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The relationship between static postural measurements and biomechanical characteristics during landing

James Craig Garrison
University of Tennessee

To the Graduate Council:

I am submitting herewith a thesis written by James Craig Garrison entitled "The relationship between static postural measurements and biomechanical characteristics during landing." I have examined the final electronic copy of this thesis for form and content and recommend that it be accepted in partial fulfillment of the requirements for the degree of Master of Science, with a major in Human Performance and Sport Studies.

Songning Zhang, Major Professor

We have read this thesis and recommend its acceptance:

Wendell Liemohn, Lorna Swanson

Accepted for the Council:
Carolyn R. Hodges

Vice Provost and Dean of the Graduate School

(Original signatures are on file with official student records.)

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The relationship between static postural measurements and biomechanical characteristics during landing

A Thesis
Presented for the
Master of Science
Degree
The University of Tennessee, Knoxville

James Craig Garrison
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DEDICATION

I would like to dedicate this thesis to my wife, Laura Garrison, and my parents, John and Dianna Garrison. Laura, thank you for your support and understanding during these last two years of study. I look forward to our future and even greater achievements. Mom and dad, thanks for instilling within me a faith in God and a desire to pursue my goals. I am thankful to have such a wonderful wife and family.

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ABSTRACT

The purpose of this study was to examine the relationship between static postural measurements of the lower extremity and biomechanical characteristics during landing. Thirteen healthy and active subjects performed five trials of drop landing from three different heights (45, 60, and 75 cm) with two different landing techniques (soft and stiff). Standing static knee extension of the right side was measured in a screening session. Ground reaction forces (GRF) and right sagittal kinematic data were sampled simultaneously. Results were analyzed in a 3x3x2 repeated measures ANOVA using SPSS (SPSS Inc., Chicago, Illinois, USA) statistical software. Unilateral knee extension was significantly greater in Excessive Knee Extension Female (EKE-Fem) than either Male (ML) or Normal Knee Extension Female (NKE-Fem). As height and landing stiffness increased, results generally showed increases in the first (F1) and second (F2) peak GRF and the loading rate of F1 (LRF1) and F2 (LRF2) across all groups. Overall, NKE-Fem produced higher mean peak values for the selected vertical GRF variables regardless of height or landing technique. EKE-Fem generated lower F2 values than ML in the stiff landing at 75 cm and than NKE-Fem in F2, LRF1, and LRF2 in all landing conditions. A greater degree of plantarflexion and knee contact velocities were also observed in EKE-Fem during landing. These results suggested that an adaptive landing strategy was adopted by EKE-Fem to minimize impact forces. NKE-Fem landed with significantly greater knee extension angle and velocities at contact than ML at the two higher heights and greater F2, LRF1, and LRF2 than EKE-Fem in all conditions, indicating a poor capacity of the subject group in impact attenuation

which could potentially overload the knee joint. In addition, the results suggest that while NKE-Fem might find difficulty in dissipating impact forces, EKE-Fem demonstrated a tendency to avoid full knee extension through compensatory measures at the ankle (increased ContAng and ROM). These findings demonstrate differences between males and females in kinetic and kinematic variables during landing, which may have implications in injury mechanisms of the anterior cruciate ligament.

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Chapter I

Introduction

Women's participation in athletics has dramatically increased over the past fifteen to twenty years. The passage of Title IX in 1972 created a greater opportunity for women to become involved in sports.[11,14] Since that time, the number of females participating in intercollegiate athletics has risen.[2] The National Collegiate Athletic Association (NCAA) showed an increase of 9% in female participation of all NCAA athletic programs over a three year period (1989 to 1992). With this increase comes greater potential for athletic injuries.

With the rise in female athletic participation, a proportional increase of injuries in female athletes has been noted.[14] When compared to male athletes competing in the same sports, females who participate in sports that involve jumping and cutting maneuvers have a four to six times greater incidence of knee injuries.[12] This increased incidence may be related to a larger population of female participants, as well as an elevated risk for certain types of injuries to the knee. In particular, the anterior cruciate ligament (ACL) has been reported to have a more significant chance of being injured (2 to 8 times) in women.[9,11] While research has focused on extrinsic factors such as conditioning, muscle recruitment patterns, and landing techniques, underlying intrinsic causes like subtalar joint pronation, genu recurvatum (knee hyperextension), and pelvic alignment should not be ignored. Loudon et al.[16] has demonstrated a relationship between static postural measurements and ACL injuries in the female athlete. In addition, greater subtalar joint pronation and knee joint laxity have been suggested by Woodford-Rogers et al.[27] as risk factors for ACL tears.

The mechanism of injury (MOI) for the ACL has been a main focus for many studies. Activities of high risk include decelerating and landing in an "awkward" position.[9] A non-contact injury mechanism accounts for approximately 80% of injuries to the ACL,[2] most of which occur while landing from a jump.[12] In a study involving Italian volleyball players, Ferretti et al.[8] found landing to be the most frequent MOI in 52 knee ligament injuries. Two types of landings that have been identified as potential mechanisms of injury are straight knee landing and one-step stop landing with the knee hyperextended.[2] During landing, a tremendous amount of impact force is applied to the body, which may lead to injuries to lower extremity joints.[22] Muscles, tendons, menisci, and other anatomic shock absorbers help to minimize the magnitude of impact forces. These forces have shown a strong correlation with anterior acceleration of the tibia. Anterior shear forces at the knee joint may be correlated with GRF (ranging from 2.2 to 6.9 BW) in landing activities[17] and the ACL is responsible for approximately 86% of restraint to these forces.[4] Therefore, by minimizing these forces, it is believed that ACL ruptures might decrease.

Two types of foot contact patterns during landing have been identified through research.[6] A toe-heel landing is widely used and produces a bi-modal ground reaction force-time history curve. A flatfoot contact pattern normally produces a unimodal ground reaction force curve. A toe-heel landing is characterized by a longer time period of deceleration and lower ground reaction forces (GRF),[19] while a flatfoot landing is typically associated with a stiff landing and a more upright posture.[5,22] The latter type of landing also induces greater ground reaction forces. Results from Zhang et al.[29] demonstrated increased loading to the body with increasing stiffness during landing. It has also been reported that mean GRF are as high as 4.66 times BW when landing flatfooted, while those who land toe to heel have mean GRF of 2.22 times BW.[22]

Since increased GRF have been correlated with stiff landing from a jump,[6] it is important to consider the absorption at the knee and the possible stresses placed upon the ACL. In their report on non-contact ACL injuries, Griffin et al.[9] suggest that the ACL acts as a major restraint of anterior tibial displacement on the femur when the knee is straight and in a weightbearing position. If this is the case, the ACL might come under significant tensioning when an individual lands in a stiff-knee position. A stiff landing in combination with the previously mentioned postural faults could play a role in contributing to possible ACL injuries.

Problem Statement

Although research has examined biomechanical factors associated with injuries during landing[6] and static postural measurements related to ACL injuries,[16] little research has been done to combine the two. McNair and Prapavessis[18] measured the normative data of vertical GRF of 234 adolescents (mean age: 16). Although results showed no significant differences across gender, there was no account for differences in step-off height, stiff and soft landings, or static postural measurements. Because it is believed that a flatfooted (stiff-leg) landing will normally produce higher GRF[22] and increased knee hyperextension has been shown to be associated with ACL injuries,[16, 27] a combination of these conditions may predispose the ACL to significant stresses. Therefore, the purpose of this study is to examine the relationship between selected static postural measurements and the biomechanical characteristics in landing activities. While there is an appreciation for extrinsic risk factors for the ACL (neuromuscular control and landing techniques), the focus of this study will be on intrinsic factors of subtalar joint pronation, knee hyperextension, and pelvic alignment. These factors in association with injury potentials across gender in landing will be assessed.

Hypotheses

1. An increase in landing height and stiffness would cause significant changes in VGRF and kinematic variables across groups.

2. Significant gender differences in VGRF and kinematic variables would be seen.

3. The female group with excessive knee extension (EKE-Fem) would produce significantly different VGRF and kinematic variable values than both the female group with normal knee extension (NKE-Fem) and the male group (ML).

Delimitations

The study was conducted within the following delimitations:

1. Sixteen active and healthy subjects (5 males and 10 females) between the ages

of 18 and 25 years of age were selected from the student population at The University of Tennessee to participate in the study. They had no impairments of the lower extremities.

2. Six test conditions included drop landings from three different heights (45, 60, and 75 cm) with two different landing techniques (stiff and soft landing).

3. Biomechanical signals were collected and analyzed for duration from 100ms prior to the contact to the end of the landing phase for each trial.

4. Data were collected at 1000 Hz from a force platform (AMTI), one electrogoniometer (Penny+Giles Biometrics Ltd.), and at 120 Hz from a digital camera (JVC, GR-DVL9800) for each trial of the landing activity.

