.0707
	right2	Right	37.0002	3.1324	1.6634	84.2049	30.6205	13.9211	1125.6151
		Left	46.2808	-5.8188	0.7317	115.9835	-14.3249	-10.6726	2036.9951
	straight1	Right	34.9043	7.8208	0.5353	101.8730	58.7319	-10.6878	1428.3904
		Left	-10.8455	-24.3221	12.0460	91.2705	-29.6446	15.1450	1527.0583
	straight2	Right	45.0457	0.4212	0.7154	119.5952	59.5680	-10.9930	1889.8540
		Left	34.8681	1.5840	-0.4042	104.1432	-26.3859	3.3393	1896.0959

Table 17 (continued, 2)

Kinetic Data Taped

Participant	Trial	Leg[b]	Knee Moment[a] at Foot Strike (Nm)			Maximum Knee Moment[a] (Nm)			Maximum GRF[f] (N)
			Extension[c]	Frontal[d]	Rotation[e]	Extension	Frontal[d]	Rotation[e]	
GCD31	left1	Right	27.3754	13.1349	-1.7636	131.3609	25.4907	-8.8950	2598.1021
		Left	19.2723	-7.2042	3.4919	40.2983	17.8418	11.0034	767.5833
	left2	Right	24.3008	12.6750	-2.7408	125.8374	20.5100	-9.3205	2406.9976
		Left	28.2018	-4.9909	1.6107	54.0874	19.2054	-11.5529	864.4326
	right1	Right	11.1437	6.6352	-1.8568	41.8084	34.5779	-12.2729	1450.7217
		Left	9.1358	-15.0548	1.8685	81.0427	-17.7766	6.3271	2744.6162
	right2	Right	14.5352	10.8245	-2.7086	83.9323	-30.2194	20.0613	1445.0948
		Left	18.9884	-3.4259	0.3039	78.1021	-16.0611	-7.3894	2318.2969
	straight1	Right	20.0021	5.3246	-0.7170	90.1985	-30.0776	14.1746	1425.9048
		Left	57.6135	5.1727	-3.0344	91.9867	29.9176	-16.6803	1862.0983
	straight2	Right	45.7989	11.5818	-0.2714	83.3052	-33.9832	17.0810	1530.5226
		Left	53.0670	8.1576	-3.4951	95.6353	35.7771	-13.6395	1734.1693

Table 17 (continued, 3)

Kinetic Data Taped

Participant	Trial	Leg[b]	Knee Moment[a] at Foot Strike (Nm)			Maximum Knee Moment[a] (Nm)			Maximum GRF[f] (N)
			Extension[c]	Frontal[d]	Rotation[e]	Extension	Frontal[d]	Rotation[e]	
GCD32	left1	Right	18.0913	13.6810	-3.8382	158.4932	35.2904	15.4503	2589.8018
		Left	26.0280	6.1392	-0.1977	82.8441	25.9722	-11.3087	790.6862
	left2	Right	10.0938	14.2892	-2.5066	160.1978	32.9249	19.9718	3071.4709
		Left	6.5724	-7.7240	5.8394	71.6511	-26.9891	21.2614	793.8388
	right1	Right	42.7969	-9.4048	4.4212	80.3547	32.4705	7.1254	994.0701
		Left	33.5570	-12.5217	0.8233	139.7199	35.8917	-20.3384	2393.5725
	right2	Right	21.3377	-6.7927	3.5003	96.2336	39.2663	8.1159	956.7839
		Left	7.9767	-16.8710	4.2602	148.6174	-28.5997	-20.1180	1969.6117
	straight1	Right	21.6964	3.2955	-0.2976	128.1558	32.0359	13.3197	2303.2275
		Left	21.6011	-3.9666	1.0590	84.5993	-19.6139	7.4126	1171.2355
	straight2	Right	52.2599	-7.6302	3.6516	98.6095	16.7727	10.8229	1504.9392
		Left	24.3166	3.1121	-0.7219	72.8989	-5.2054	-7.9673	1511.5907