5. Collection of data for each subject was completed in one session.

Limitations

The study was limited by the following factors:

1) Subjects were limited to the student population at The University of Tennessee.

2) Possible errors from placement and digitizing of the reflective markers. Other errors such as perspective error and marker placement are acknowledged.

3) Inherent errors from the force platform, accelerometer, and/or digital video systems. Errors of force platform and high-speed video systems are always present but were considered acceptable within the specifications of the manufacturers. Confining the activity to the sagittal plane controlled errors caused by out of plane motion.

4) Potential errors due to the difference in sampling frequency of the force platform (1200Hz), the digital video system (120Hz), and synchronization of the systems. Synchronization accuracy between the force and video systems was limited by the sampling rate of the slower system. The video system has a sampling error of ± 0.08 frames/second, resulting in a maximum error of only 0.67 ms.

Assumptions

The following assumptions were made for this study:

1. The biomechanical measurements used were sufficient for analyzing the effects of drop landings with different landing heights and techniques.
2. Biomechanical instruments used were accurate.
3. All of the subjects were free of lower extremity injuries at the time of testing.
4. The performance of the subjects was symmetrical, therefore, only the right side was assessed for kinematics and GRF.

Chapter II

Literature Review

In 1972, the US enacted the Title IX Educational Assistance Act which provided an expansion of opportunities for women to compete in sports.[11] Since its passage, there has been an increase in female participants in all sports and at all skill levels. [15] Before this time there were fewer than 10,000 female college athletes. Twenty years after Title IX, the number had risen to nearly 100,000 women. The overall number of NCAA sponsored varsity soccer programs for females climbed from 308 to 455 between 1990 and 1995.[2] With this increase came a concern that women were likely to suffer more injuries. Although the overall injury rates for males and females have been similar, a two to eight times[11] greater predisposition of anterior cruciate ligament (ACL) rupture has been shown in women.[12,16]

Female soccer players, gymnasts, handball, and basketball players appear to be at the greater risks for tearing the ACL.[2,15,16] With these tears comes the high cost of surgery, rehabilitation, and time lost from the sport. Roughly 50,000 ACL reconstructions are done every year with an approximate cost of $17,000 per surgery.[11] This cost does not even include the initial care or the possible long term effects of altered biomechanics of the knee joint. Many patients who have sustained ACL injuries and undergone a surgical procedure may suffer from post-traumatic degeneration or early onset of arthritis. The emotional and physical burdens applied to the injured athlete justifies efforts toward decreasing these injuries.

A review of the National Collegiate Athletics Association (NCAA) data for knee injury patterns in college basketball and soccer athletes revealed a significantly higher ACL injury rate for women compared to their male counterparts.[2] Data was collected by the NCAA Injury Surveillance System which examines men's and women's sports during the fall, winter, and spring. Random selections of injuries are made with a minimum 10% representation in NCAA divisions I, II, and III as well as by region (East, West, South, North). This procedure provides a representative national cross-section of injury rates of the NCAA. Injury figures were collected during a five-year period (1989 to 1993) on a total of 471 men's teams and 278 women's teams. The results show a statistically significant difference ($P < 0.05$) in ACL injury rate in female athletes in soccer (0.31/1000) and basketball (0.29/1000) compared to their male counterparts (0.13/1000 and 0.07/1000).

Ferretti et al.[8] reported 52 cases of serious knee injuries in volleyball players in the Italian Volleyball Federation. All of the injuries occurred over a ten year period and included tears to the ACL that required surgical intervention. The ACL athletes consisted of 10 men and 42 women with an average age of 22.2 years. There were 30 acute injuries and 22 cases of chronic instability. The most common mechanism of injury was landing from a jump, with 38 of the injuries involving a twisting motion during the landing. Other important findings included a greater percentage of injuries during play rather than practice, as well as the failure to support fatigue as a factor of injury. Female athletes were more affected than males which were similar to the athletes in other high-risk sports such as basketball, soccer, and gymnastics.[2]

Risk Factors

Harmon and Ireland [11] describe two different types of risk factors (intrinsic and extrinsic) upon which research has been focused regarding gender differences in anterior cruciate ligament injuries. Intrinsic factors include joint laxity, hormonal influences, notch size, limb and pelvic alignment, and ligament size. These factors are inherent and more difficult to control. Extrinsic factors are those like conditioning, skill and experience, muscle recruitment patterns, and neuromuscular control during landing. Although these factors are difficult to quantify functionally, they may be easier to modify. For the purpose of this study, both intrinsic and extrinsic factors will be reviewed, but with an emphasis on correlation between static postural measurements and potential for ACL injuries across gender.

<u>Extrinsic</u>

Researchers have examined the association between conditioning and ACL injuries in female athletes.[12,15] Although good conditioning has been shown to increase performance and decrease the risk for injury, no solid research evidence suggests that female athletes are less conditioned than the men. The increased conditioning level of females has, in part, contributed to an increase in experience and participation. From 1971 to 1998 female participation in high school sports grew from 3.7% to 33.3%. It was suggested that these numbers cause not only an increase in overall awareness of females in sports, but also a possible decreased risk of ACL injuries secondary to inexperience.

Another extrinsic factor of interest is muscle strength and recruitment. Dynamic stabilization of the knee occurs when the quadriceps, hamstrings, and gastrocnemius work together to protect the ACL.[12] It is believed that the ACL might even fail during activities of daily living if the dynamic stabilizers were not active. These stabilizers come into play during cutting and landing maneuvers that take place in sport activities. In particular, the hamstring muscles are activated for protection of the ACL. The problem is that female athletes have less dependence on the hamstrings and more on the quadriceps and gastrocnemius than do the male athletes. Hamstring-to-quadriceps strength ratios are thought to be less in women when compared to men.[2,12] This becomes important in landing when an athlete is trying to decelerate from a jump.

It has been reported that nearly 80% of ACL injuries occur in a non-contact situation,[2] with the majority of these occurring while landing from a jump.[13] These injuries can occur during deceleration of the lower limb.[16] In a review of literature involving NCAA data of female basketball players, straight-knee landing and one-step stop landing with the knee hyperextended accounted for 28% and 26% of non-contact injuries respectively.[2] These females were taught to land with the knees slightly flexed, enabling the hamstrings to be in a more favorable position to stabilize the knee joint by controlling rotation and anterior displacement of the tibia. Therefore, the hamstring muscles must have sufficient strength and neuromuscular control to maintain the slightly flexed knee position during landing.

When landing from a jump females have been shown to have greater adduction and abduction moments at the knee, take a longer period of time to develop peak hamstring torque,[10] and have difficulty decelerating the body.[11,12,16] In order to land and decelerate the body safely, an athlete must have functional joint stability. This type of stability is maintained by both static and dynamic stabilizers.[11] Contributions of the dynamic stabilizers include precise neuromuscular control on skeletal muscles. Neuromuscular control arises from the concept of unconscious activation of dynamic restraints (muscles) when responding to a stimulus. To respond, one must have an accurate awareness of where the joint is in space, or proprioception. This is the best source for providing sensory information in order to mediate neuromuscular control and enhance functional joint stability. When proprioceptive signals are supplied to the muscles in preparation for activation, appropriate adaptations can be made that might shield the ACL from extreme forces via prophylactic mechanisms.

Sources for proprioception involve the mechanoreceptors located in muscles, joints, and other soft tissues.[11] These sources are found throughout the lower kinetic chain (ankle to knee to hip) and stimulated as the "chain" moves through the three different planes of motion (sagittal, frontal, and transverse). In the sagittal plane, flexion and extension occur. If hyperextension is allowed, the ACL can be stressed. Abduction and adduction forces arise in the frontal plane. Excessive moments in this plane can damage the medial collateral ligament

(MCL), which is a secondary stabilizer to the ACL. Finally, in the transverse plane, rotational forces and pronation/supination can occur. Shock absorption and deceleration occur during normal lower extremity kinetic chain pronation.[16] When excessive pronation is allowed to take place, the tibia will internally rotate and tension the ACL. At the more proximal hip joint, the femur follows the lead of the tibia as it internally rotates. The close connection of the knee and hip has been described as the moment produced at the knee being "slaved" to the moment produced at the hip.[9] Thus, one influences the other.

According to Griffin et al.,[9] decreased activation of the hip extensors in females might explain the upright hip position and greater knee extension angle seen during landing from a jump. This position could serve as a protective mechanism from having to decelerate the trunk over the hips. With decreased hip-muscle activation (gluteus maximus and medius), there will be less possibility of maximal quadriceps and hamstrings activation. Hewett et al.[12] investigated the effects of neuromuscular training on the incidence of female knee injuries in high school sports teams (soccer, volleyball, and basketball). An instructional video and training manual demonstrating correct technique of a 6-week preseason neuromuscular training program was sent out to the coaches and trainers of these schools. The program consisted of flexibility, plyometrics, and weight training activities. Of the 43 sports teams, 15 female teams elected to use the program and 13 untrained male teams were used as controls. A one year study period involved monitoring injuries in these sports during the respective season. Ninety-four percent of the athletes (n=1263) were monitored throughout the season and were included in the data analysis. Results of the study show a significant effect of training on the incidence of serious knee injuries.[12] While the untrained female group demonstrated a significantly higher injury incidence than the male group, the trained female group showed no significant difference in knee injuries when compared to the males. The injury rate for the untrained female group was 3.6 times higher than the trained females and 4.8 times higher than the males. The trained female group had an injury rate incidence only 1.3 times that of the male control group. These results demonstrate a decrease in female knee injuries following a specific neuromuscular plyometric training program.[12]

Intrinsic

Increased ligamentous laxity in females has been closely examined in regards to the rate of ACL tears.[1,2,11,16,27] The KT 1000 ligament arthrometer (Med-Metric, San Diego, CA) has been proven reliable in testing anterior-posterior (A-P) translation of the ACL.[11] Although some studies have shown a correlation between increased laxity and ACL tears in females,[16,27] other reports have found no significant differences in knee laxity across gender.[2,14] Because A-P tibial translation increases with exercise, these findings present some concern over whether or not static measurements correctly represent ligamentous laxity at the time of injury.[11]

Hormones may also play a role in increased laxity of ligaments in females.[11,14,15] Levels of estrogen and progesterone and their ratio change during the menstrual cycle. Receptors of estrogen and progesterone have been found on the ACL in men and women.[11] Because of female sex hormones and the fact that physiologic levels of estrogen affect fibroblastic production of collagen, it is believed that anterior cruciate ligament injuries may occur more often in females. Another female sex hormone thought to be related to ligamentous laxity is relaxin. This hormone is found in women during pregnancy and the luteal phase of the menstrual cycle. It is associated with ligamentous relaxation during the birth process and increases the risk for injuries during pregnancy.

Although research has shown a relationship between ACL injuries in females and the menstrual cycle, limitations of the studies have made it difficult to make accurate conclusions.[11] A lack of hormone blood level confirmations and irregularities in women during their menstrual cycles have clouded the relationship. Because of the complicated interplay between hormones and the association to ligamentous laxity, no clear evidence of one point in the menstrual cycle being riskier than another has been demonstrated.

Recently, the intercondylar notch width of the femur has become a popular topic of discussion.[11,14,15,26] The notch width index (NWI) is the ratio between the intercondylar notch width to the width of the distal femur at the popliteal groove.[15] Those with a low NWI are thought to be predisposed to ACL ruptures. The range of NWI in males and females is large and makes it difficult to make accurate conclusions about the contributions of a small notch to gender differences in ACL injury rate. Anderson et al.[1] found no significant difference in NWI between male and female highschool basketball players. Among the one hundred highschool basketball players (50 male and 50 female) who participated in the study, results did reveal a smaller size of the anterior cruciate ligament in females when adjustments for body weight were made. When correlation of the ACL area with strength measurements were examined, data implied that those with stronger quadriceps muscles are likely to have larger anterior cruciate ligaments.

In a similar study, Shelbourne et al.[26] studied the relationship between intercondylar notch width of the femur and the incidence of ACL tears. The median notch width was 13.9 $\pm$ 2.2 mm for women and 15.9 $\pm$ 2.5 mm for men (n=714). For notch width measurements between height groups, no significant differences were found for either males or females. However, results suggest that the ACL tear rate is affected by notch width. Even when women are the same height as men, on average, they have smaller notches. Finally, because the femoral condylar width increases with increasing patient height, while the notch width does not, the NWI should not be used as a reflection of notch width. Instead, the authors suggest that the absolute notch width is a more accurate indicator.

From a biomechanical standpoint, variations in lower extremity alignment have been implicated as a possible source for ACL injuries in females.[2,11,14,15,16] A wider pelvis, increased femoral anteversion, increased genu recurvatum and valgum, increased anterior pelvic tilt, and more foot pronation are a few of the differences between males and females. Beginning with the pelvis, there is a direct link between the foot, ankle, knee, and hip. As the pelvis tilts anteriorly, the hip and knee internally rotate while the foot pronates at the subtalar joint. Lower extremity internal rotation occurs in the transverse plane and can proceed too fast and too far if hip weakness is present. Hip external rotator musculature is responsible for decelerating the femur and subsequently the tibia as the foot hits the ground during gait. If the femur and tibia are not properly decelerated, excessive pronation will occur. These motions of the lower extremity take place in order to absorb the forces transmitted from the ground.

In the frontal plane females tend to display greater coxa varum and genu valgum alignment.[15] This is concurrent with the rotational forces at the tibiofemoral joint and can contribute to excessive pronation. The valgus force at the knee loads the medial collateral ligament (MCL) and is common to the ACL injury mechanism. Genu valgum alignment can also be associated with an incidence of increased quadriceps angle (Q angle). It is an acute angle formed by drawing lines from the anterior superior iliac spine (ASIS) to the middle of the patella, and from the tibial tuberosity to the center of the patella.[28] Although it can be measured in supine or standing, the Q angle is believed to be more functional when standing. Because of the wider pelvis of the female, increased Q angles might be associated with predisposing the knee to injury.

Woodland et al.[28] determined the normal mean Q angle of college-aged men and women. Two hundred sixty-nine males and 257 females were randomly selected from college physical education classes. The testing procedure consisted of measuring the right knee Q angle of each subject in standing and supine. Results showed a statistically significant difference in Q angle between men and women, as well as significant differences between standing and supine positions. Also, women showed a larger increase in Q angle from supine to standing than did the men. The men had average measurements of 12.7 degrees in supine and 13.6 degrees in standing while the women had mean Q angles of 15.8 degrees supine and 17.0 degrees standing. This study provided quantitative values of Q angles for men and women as baseline information for examination of possible biomechanical gender differences associated with knee injuries.

Genu recurvatum and increased anterior pelvic tilt are seen in the sagittal plane and can create an impingement of the ACL in the intercondylar notch while also increasing tensile strain on the ligament.[15] An anterior pelvic tilt position will cause the gluteus medius and hip external rotators to work harder in order to maintain a neutral lower extremity alignment while opposing a more internally rotated posture. A combination of these intrinsic factors can pre-load the anterior cruciate ligament before a female athlete even participates in her sport. If the neuromuscular system does not accommodate for these postural faults during dynamic activities, the passive ligamentous restraints are called upon to provide stability. When this happens, there is a greater likelihood for ACL injury.

The relationship between static posture and ACL injuries in female athletes has been studied.[16] Twenty females (average age 26.5 years old) with a unilateral ACL injury participated in the study. Injury occurrence was within 2 years of the test date and each knee injury was either reconstructed or the patient chose conservative nonoperative treatment. The control group consisted of 20 age-matched athletic females with non-pathological knees. Seven postural measurements were evaluated for each subject, including the involved-side for the ACL-injured group. The variables consisted of femoral anteversion, standing sagittal knee extension, pelvic position, hamstring length, standing knee angle in the frontal plane, navicular drop, and subtalar joint neutral position. Each variable was placed into one of two categories (normal and abnormal) for statistical purposes. Univariate and multivariate statistics were used for assessing the variables. To determine the significance of individual variables as discriminators between ACL-injured and normal subjects, the McNemar test of symmetry was used. For multivariate measurements (logistic regression model), the variables of excessive sagittal knee extension, excessive navicular drop, and excessive subtalar joint pronation were found to be associated with the ACL group. Pelvic position was found to have a strong link to injury of the ACL when analyzed as a univariate measure. It is believed that with increased anterior pelvic tilt, the hip is placed in a flexion moment. This is counteracted by an extension moment at the knee, placing tension on the ACL. Because tibial rotation follows subtalar joint motion, excessive pronation will lead to tautness of the ligament as well. A combination of these factors will cause greater strain to the ACL than any single postural abnormality. For clinical purposes, the results of this study demonstrate a strong association between a standing posture of genu recurvatum with subtalar joint hyperpronation and non-contact anterior cruciate ligament injuries.[16] Because static posture is usually the starting position from which a dynamic activity begins, one with genu recurvatum and overpronation might unconsciously predispose the ACL to injury.

In a similar study of risk factors for anterior cruciate ligament injury, Woodford-Rogers et al.[27] examined the subtalar joint and internal tibial rotation.