Table 17 (continued. 4)

Kinetic Data - Taped

Participant	Trial	Leg[b]	Knee Moment[a] at Foot Strike (Nm)			Maximum Knee Moment[a] (Nm)			Maximum GRF[f] (N)
			Extension[c]	Frontal[d]	Rotation[e]	Extension	Frontal[d]	Rotation[e]	
GCD35	left1	Right	25.9496	13.2714	-0.9208	118.6421	45.7498	-13.4884	2739.4578
		Left	16.7150	-11.6308	10.4209	78.2115	-31.8448	29.7491	615.3849
	left2	Right	39.3920	12.0173	-1.4603	132.4903	50.9041	-15.1460	2144.7136
		Left	33.9982	-14.4771	11.8342	96.6830	-37.1339	30.6180	559.3179
	right1	Right	26.5735	17.6309	-3.6422	89.1779	68.4131	-26.1821	1100.4425
		Left	19.0861	1.2953	1.6772	115.9820	-23.1194	10.4220	1757.0876
	right2	Right	19.1339	7.2397	-1.6460	64.4527	61.6168	-24.7614	804.7402
		Left	52.7408	-15.0944	5.5761	137.6941	-28.2309	17.2403	2131.3376
	straight1	Right	12.3785	3.8346	-0.1954	106.3748	64.2125	-15.1299	1270.5444
		Left	25.8838	-9.5854	6.0457	97.3082	-15.5792	13.9825	853.8127
	straight2	Right	20.0954	6.4877	-0.6967	97.0893	57.5972	-17.6619	1043.1439
		Left	26.5571	-19.9093	10.1611	94.6710	-27.3787	21.4588	1036.7389

Table 17 (continued, 5)

Kinetic Data - Taped

Participant	Trial	Leg[b]	Knee Moment[a] at Foot Strike (Nm)			Maximum Knee Moment[a] (Nm)			Maximum GRF[f] (N)
			Extension[c]	Frontal[d]	Rotation[e]	Extension	Frontal[d]	Rotation[e]	
GCD36	left1	Right	27.8378	13.3703	-3.8702	111.4873	-28.2707	15.7752	1155.3995
		Left	38.1172	-5.1086	1.0040	74.1617	-35.8857	18.2184	833.7001
	left2	Right	32.5843	14.8982	-1.8003	137.2602	44.7806	-17.5725	1869.3101
		Left	30.9548	-15.0503	3.7750	84.8661	-50.8556	20.8627	811.0284
	right1	Right	19.3426	15.3305	-3.3069	50.6102	30.5017	-12.4697	790.9645
		Left	26.0699	-3.2881	0.0507	118.3915	-16.3865	7.3307	1767.0656
	right2	Right	0.6998	14.6140	-3.5127	69.6763	38.4721	-20.8587	854.3664
		Left	32.2794	-14.0121	1.4667	131.8995	-30.3000	9.6681	1751.1715
	straight1	Right	5.4764	9.9159	-2.4876	78.2974	25.5550	-8.3504	694.8973
		Left	-3.1273	-2.1482	0.0897	71.3938	-12.5920	5.1859	1223.5717
	straight2	Right	35.1731	15.6317	-4.7174	110.3897	31.8277	-14.0187	1035.3201
		Left	30.8431	-9.4639	3.8289	96.4598	10.4029	-7.3968	1166.1255

Table 17 (continued, 6)

Kinetic Data Taped

Participant	Trial	Leg[b]	Knee Moment[a] at Foot Strike (Nm)			Maximum Knee Moment[a] (Nm)			Maximum GRF[f] (N)
			Extension[c]	Frontal[d]	Rotation[e]	Extension	Frontal[d]	Rotation[e]	
GCD37	left1	Right	-10.7716	15.8284	-7.9249	119.9337	50.4854	15.4153	1864.6348
		Left	52.4432	-6.3577	-0.2184	132.7148	-50.1070	-8.2470	1219.6987
	left2	Right	9.5215	24.5873	-10.3962	159.7046	75.8425	17.0164	1866.9919
		Left	51.0807	-2.8616	3.1828	132.2790	-42.7008	-8.9567	1290.6616
	right1	Right	59.1943	41.2373	-6.9104	118.5645	103.6797	-19.9775	1124.9750
		Left	33.5299	-24.8073	-4.7404	143.3425	27.7759	-12.9127	1753.2429
	right2	Right	54.5863	6.2689	-0.7032	142.1645	78.5464	-13.0156	1222.1046
		Left	36.1368	-22.2610	2.2119	124.1017	-27.8581	-11.0391	1760.9789
	straight1	Right	11.3459	6.3288	-1.9816	124.7497	68.4062	-9.6801	1177.5996
		Left	70.8453	1.0319	1.4612	151.8445	-38.6707	-9.9215	1342.0894
	straight2	Right	17.2301	8.5866	-1.1062	111.4275	66.3046	-12.8413	1153.0011
		Left	65.5410	-9.4162	2.5964	140.8486	-63.4253	9.0675	1508.7423

Table 17 (continued. 7)

Kinetic Data Taped

Participant	Trial	Leg[b]	Knee Moment[a] at Foot Strike (Nm)			Maximum Knee Moment[a] (Nm)			Maximum GRF[f] (N)
			Extension[c]	Frontal[d]	Rotation[c]	Extension	Frontal[d]	Rotation[c]	
GCD38	left1	Right	16.3877	4.7600	-0.9058	99.4964	39.6526	25.8959	1284.8311
		Left	7.5556	-3.5540	2.0493	62.0934	-10.0811	-5.9776	446.2316
	left2	Right	-1.2882	18.7798	-16.3744	134.1050	32.0562	-21.0604	1404.1873
		Left	28.8752	8.1686	-0.0526	102.9735	-21.1331	-6.3788	874.8521
	right1	Right	27.4010	1.7734	1.3353	57.2273	14.7240	3.1954	656.1290
		Left	16.8688	-57.9758	13.0686	83.4905	-77.6249	-33.0112	586.8839
	right2	Right	46.9186	-1.3018	-0.9316	86.6885	22.9555	1.8658	1167.6718
		Left	-0.7380	-16.2778	5.4312	90.4983	-26.2075	-15.6272	994.9138
	straight1	Right	29.0317	8.7326	-1.9486	83.4470	32.7468	-4.4842	1421.0549
		Left	-11.8934	-9.1488	1.3140	70.5265	-11.6097	4.2127	733.5176
	straight2	Right	41.5003	6.3696	-3.3864	74.8163	31.5350	-5.4241	1319.2648
		Left	14.4559	-4.2911	3.3061	75.8744	-26.7095	5.8875	564.7505

Table 18

Kinetic Data Braced

Participant	Trial	Leg[b]	Knee Moment[a] at Foot Strike (Nm) Extension[c]	Frontal[d]	Rotation[e]	Maximum Knee Moment[a] (Nm) Extension	Frontal[d]	Rotation[e]	Maximum GRF[f] (N)
GCD25	left1	Right	2.5123	18.2797	-3.7746	120.8785	29.5560	19.2090	1994.1469
		Left	16.5871	14.9366	-1.8933	81.4076	28.5966	-4.6501	1356.0549
	left2	Right	11.6431	38.0599	-9.4116	146.5221	52.5319	22.8069	2290.0278
		Left	29.9739	9.5910	-0.4153	94.6246	15.3110	6.5041	1608.3779
	right1	Right	47.4670	-36.8308	4.9058	109.1078	-42.0986	7.1602	1725.0491
		Left	16.6694	-31.1651	7.9450	150.8884	-44.9653	18.4249	2580.9087
	right2	Right	27.4187	-22.4647	3.4913	101.3247	-30.7366	8.7264	1781.8406
		Left	28.3913	-18.4885	5.0731	153.5057	47.3697	-17.8672	2548.9084
	straight1	Right	14.0527	8.6128	-2.8132	127.9156	19.1874	-8.3539	1122.3574
		Left	25.5072	-6.9830	2.6310	113.9332	-22.7143	13.5758	1478.1276
	straight2	Right	21.7641	-4.7029	0.3490	94.4856	-14.5373	7.6451	1479.8362
		Left	28.8641	-9.0925	3.5416	94.6989	-17.2183	13.7373	1467.6525