The uninjured lower extremity of fourteen ACL-injured male highschool and college football players was assessed, along with a total of eight female ACL-injured highschool and college gymnasts and basketball players. The control group consisted of an equal number of subjects matched for age, sport, and playing time with no history of an ACL injury. Measurements of calcaneal alignment changes during stance, navicular drop, and anterior translation of the tibia (with a KT-1000) were taken. For the uninjured football players, no significant differences were found between right and left lower extremities in navicular drop, calcaneal changes, or knee hyperextension. Navicular drop, anterior drawer with 20 lb of force, and maximum manual drawer were most indicative of predicting group classification (ACL-injured and uninjured) in a discriminative analysis with 71.4% being classified correctly ($P < 0.01$). In the females, navicular drop, anterior drawer with 20 lb force, and maximum manual drawer were also the best classifying predictors. Correct classification occurred in 87.5% of the female subjects. The findings suggest that increased navicular drop and knee joint laxity are associated with increased risk of anterior cruciate ligament injury.[27] When knee joint laxity is added to the equation, the ligament is predisposed to abnormal stresses and an injury is likely.

The behavior of ACL strain in knee extension has been determined by researchers.[3,10] During knee flexion angles of 90 degrees, ACL strain values are at a minimum, while the values significantly increase as the knee approaches full extension. Beynnon et al.[3] reported mean ACL strain values of 3.8% $\pm$ 0.5% with 45 N of weight and 2.8% $\pm$ 0.6% without weight during active flexion/extension of the knee. Even when there was simultaneous contraction of the quadriceps and hamstrings, significant strain (mean 2.8% ± 0.9%) was applied to the ACL at 15 degrees of knee extension in comparison to 30 degrees of extension (mean 0.4% ± 0.5%). When the ACL is removed, average anterior tibial displacement increases 1.5 mm (0.5 to 2.0 mm) as the knee extends from 30 to 0 degrees.[10] Along with this finding was the fact that an increased quadriceps force was shown to occur during the last 15 degrees of knee extension. This data provides important clinical significance when considering strain of the ACL in relation to knee extension not only during open-chain activities, but also in closed-chain movements such as landing.

Ground Reaction Forces

Landing is a task involved with many recreational and sporting activities. It is associated with injuries to the lower extremity as a result of the impact forces absorbed by the body.[22] These forces are often transferred up the kinetic chain to the knee joint where they are reported to be highly correlated with anterior tibial accelerations and possible stress to the ACL, which is responsible for restraining approximately 86% of the forces.[4] Straight knee landing (28%) and one-step stop landing with the knee hyperextended (29%) are two mechanisms of injury to the ACL found among female basketball players.[2] In 52 cases of ACL injuries in Italian volleyball players, the most common mechanism of injury involved landing from a jump.[8]

Research has identified 2 types of foot contact during landing from a jump: toe-heel and flatfoot.[6] Flatfoot landing typically produces a unimodal GRF pattern while toe-heel landing exhibits bimodal GRF. Those individuals who land flatfooted normally have larger GRF (4.66 times body weight) secondary to increased heel velocity at footstrike, while toe-heel strikers have smaller GRF (2.22 times body weight).[22] McNair and Prapavessis[18] examined normative values of vertical ground reaction forces during landing from a jump. A total of 234 subjects (154 males and 80 females) between the ages of 13 and 19 years of age participated. Each subject jumped from a box (0.3 meters in height) to land on a force plate and was instructed to land so as to minimize the stress. The results showed a mean GRF of 4.5 body weight (BW) with ranges from 2.0 BW to 10.4 BW. No significant differences ($p > 0.05$) were noted across gender or activity levels. However, landing techniques and drop heights were not varied in order to examine the different effects.

In a study of energy dissipations during landing by Zhang et al.,[29] three different landing techniques and landing heights were used. Nine males (age 25 $\pm$ 5 yr.) dropped from three different landing heights (0.32 m, 0.62 m, and 1.03 m) using either a soft (SFL), normal (NML), or stiff (STL) technique. Variables of range of motion (ROM) for the ankle, knee, and hip, peak GRF, peak joint moments and powers, and total eccentric work for the different lower extremity muscle groups were examined. Results indicated increases in peak GRF with increasing landing height and stiffness conditions. In general, as landing heights increase, biomechanical mechanisms respond accordingly. A shift from distal to proximal muscles groups was seen as mechanical demands increased. Most of the energy absorption was performed by the larger knee and hip extensors. Although the plantar flexors (STL) and hip extensors (SFL) varied as to when major contributions were made, the knee joint extensors remained consistent across conditions when contributing to energy dissipation. These findings suggest that landing height and technique are significant interactions when investigating lower extremity energy dissipation during landing activities.[29]

Dufek and Bates[7] looked at predicting impact forces in order to better understand the body's accommodative mechanisms during landing. A total of three males (27 – 30 years old) participated in the study by performing a series of landings from three jump distances (40, 70, and 100 cm) and three jump heights (40, 60, and 100cm) while using three different landing techniques: stiff knee (ST), slightly flexed knee (SL), and fully flexed knee (FF). GRF data were examined and three regression models (mechanical, biomechanical, and refined biomechanical) were used to predict Fl and F2. Three main effects of height (H), distance (D), and technique (T) were identified from ANOVA for F1. In the mechanical model, H was found to be the dominant variable when predicting F1 among all the subjects, while all three biomechanical models targeted T as the most important variable. Across all the main effects F1 and F2 increased with both H and ST landing technique. Overall, the data suggest increasing vertical forces are accompanied by greater landing heights and increased knee extension techniques which might functionally predispose one to injury.[7]

Because injury to the ACL is associated with jumping activities, landing characteristics of normal and ACL-deficient knees has been investigated.[17] A total of sixteen subjects (10 men, 6 women) who had confirmed ACL rupture participated in the study. The control group consisted of 16 subjects with no history of lower extremity musculoskeletal conditions. Average age of all the subjects was 27 years old. Functional ability was assessed with a questionnaire and a series of hop tests prior to testing. Each subject was instructed to hop from a 300 mm (height) box and land on the force platform in a balanced position on one leg. For trials and group, a two factor MANOVA revealed no significant differences ($p<0.05$) between the ACL-deficient and normal knee groups. However, when lateral hamstring muscle activity was correlated with peak vertical GRF, there was a significant difference noted. As hamstring muscle activity increased, GRF decreased secondary to lower anterior tibial accelerations across all subjects ($r = 0.87$).[17] Finally, based upon the results of the study, the authors concluded that teaching skills of landing "softly" should be incorporated into ACL rehabilitation programs.

The effects of instruction on GRF during landing activities have been studied.[19,20,22] Prapavessis and McNair[22] looked at the roles that augmented and sensory feedback played when individuals land from a jump. A total of 91 (35 females, 56 males) subjects with an age range of 13 to 19 years old participated in the study. The peak vertical GRF was established as subjects landed on the force platform from a height of 300 mm. They were then randomly assigned to either an augmented or sensory feedback group. Data were analyzed using a 2-factor (feedback and trials) repeated measures ANOVA with a significance level of 0.05. Results showed no significant main effect for the "feedback" factor, but did show a significant main effect for the "trial" factor. GRF after feedback (trial 2) were significantly less than without feedback (trial 1). Also, a significant difference was found between GRF of sensory and augmented groups. Subjects with augmented feedback had lower GRF (mean = 3.57 ± 1.10) when compared to the sensory feedback group (mean = 4.33 ± 1.54).[21] These findings demonstrate that individuals who have received augmented feedback are able to land with less force than those who rely on previous experience.

In a similar study, Onate et al.[20] looked at the effect of augmented feedback in reducing jump landing forces. Sixty-three subjects (18 to 25 years old) were randomly placed into one of four groups (augmented, sensory, control I, control II). A Vertec jumping instrument (to assess maximal vertical jump height) and two series 4060 Bertec force plates were used to collect the data. Each subject was instructed to jump as high they could while hitting the Vertec vanes and to land onto the force platform as soft as possible. Testing occurred over a one-week period which included initial testing with videographic data, followed by an immediate post-test (2 minutes) and a one-week post-test. The augmented group received visual and verbal feedback while the sensory group relied solely upon feedback from their jumping experience. Neither control group received feedback. Results from using a 1-way ANCOVA showed a significant difference between baseline scores and both post-test scores of peak vertical GRF (PVGRF) by group. When compared to the Sensory and Control I groups, the Augmented group significantly reduced their PVGRF from baseline to immediate post-test as well as for baseline and delayed post-test scores when compared to both control groups. Scores for within-group comparison revealed a greater reduction of PVGRF in Augmented when compared to Sensory and Control I groups.[20] Overall, findings display the fact that augmented feedback significantly decreases jump landing impact forces in performances for both immediate (2 minute) and delayed (1 week) testing by approximately 0.80 BW.

Finally, McNair et al.[19] looked at the effect of instruction on decreasing forces from landing. In this study 80 subjects (27 men, 53 women) with a mean age of 24 years were instructed to perform two sets of eight landings from a box 300 mm in height. Following the first eight landings, subjects were randomly assigned to one of four different groups: (1) technical instruction, (2) auditory cue, (3) imagery rehearsal, and (4) control. After being assigned to their respective groups, each subject repeated the eight landings based upon the instructions received. An ANCOVA was used to assess the dependent variable (mean VGRF) to the different groups and revealed a significant difference between the technical instruction and auditory cue groups when assessed against the control group. These findings confirm the use of instruction with an external cue (sound of foot hitting ground when landing) in decreasing VGRF during landing, and possibly helping to diminish stresses to the ACL.

Summary

This chapter has illustrated the incidence of ACL injuries and research involving GRF during landing. Evidence from the literature demonstrates the correlation between both extrinsic and intrinsic MOI to the ACL. Research has implicated certain static postural measurements as factors[27] and predictors[16] for ACL-injured group classification. The impact forces absorbed by the body during landing have been shown to differ with landing height and technique,[29] which might predispose the ACL to undue stresses. Research on the relationship between static postural measurements and possible correlation to biomechanical characteristics during landing is limited. Therefore, future investigations of biomechanical characteristics during landing and the possible relationship to MOI to the ACL are warranted.

Chapter III

Methods

Experimental Methods

The purpose of the study was to investigate the relationship between static postural measurements and biomechanical characteristics during landing. The protocol consisted of a warm-up, anthropometric measurements, static postural measurements, and six test conditions. Each subject performed five trials of drop landings in each of six different conditions, for a total of 30 trials.

Subjects

Subjects were recruited from the student population at The University of Tennessee. Thirteen healthy and active male (5) and female (8) subjects volunteered to participate in the study. The subjects were divided into three groups: male (ML), female with normal knee extension (NKE-Fem), and female with excessive knee extension (EKE-Fem). The number of subjects in each group included five in ML, four in NKE-Fem, and four in EKE-Fem respectively. Healthy and physically active subjects were defined as one who had no lower extremity impairments to his or her lower extremities and who exercised regularly 2 to 3 times per week. Prior to their participation, all subjects were briefed on the purpose, procedures, risks, and benefits of the study. All subjects signed an informed consent form approved by the Institutional Review Board at The

University of Tennessee prior to their participation in the study. Subject information is provided in Appendix B.

Instrumentation

All testing was conducted in the Biomechanics/Sports Medicine Lab, Room 135, HPER Building at The University of Tennessee. The biomechanical instruments used for the study included a force platform, a digital camera, an electrogoniometer, lab shoes, a trigger device, a reference frame, reflective markers, an analog/digital (A/D) converter, and an Ariel Performance Analysis System for data collection and processing.

Kinematics

During the test, a digital camera (JVC, GR-DVL9800) was used to record kinematic data of the right sagittal view of each subject. A reference frame (width = 140.97 cm, length = 186.69 cm) was used to obtain scale factors in order to convert coordinates of reflective markers. The reference frame contains four reflective markers on each corner of the structure.

Reflective markers were placed on the right side of the body at the shoulder, hip, knee, ankle, heel, and the head of the fifth metatarsal (Figure 1). Recorded video images were digitized to obtain coordinates of these markers during the activity using Ariel Performance Analysis System (APAS, Ariel Dynamics, Inc.). Following digitization, the coordinates were imported into a customized program to determine the time-history and discrete events of linear and angular positions, velocities, and accelerations.

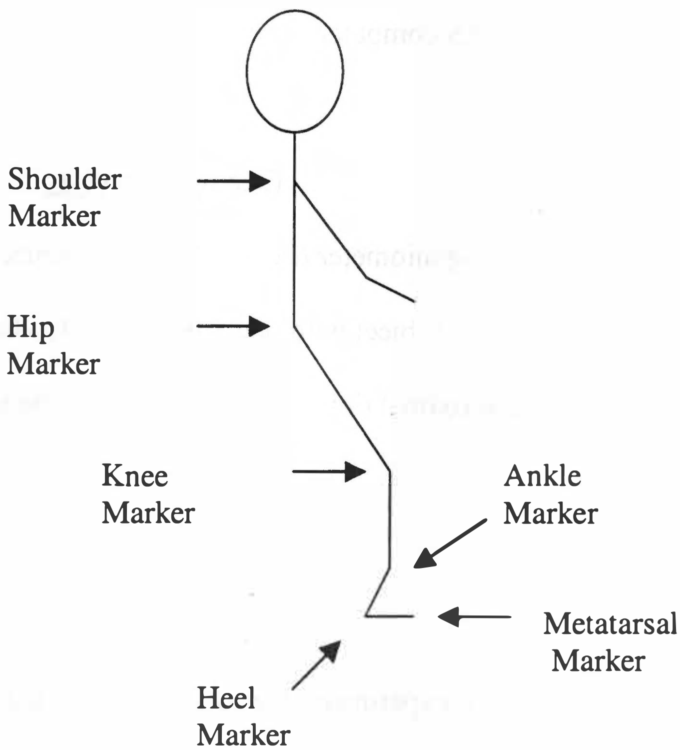

Figure 1. Placement of reflective markers

Force Platform

In order to measure ground reaction forces (GRF) during the test, a force platform (OR6-7, AMTI) that was flush with the floor was used. The GRF data included Fz (vertical), Fx (medial-lateral), and Fy (anterior-posterior) forces. Signals from the force platform were sampled for 1.5 sec at a frequency of 1200 Hz. These signals were also amplified through the A/D converter prior to being stored in the APAS computer.

Electrogoniometer

An electrogoniometer (Penny+Giles Biometrics Ltd) was placed around the left knee of the subject with one arm strapped to the distal femur and the other strapped to the proximal tibia in order to monitor the knee joint angle during data collection.

Synchronization

During the experiment, the force platform, the electrogoniometer, and the sagittal view digital video were recorded simultaneously. A customized trigger with a light emitting code (LED) was used to synchronize kinematic and analog signals.

Shoes

Identical lab shoes (Adidas) were provided by the Biomechanics/Sports Medicine Lab and worn by subjects during testing.

Experimental Protocol

The principal investigator outlined the purpose and procedures of the study for each subject prior to their participation. Subjects were further informed about the purpose, the number of conditions, the number of repetitions, and the performance requirements of the study on the day of testing. The testing session was completed in approximately one and one-half hours.

Subjects began their test session by riding a stationary bike for five minutes as a warm-up. Anthropometric measurements were then obtained for the proximal and distal circumferences and the length of lower extremity segments. Each measurement was taken three times and the average was used for further analysis.

In the screening session, static postural measurements (unilateral knee extension, navicular drop, and standing pelvic angle) were also evaluated three times and the average value was recorded (Table 1). In unilateral knee extension, the subject stood in an upright posture with the right lower extremity in hip and knee extension.[16] After shifting the body weight to the right lower extremity, the subject was asked to bring the knee into a position of maximal extension. A standard goniometer was used to record the measurement. Subjects were classified into either a normal knee extension (0 to 5 degrees) or excessive knee extension (> 5 degrees) group depending upon the measurement. The navicular drop was measured in both sitting and standing. While sitting, subtalar neutral

Table 1. Static postural measurement and definitions

Static Postural Measurement
UKE = Unilateral Knee Extension PA = Pelvic Angle ND = Navicular Drop
Measurement Definition
UKE = Angle of the tibiofemoral joint between the greater trochanter and the lateral malleolus PA = Angle between the PSIS and the ASIS ND = Difference in height of the navicular bone while sitting with subtalar neutral and the height of the navicular bone in standing while relaxed

was palpated. This is performed by palpating the head of the talus until the medial and lateral sides are felt equally.[25] Once this was found, the height of the navicular was measured from the floor to the distal most portion of the navicular bone. The measurement was taken again while the subject was standing with a relaxed foot. The difference between the two measurements was calculated. Classification of subjects was either normal (0 to 6 mm) or high ($\geq$ to 9mm). A set of calipers and a metric ruler were used to assess standing pelvic tilt by measuring the angle between the anterior superior iliac spine (ASIS) and the posterior superior iliac spine (PSIS).[23] After taking the measurements, the *sin* of the angle was calculated to determine the number of degrees. Alignment fell into a classification of normal (0 to 10 degrees) or high/anteriorly tilted ($\geq$ to 10 degrees).

After the above measurements, the subject was asked to perform testing trials. Thirty trials of drop landing were performed by each subject in six different test conditions. The six conditions included stiff (ST)and soft (SF) drop landings from three different heights: 45 cm, 60 cm, and 75 cm. The range for the stiff and soft landings was determined by taking an average of three landings in each condition with a standard deviation of $\pm$ 9 degrees.[29]

The subject was instructed to land with the right foot on the force platform and the left foot on the adjacent floor at the same time. Five drop landings in the six test conditions were completed. The order of the landing techniques (soft and stiff) and heights (45, 60, and 75 cm) were randomized for each subject. A diagram of the experimental setup is shown in Figure 2.