Table 18 (continued)

Kinetic Data – Braced

Participant	Trial	Leg[b]	Knee Moment[a] at Foot Strike (Nm)			Maximum Knee Moment[a] (Nm)			Maximum GRF[f] (N)
			Extension[c]	Frontal[d]	Rotation[e]	Extension	Frontal[d]	Rotation[e]	
GCD30	left1	Right	24.7579	21.8573	-3.7520	76.2310	-2.6511	4.7931	2526.6160
		Left	41.1295	-15.4520	12.0571	12.3966	6.6995	-12.8322	1038.1768
	left2	Right	25.2281	11.4772	-1.8330	103.3800	19.5225	10.4461	1608.5836
		Left	12.4000	3.9593	0.0148	72.1940	7.5660	-12.2575	1394.4153
	right1	Right	30.1643	13.9565	-4.3412	49.4963	28.4249	-12.0384	1299.5419
		Left	7.6546	-18.3977	3.5380	81.9053	39.6450	-8.4701	1938.1130
	right2	Right	28.4796	17.1211	-4.9264	88.7528	39.7617	-14.4527	954.6335
		Left	16.8936	-5.7725	3.5205	100.5517	29.1580	3.8462	1815.1666
	straight1	Right	40.1320	16.1651	-3.9189	98.9690	45.1242	-12.2413	1541.1122
		Left	26.7874	-4.8152	3.7402	89.5462	-14.4015	5.1242	1631.5414
	straight2	Right	43.2628	14.5209	-2.5824	94.1028	51.5535	-5.9800	1772.3942
		Left	42.0115	1.5152	1.4518	105.9909	-26.9194	8.1883	1823.7267

Table 18 (continued, 2)

Kinetic Data Braced

Participant	Trial	Leg[b]	Knee Moment[a] at Foot Strike (Nm)			Maximum Knee Moment[a] (Nm)			Maximum GRF[f] (N)
			Extension[c]	Frontal[d]	Rotation[e]	Extension	Frontal[d]	Rotation[e]	
GCD31	left1	Right	34.1310	30.3103	-4.7371	146.7203	50.2290	-10.0526	2618.9021
		Left	6.9849	-4.3517	1.9261	63.2863	32.4325	-13.8876	1229.3854
	left2	Right	30.0842	7.4021	-0.2792	148.4142	30.8679	17.2802	2703.7678
		Left	18.3114	-6.0587	1.8837	53.7118	19.6053	-11.8701	640.8115
	right1	Right	13.4324	10.2561	-4.7504	36.8113	-36.8977	-18.2871	942.0213
		Left	12.1335	-5.8609	0.7996	88.9072	28.7598	-8.0618	2622.2268
	right2	Right	21.1656	26.0357	-11.5844	21.1656	34.8847	-18.5574	659.7910
		Left	20.7743	-1.4391	-1.6022	106.7581	-10.6637	-14.4640	2494.9485
	straight1	Right	7.9128	6.8821	-1.1567	131.5240	27.4718	18.2851	1750.7909
		Left	-3.0556	-10.4696	3.6750	90.0759	15.1476	-7.9958	857.9050
	straight2	Right	-2.0218	-10.7475	3.2886	66.6194	-18.3433	8.1104	1454.5443
		Left	-6.1246	-48.0073	-5.4354	77.0553	-54.8143	10.9603	1801.3405

Table 18 (continued, 3)