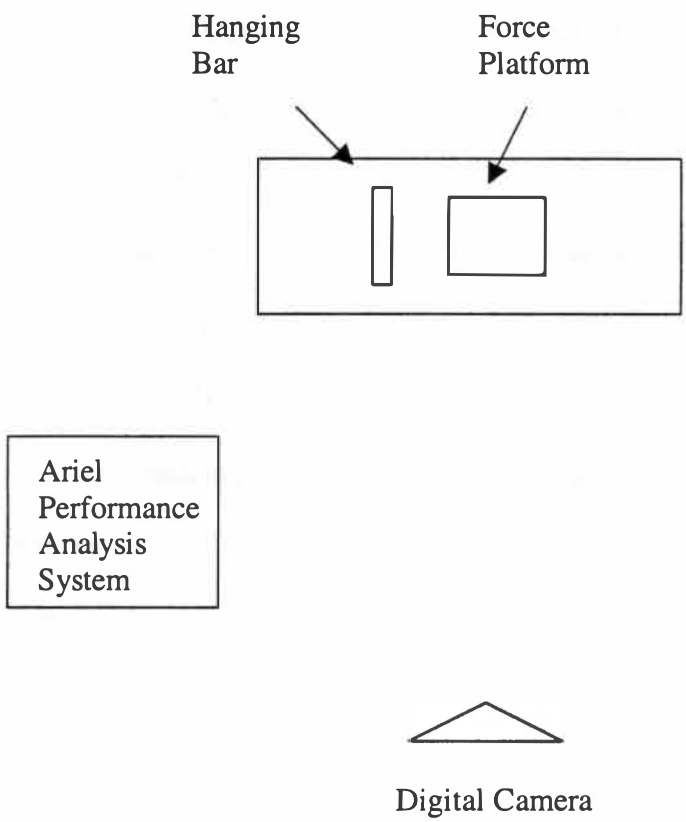

Figure 2. Diagram of Experimental Setup

Data Processing

For the static postural measurements, only the knee extension was used for further analysis due to the nature of the study. For data processing, the procedure was divided into kinetic and kinematic categories.

Kinematic Data

Kinematic variables were obtained from images collected by the digital camera. Data processing occurred in four steps: capturing, digitizing, decoding/smoothing, and computing. A total of 120 frames of video images were captured and stored for each trial on APAS, with 20 frames prior to and 100 frames following foot contact with the force platform. APAS was used to digitize the reflective markers. Digitization was also performed on the reference frame in order to produce scale factors that are needed for conversion of digitized reflective marker coordinates from a screen reference system to a lab reference system. A customized computer program was utilized to decode, smooth, and reconstruct the digitized coordinates. The x and y coordinates of each reflective marker were smoothed individually using an algorithm in order to obtain optimal cutoff frequencies. Finally, a second customized program for computing linear and angular kinematics and determining subsequent discrete events was used. The computed angular kinematic variables included contact position/velocity, maximum and time to maximum position/velocity, and minimum and time to minimum position/velocity for the hip, knee, and ankle joints.

Kinetic Data

Analysis of data collected from the force platform was completed in two steps. First, a customized program was used to decode the analog GRF data file stored on APAS and saved to an ASCII file. Second, the decoded data files were imported to compute and obtain GRF variables via another program. The GRF variables included the first (F1) and second (F2) maximum vertical GRF, the associated times (T1 and T2), and impulse.

Statistical Analysis

For each kinematic and kinetic variable, means and standard deviations were calculated. For the selected variables, a 3x3x2 (group x height x landing technique) repeated measures analysis of variance (ANOVA) was computed using SPSS (SPSS Inc., Chicago, Illinois, USA) statistical package. Significance level was set at ($p < 0.05$). A Tukey procedure was used in post hoc comparisons.

Chapter IV

Results

The purpose of this study was to investigate the relationship between static postural measurements of the lower extremity and biomechanical characteristics during landing activities from three different heights using two different landing techniques. The static measurement showed significantly greater knee extension in the female group with excessive knee extension (EKE-Fem) than the other two groups ($p < 0.05$).

Kinetics

Vertical Ground Reaction Force

A main effect of height, technique and group was found significant for the first (F1) and second (F2) peak GRF, and the loading rate of F1 (LRF1) and F2 (LRF2). Post hoc analyses indicated greater F1 values for NKE-Fem than EKE-Fem in the soft landing (SF) at 45 cm (Table 2, see Appendix D for individual and group tables). Also noted were significantly higher F1 for NKE-Fem than ML in both types of landing at 75 cm, and EKE-Fem than ML in SF at the high height. The results also revealed significantly greater F2 for ML than for EKE-Fem in the soft landing at 45 cm and in the stiff landing (ST) at 75 cm (Table 2). The ML group demonstrated a lower F2 than the NKE-Fem in SF at the high height. The EKE-Fem group also produced significantly lower values than the NKE-Fem group in F2, LRF1 and LRF2 across all six landing conditions.

Table 2. Means and standard deviations of selected VGRF variables

		F1			F2			LRF1			LRF2		
Ht.	Land	ML	NKE-Fem	EKE-Fem	ML	NKE-Fem	EKE-Fem	ML	NKE-Fem	EKE-Fem	ML	NKE-Fem	EKE-Fem
Low	Soft	19.7 [*]	26.5 [b]	19.1 [*]	42.7 [*]	42.7 [b,*]	31.5 [c,*]	1886.4 [a]	2780.8 [b]	1593.7	818.4	1471.1 [b]	402.3
		(4.9)	(13.5)	(3.5)	(13.2)	(15.1)	(5.4)	(675.1)	(1905.6)	(418.3)	(479.2)	(1536.8)	(149.1)
	Stiff	24.0	25.8	24.8	51.7	57.4 [b]	42.3	2172.5	2335.4 [b]	1770.1	1025.8	1535.9 [b]	499.7
		(4.4)	(6.9)	(6.4)	(13.1)	(21.7)	(8.5)	(523.1)	(729.6)	(439.8)	(432.5)	(1230.1)	(176.2)
Med	Soft	25.1 [*]	31.1 [*]	30.7 [*]	46.3 [*]	51.6 [b,*]	41.7 [*]	2306.8 [*]	2914.4 [b,*]	2437.6 [*]	1410.8 [*]	2290.3 [b,*]	624.1 [*]
		(7.1)	(4.8)	(3.7)	(20.1)	(16.7)	(6.6)	(897.8)	(555.2)	(407.6)	(1985.2)	(2937.4)	(201.9)
	Stiff	33.8	37.8	38.3	58.3	68.5 [b]	49.3	3454.2	3646.2 [b]	2850.8 [c]	1424.2	2022.0 [b]	856.3
		(3.4)	(11.6)	(9.5)	(10.9)	(19.4)	(18.8)	(459.7)	(1038.7)	(884.5)	(477.2)	(1393.6)	(625.7)
High	Soft	36.3 [a,*]	48.7	45.3 [c]	56.0 [a,*]	72.5 [b,*]	51.4 [*]	3441.1 [a,*]	4608.6 [b,*]	3557.1 [*]	1440.96 [a]	2380.2 [b,*]	931.9 [*]
		(8.7)	(9.3)	(8.1)	(12.0)	(19.6)	(8.5)	(1311.1)	(739.6)	(561.8)	(551.8)	(1243.9)	(348.5)
	Stiff	42.9 [a]	48.1	45.1	77.8	80.6 [b]	55.8 [c]	4123.7	4953.8 [b]	3653.5	2267.4	2502.7 [b]	1524.6
		(7.7)	(10.9)	(10.7)	(11.8)	(22.9)	(18.8)	(1044.5)	(779.2)	(1056.5)	(605.6)	(1562.6)	(1520.6)

Note: F1 and F2 units are in N/kg and LRF1 and LRF2 units are in N/kg/s. Standard deviation values are in parentheses.
F1: First maximum vertical ground reaction force
F2: Second maximum vertical ground reaction force
LRF1: Loading rate of the first maximum vertical ground reaction force
LRF2: Loading rate of the second maximum vertical ground reaction force
Male –Normal male, NKE-Fem – Normal knee extension female, EKE-Fem – Female with Excessive knee extension.
[a] denotes significant difference between Male and N-Fem in the same landing condition.
[b] denotes significant difference between N-Fem and H-Fem in the same landing condition.
[c] denotes significant difference between Male and H-Fem in the same landing condition.
[*] denotes significant difference between soft and stiff landing at each height.
Comparisons among all heights within the same landing technique and same group were significant except for LRF2 in soft landing across all three heights.

In addition, LRF1 for NKE-Fem was significantly greater than ML in SF at both 45 and 75 cm. EKE-Fem demonstrated lower LRF1 than ML in ST at 60 cm. For LRF2, ML recorded smaller values than NKE-Fem in the soft landing at 75 cm.

For landing stiffness, significant differences were noted between SF and ST techniques in all three heights for F1 in ML, 60 cm NKE-Fem, and 45 & 60 cm EKE-Fem. Landing differences were also noted for each height in F2, and in the medium and high heights for LRF1 and LRF2. Finally, comparisons among all heights were significant for the same group and landing technique except for LRF2 in the soft landing at all three heights.

Kinematics

Knee

The kinematic results of the knee joint indicated a significant main effect of height, technique, and group for contact angle (ContAng), contact velocity (ContVel), maximum velocity (MaxVel), and ROM. Post hoc comparisons showed NKE-Fem to have significantly greater ContAng than ML in SF at 60 cm, in both SF and ST landing at 75 cm, and than EKE-Fem at 75 cm of ST (Table 3, see Appendix C for individual and group tables). NKE-Fem generated significantly greater contact velocities n ML in all conditions except for the soft landing at the medium height. EKE-Fem also showed higher contact velocities across all conditions than ML except for ST at 45 cm. For maximum knee velocity (Table 3), NKE-Fem was greater than

Table 3. Means and standard deviations of selected knee kinematic variables

		ContAng			ContVel			MaxVel			ROM		
Ht.		ML	NKE-Fem	EKE-Fem	ML	NKE-Fem	EKE-Fem	ML	NKE-Fem	EKE-Fem	ML	NKE-Fem	EKE-Fem
Low	Soft	30.9 [*]	28.9 [*]	29.9 [*]	340.2 [a,*]	419.6 [*]	387.4 [c,*]	478.9 [a,*]	580.3 [b,*]	522.9 [*]	64.5 [a,*]	76.5 [*]	78.4 [c,*]
		(4.4)	(7.9)	(5.5)	(38.4)	(96.4)	(42.9)	(83.0)	(54.3)	(30.9)	(17.5)	(10.3)	(8.7)
	Stiff	25.5	22.6	23.2	289.5 [a]	342.8	335.9	439.8 [a]	507.2 [b]	438.9	49.4	51.1	49.5
		(4.4)	(4.7)	(4.3)	(45.7)	(104.5)	(32.0)	(77.1)	(48.7)	(52.9)	(18.0)	(13.5)	(9.1)
Med	Soft	33.3 [*]	29.5 [*]	29.7 [*]	395.5 [*]	436.4 [*]	388.3 [*]	529.3 [a,*]	620.1 [b,*]	563.5 [*]	73.9 [a,*]	80.4 [*]	78.9 [*]
		(4.4)	(7.2)	(7.1)	(51.3)	(112.1)	(120.7)	(73.7)	(60.9)	(43.9)	(10.7)	(6.4)	(8.0)
	Stiff	27.8 [a]	23.6	25.9	337.7 [a]	392.7	388.9 [c]	502.0 [a]	552.9	513.6	46.5 [a]	55.1	55.7 [c]
		(2.8)	(5.2)	(3.9)	(41.3)	(76.4)	(33.4)	(72.3)	(40.4)	(52.6)	(8.3)	(11.0)	(9.5)
High	Soft	32.8 [a,*]	28.2 [*]	32.3 [*]	390.3 [a,*]	479.3 [*]	458.5 [c,*]	579.7 [a,*]	652.5 [b,*]	603.2 [*]	78.7 [*]	76.9 [*]	78.9 [*]
		(4.8)	(7.7)	(3.8)	(45.6)	(108.1)	(27.5)	(63.8)	(54.5)	(33.7)	(8.9)	(8.4)	(5.4)
	Stiff	29.0 [a]	24.3 [b]	30.3	360.3 [a]	432.7	400.3 [c]	538.8 [a]	607.1 [b]	538.4	51.6 [a]	63.5	56.4
		(3.8)	(3.3)	(3.5)	(48.3)	(63.1)	(38.1)	(86.3)	(50.3)	(62.5)	(10.5)	(15.5)	(11.6)

Note: ContAng and ROM units are in degrees and ContVel and MaxVel are in m/s. Standard deviation values are in parentheses.
ContAng: Contact joint angle at ground contact
ContVel: Angular joint velocity at ground contact
Max Vel: Angular joint maximum velocity
ROM: Range of motion of joint
Male – Normal male, NKE-Fem – Normal knee extension female, EKE-Fem – Female with Excessive knee extension.
[a] denotes significant difference between Male and N-Fem in the same landing condition.
[b] denotes significant difference between N-Fem and H-Fem in the same landing condition.
[c] denotes significant difference between Male and H-Fem in the same landing condition.
[*] denotes significant difference between soft and stiff landing at each height.
Comparisons among all heights within the same landing technique and same group were significant except for ContAng in soft landing across all three heights.

ML across all heights and landing conditions, and EKE-Fem in 5 out of 6 conditions (except ST at 60 cm). Knee ROM variables were significantly less for ML and EKE-Fem than for NKE-Fem in the soft and stiff landing at 45 cm and 60 cm respectively. NKE-Fem also produced greater ROM values than ML in SF at 60 cm and ST at 75 cm. Significant differences in landing technique were observed at each height across all the groups for all variables. Height differences for the same landing technique were all significant except for ContAng in SF at 45, 60, and 75 cm.

<u>Ankle</u>

For the ankle joint, a main effect of height and group was found for ContAng, ROM, ContVel, and MaxVel (technique for ROM). In post hoc comparisons, ML displayed significantly lower values for ContAng and ROM than EKE-Fem in all conditions, and NKE-Fem for ContAng in ST at both 60 and 75 cm, and for ROM in ST at 60 cm (Table 4, see Appendix C for individual and group tables). EKE-Fem also generated greater ContAng and ROM than NKE-Fem across all conditions, except for ContAng in SF at 75 cm.

<u>Hip</u>

Statistical analyses of the hip related variables (Table 5, see Appendix C for individual and group tables) demonstrated greater ContVel for NKE-Fem than ML at 45 cm (SF and ST), 60 cm (ST), and 75 cm (both SF and ST). EKE-Fem had lower contact velocities than NKE-Fem at 45 cm (SF and ST) and 75 cm (ST). For MaxVel,

Table 4. Means and standard deviations of selected ankle kinematic variables.

		ContAng			ContVel			MaxVel			ROM		
Ht.	Land	ML	NKE-Fem	EKE-Fem	ML	NKE-Fem	EKE-Fem	ML	NKE-Fem	EKE-Fem	ML	NKE-Fem	EKE-Fem
	Soft	-12.0 [*]	-14.2 [b,*]	-23.6 [c,*]	363.2 [*]	405.9 [*]	435.3 [c,*]	423.4 [*]	459.1 [b,*]	536.3 [c,*]	35.1	36.6 [b]	49.7 [c]
Low		(4.1)	(14.0)	(2.4)	(48.8)	(134.9)	(84.8)	(60.2)	(161.5)	(52.5)	(6.0)	(15.8)	(2.7)
	Stiff	-14.0	-19.1 [b]	-28.0 [c]	394.2	423.7	458.8 [c]	457.9 [a]	513.8 [b]	587.3 [c]	35.3	41.1 [b]	48.7 [c]
		(4.2)	(11.7)	(4.1)	(40.3)	(129.9)	(59.2)	(51.8)	(101.1)	(44.8)	(6.3)	(12.4)	(5.0)
	Soft	-14.0 [*]	-15.3 [b,*]	-24.6 [c,*]	388.8 [*]	431.9 [*]	414.0 [*]	457.5 [*]	491.4 [*]	560.8 [c,*]	36.8	39.1 [b]	50.3 [c]
Med		(6.4)	(13.0)	(8.0)	(93.7)	(133.0)	(158.7)	(94.6)	(154.2)	(37.5)	(6.7)	(13.6)	(7.3)
	Stiff	-15.0 [a]	-21.5 [b]	-27.2 [c]	398.1 [a]	501.5	506.6 [c]	467.8 [a]	575.4	616.4 [c]	36.9 [a]	43.0 [b]	49.0 [c]
		(3.1)	(9.2)	(5.3)	(43.9)	(87.3)	(69.4)	(48.8)	(92.7)	(48.0)	(4.5)	(9.0)	(4.6)
	Soft	-15.0	-19.5	-24.4 [c]	434.2 [a]	491.5	498.9 [c]	489.0 [a,*]	549.5	587.6 [c,*]	39.0	40.8 [b]	49.5 [c]
High		(4.8)	(11.4)	(4.7)	(41.5)	(101.9)	(39.5)	(47.0)	(121.8)	(23.5)	(5.9)	(11.4)	(4.0)
	Stiff	-15.0 [a]	-20.9 [b]	-26.1 [c]	439.0 [a]	493.0	496.7 [c]	504.8 [a]	556.5 [b]	610.2 [c]	38.2	40.0 [b]	48.0 [c]
		(3.4)	(8.6)	(5.9)	(38.4)	(80.9)	(73.7)	(45.5)	(83.0)	(67.3)	(5.4)	(7.5)	(6.0)

Note: ContAng and ROM units are in degrees and ContVel and MaxVel units are in m/s. Standard deviation values are in parentheses.
ContAng: Contact joint angle at ground contact
ContVel: Angular joint velocity at ground contact
MaxVel: Angular joint maximum velocity
ROM: Range of motion of joint
Male –Normal male, NKE-Fem – Normal knee extension female, EKE-Fem – Female with Excessive knee extension.
[a] denotes significant difference between Male and N-Fem in the same landing condition.
[b] denotes significant difference between N-Fem and H-Fem in the same landing condition.
[c] denotes significant difference between Male and H-Fem in the same landing condition.
[*] denotes significant difference between soft and stiff landing at each height.
Comparisons among all heights within the same landing technique and same group were significant except for ContAng and ROM of ankle in stiff landing.