Kinetic Data Braced

Participant	Trial	Leg[b]	Knee Moment[a] at Foot Strike (Nm)			Maximum Knee Moment[a] (Nm)			Maximum GRF[f] (N)
			Extension[c]	Frontal[d]	Rotation[e]	Extension	Frontal[d]	Rotation[e]	
GCD32	left1	Right	22.0050	14.7128	-3.0003	160.4021	23.3398	14.3885	2349.0654
		Left	12.5640	-2.2088	1.7655	72.6169	-31.0062	12.0814	723.6220
	left2	Right	28.1928	12.3036	-1.8115	177.5282	-37.4202	11.0706	2485.1191
		Left	8.0189	-5.0178	3.0652	54.5981	-40.2251	19.6940	704.2911
	right1	Right	55.6292	-6.9567	0.1460	75.4998	40.0324	-6.3379	1190.7036
		Left	20.6143	-11.4020	2.8904	152.5326	-25.2595	-15.2884	2191.5950
	right2	Right	59.8895	-2.1580	1.4934	76.8693	36.5088	3.7639	1154.7295
		Left	17.9004	-15.1110	2.5998	135.1648	-29.3381	-11.8609	1721.1034
	straight1	Right	29.2425	-8.2112	-0.2201	126.4940	19.3745	7.7576	1714.5654
		Left	53.1584	-7.3024	3.7696	124.5690	-37.4787	6.1214	1660.9651
	straight2	Right	46.2607	-3.3872	0.0428	118.9230	32.2455	3.9870	1746.6460
		Left	33.6908	-8.0744	3.0232	108.9886	-30.0375	6.2583	1592.3956

Table 18 (continued, 4)

Kinetic Data – Braced

Participant	Trial	Leg[b]	Knee Moment[a] at Foot Strike (Nm)			Maximum Knee Moment[a] (Nm)			Maximum GRF[f] (N)
			Extension[c]	Frontal[d]	Rotation[e]	Extension	Frontal[d]	Rotation[e]	
GCD35	left1	Right	41.3338	13.0198	0.8293	135.5710	75.6567	-21.2835	1914.8761
		Left	16.3693	-4.6295	4.4151	90.8226	-26.6975	22.6098	553.0262
	left2	Right	39.3747	27.4892	-3.5476	144.8391	77.1279	-20.0936	2696.3022
		Left	16.7860	-8.9823	7.6973	88.8706	-43.5610	32.5784	568.7045
	right1	Right	41.4170	7.9017	-1.7228	76.8978	58.7361	-15.4183	1127.5433
		Left	28.5177	-4.5778	2.5609	124.6095	-37.7117	20.2710	1314.1185
	right2	Right	21.9698	5.8282	-1.3879	85.6014	65.0371	-16.2573	595.3453
		Left	45.2359	-12.8885	4.5518	133.8129	-42.1839	23.6933	1924.8856
	straight1	Right	41.1386	10.1228	0.6428	78.1303	41.8386	-21.4446	1178.6273
		Left	19.2730	-2.8012	1.9872	83.0633	-18.4172	13.0609	791.1356
	straight2	Right	30.6937	11.1700	0.1904	68.7724	43.4361	-17.8370	1011.9670
		Left	12.6176	-10.9763	3.4480	78.9916	-27.0397	16.4700	794.2550

Table 18 (continued, 5)

Kinetic Data Braced

Participant	Trial	Leg[b]	Knee Moment[a] at Foot Strike (Nm)			Maximum Knee Moment[a] (Nm)			Maximum GRF[f] (N)
			Extension[c]	Frontal[d]	Rotation[e]	Extension	Frontal[d]	Rotation[e]	
GCD36	left1	Right	30.7721	12.0536	-2.7360	131.4339	29.7509	-15.7428	1974.0837
		Left	41.9128	-8.5581	0.9020	76.1440	-48.7153	21.5974	851.6504
	left2	Right	24.4492	12.5446	-1.1460	134.8266	30.8158	-14.2146	1696.2814
		Left	12.2065	-2.8392	1.0533	75.5999	-33.6529	15.5199	660.1957
	right1	Right	31.5385	7.7042	-2.8055	73.4020	27.8457	-12.9088	742.3994
		Left	51.4943	-6.6901	1.1243	126.1641	-27.4815	11.9199	1399.0927
	right2	Right	36.1694	21.3497	-6.3469	88.5322	37.9219	-22.5529	782.6240
		Left	46.0474	-13.9416	1.2766	136.8525	-47.0474	11.4679	1391.8635
	straight1	Right	6.6102	5.9406	-1.8388	77.4660	8.2879	-8.5097	1132.1921
		Left	4.9537	-8.2691	2.1907	62.1365	-13.0368	8.8624	1498.4955
	straight2	Right	7.4546	6.6804	-2.0251	91.2360	25.3529	-10.6769	1010.4301
		Left	20.2333	-1.9702	2.2243	93.1673	-18.8610	10.7852	963.4189

Table 18 (continued, 6)

Kinetic Data Braced

Participant	Trial	Leg[b]	Knee Moment[a] at Foot Strike (Nm)			Maximum Knee Moment[a] (Nm)			Maximum GRF[f] (N)
			Extension[c]	Frontal[d]	Rotation[e]	Extension	Frontal[d]	Rotation[e]	
GCD37	left1	Right	19.6366	22.5142	-5.7168	141.8482	72.6055	14.2212	1524.0211
		Left	48.3043	1.4372	-0.1172	138.7567	-41.7609	-10.1198	1509.2474
	left2	Right	5.1240	22.6389	-9.8666	146.1043	95.0634	-14.7430	1856.8491
		Left	51.7657	9.2370	1.2440	131.6409	-55.4205	13.5607	1407.9540
	right1	Right	33.0881	9.5064	-1.7613	157.0807	84.6750	-16.0477	1046.3164
		Left	24.7817	-16.5098	2.2011	149.6155	33.5412	-16.4084	1598.0397
	right2	Right	24.5661	3.6506	0.5574	150.6796	88.3759	-10.5325	1256.4946
		Left	29.3281	-1.4445	0.2026	125.0852	51.3708	-24.4087	1572.6124
	straight1	Right	25.4888	5.5221	0.8484	111.1493	48.4956	6.3508	1592.3662
		Left	44.5607	1.5285	1.5234	127.1681	-56.7811	12.6426	1392.1174
	straight2	Right	3.7747	3.7509	-0.7121	96.5202	54.6897	-9.8793	1030.7551
		Left	38.6201	-2.4671	0.7918	146.0948	-45.5023	9.5769	1432.3253

Table 18 (continued, 7)

Kinetic Data – Braced

Participant	Trial	Leg[b]	Knee Moment[a] at Foot Strike (Nm)			Maximum Knee Moment[a] (Nm)			Maximum GRF[f] (N)
			Extension[c]	Frontal[d]	Rotation[e]	Extension	Frontal[d]	Rotation[e]	
GCD38	left1	Right	-1.9914	11.6556	-8.6966	114.2356	27.8571	17.2275	1186.9790
		Left	24.6793	12.7333	0.1989	91.8041	-29.1155	-5.8274	815.4446
	left2	Right	15.8424	14.7404	-8.6895	113.3598	23.0860	15.6794	1215.8184
		Left	19.2054	9.3445	-0.6765	77.9410	-30.5883	-17.6439	738.5194
	right1	Right	20.8099	10.1243	-9.9287	53.6787	-42.0086	30.3523	785.0908
		Left	32.3340	2.8555	0.1743	124.7747	-43.2660	12.8956	1709.2037
	right2	Right	15.5241	17.3096	-8.0362	69.1478	17.3540	-9.1016	1415.0538
		Left	32.9183	4.2650	1.4159	92.3644	-15.1806	13.0547	720.8112
	straight1	Right	-4.5750	7.1089	-5.8926	80.0301	25.9182	9.7042	858.8311
		Left	24.1197	-1.0703	0.9953	75.3749	-37.0210	11.6215	933.6990
	straight2	Right	21.0711	10.8091	-6.7887	92.2226	46.3759	-8.0495	1111.5337
		Left	24.2594	-5.7475	3.0691	69.8551	-32.6776	12.4645	729.4382