Table 5. Means and standard deviations of selected hip kinematic variables.

		ContAng			ContVel			MaxVel			ROM		
Ht.	Land	ML	NKE-Fem	EKE-Fem	ML	NKE-Fem	EKE-Fem	ML	NKE-Fem	EKE-Fem	ML	NKE-Fem	EKE-Fem
	Soft	23.3 *	25.9 *	25.3 *	199.8 ^{a,*}	281.2 ^{b,*}	224.7 *	341.2 *	432.1 *	362.0 *	60.2	74.9	70.7
Low		(7.2)	(15.2)	(7.3)	(55.5)	(100.8)	(40.2)	(101.9)	(60.2)	(27.0)	(25.3)	(14.8)	(20.6)
	Stiff	19.9	17.4	13.8 ^{c}	148.5 ^{a}	187.4 ^{b}	141.1	241.7 ^{a}	297.8 ^{b}	224.8	29.9	30.5	25.7
		(3.5)	(8.7)	(4.6)	(39.8)	(55.1)	(45.5)	(66.6)	(57.9)	(48.1)	(15.6)	(12.2)	(7.8)
	Soft	29.0 *	25.9 *	24.2 *	243.9 *	287.2 *	230.0 *	378.5 ^{a,*}	459.4 ^{b,*}	402.1 *	73.0	81.5 ^{b}	67.9
Med		(4.5)	(14.0)	(4.0)	(53.5)	(92.1)	(81.1)	(67.0)	(53.9)	(43.8)	(12.6)	(12.7)	(11.7)
	Stiff	21.6	18.9	17.5	176.3 ^{a}	213.2	186.9	271.0 ^{a}	324.5	302.2	26.0 ^{a}	33.9	35.2 ^{c}
		(4.5)	(9.1)	(4.2)	(49.3)	(42.9)	(40.4)	(73.8)	(59.1)	(47.3)	(9.2)	(10.7)	(7.9)
	Soft	28.8 *	25.1 *	26.6 *	249.1 ^{a,*}	308.2 *	264.6 *	399.7 ^{a,*}	452.4 *	419.3 *	75.5	74.1	69.5
High		(5.4)	(15.2)	(4.8)	(47.7)	(96.2)	(56.6)	(70.9)	(78.1)	(34.7)	(10.7)	(20.7)	(11.3)
	Stiff	22.2	21.1	21.7	199.8 ^{a}	254.1 ^{b}	209.0	297.3 ^{a}	380.5 ^{b}	301.8	30.0 ^{a}	49.3 ^{b}	34.4
		(4.2)	(9.0)	(4.0)	(59.5)	(30.6)	(40.5)	81.6)	(71.5)	(56.6)	(10.8)	(28.3)	(8.7)

Note: ContAng and ROM units are in degrees and ContVel and MaxVel units are in m/s. Standard deviation values are in parentheses.
ContAng: Contact joint angle at ground contact
ContVel: Angular joint velocity at ground contact
MaxVel: Angular joint maximum velocity
ROM: Range of motion of joint
Male –Normal male, NKE-Fem – Normal knee extension female, EKE-Fem – Female with Excessive knee extension.
^{a}denotes significant difference between Male and N-Fem in the same landing condition.
^{b}denotes significant difference between N-Fem and H-Fem in the same landing condition.
^{c}denotes significant difference between Male and H-Fem in the same landing condition.
*denotes significant difference between soft and stiff landing at each height.
Comparisons among all heights within the same landing technique and same group were significant except for MaxVel of hip in soft landing.

NKE-Fem showed greater values than both ML and EKE-Fem at 45 cm of ST, 60 cm of SF, and 75 cm of ST. NKE-Fem also had greater maximum velocities than ML in ST at 60 cm and SF at 75 cm. Differences in landing technique were significant across all subject groups for ContAng, ContVel, MaxVel, and ROM at each height, while height differences within the same landing technique and group were all significant except for MaxVel in SF across all three heights.

Chapter V

Discussion

The purpose of the study was to demonstrate differences in landing kinematics and vertical ground reaction forces (VGRF) between males and females. Because of the high incidence of knee injuries in females,[2] biomechanics of landing was the main focus in this study. ACL strain behavior has been shown to be at its greatest from 30 degrees of flexion to full knee extension.[3,10] The results generally demonstrated greater F1, F2, LRF1 and LRF2 values across the groups as the height and landing stiffness increased.. These values are important when examining the functional implications for possible mechanisms of injury. The initial impact forces and loading rates during landing are absorbed up through the kinetic chain and may cause damage at the knee.

On average, of all the groups, the normal knee extension females displayed higher mean peak values for the selected VGRF variables independent of landing height or technique. The NKE group produced greater F1 (48.7 N/kg in SF & 48.1 N/kg in ST) and F2 (72.5 N/kg in SF & 80.6 N/kg in ST) than its male counterpart (F1 - 36.3 N/kg in SF & 42.9 N/kg in ST, F2 -56.0 N/kg in SF & 77.8 N/kg in ST) in both soft and stiff landings from 75 cm. This same pattern was seen in LRF1. The normal females generated a loading rate that was significantly greater than the males. These findings suggest possible differences in landing strategies adopted by the two gender groups. It has been speculated that weakness of the hamstrings as a knee stabilizer during landing may predispose females to ACL injury.[2,8] Hewett et al.[12]

observed three times greater overall hamstring activity in males than females while landing. It is thought that the enhanced activity helped to decrease peak impact forces. Similarly, McNair and Marshall[17] observed a significant correlation between lower peak VGRF and lateral hamstring muscle activity in landing. Although hamstring activity was not evaluated in this study, the higher GRF data in normal females may indicate an inability to absorb impact forces secondary to weak hamstring muscles in this group.

Significant differences for the selected GRF variables were also noted between the ML and EKE-Fem groups. The males generated 77.8 N/kg in F2 compared to 55.8 N/kg in the excessive knee extension female group in the stiff landing from 75 cm. This group comparison difference was also noted in LRF1 for a stiff landing at 60 cm. When compared to the NKE-Fem group, the EKE-Fem produced significantly lower values of F2, LRF1 and LRF2 across all landing conditions. These significant differences across the two female groups were somewhat surprising given the similarities in the reported activity levels and anthropometric measurements (circumferences and lengths of lower extremity). The EKE-Fem group had a greater degree of unilateral standing knee extension, but this would not be thought to elicit such disparity in the impact related measures. Perhaps subjects in this group implemented a compensatory strategy in landing for their increased knee laxity. In the ankle, EKE-Fem landed with significantly greater planterflexion at contact than either NKE-Fem or ML. Landing with increased plantarflexion might facilitate the toe-heel landing, which would appear to help decrease the VGRF.[7] This landing pattern at the ankle, adopted by EKE-Fem, could be an adaptive strategy whereby the subjects try to use the distal musculature to help decrease impact at the knee.

In this study, EKE-Fem landed with greater knee flexion contact angles on average than NKE-Fem, possibly avoiding end-range knee extension. In assessing sagittal knee position in ACL-injured females, Loudon et al.[16] hypothesized that individuals with genu recurvatum already had an applied load to the ACL; any additional stress to the knee, such as landing from a jump, might overload the ligament. If this is the case, then these females might subconsciously land cautiously in order to prevent ACL overload. Caster and Bates[5] support the idea of protective response mechanisms in which individuals sense a mechanical threat to the system. The authors reported that two of the four subjects implemented a Newtonian strategy and responded to the addition of mass to the body with increased peak GRF, while the other two adopted a neuromuscular strategy and ignored the added mass. The maximum knee angular velocity showed significant lower values for the EKE-Fem group across almost all conditions. Decreased hip maximum angular velocities and increased ankle ROM for this group also suggest a different strategy was used in landing.

Because landing has been shown to be a critical factor for non-contact ACL injury in females,[9,11] the contact angle, contact velocity, maximum velocity and ROM of the lower extremity were examined. The normal knee extension females landed with significantly greater knee extension at contact than ML at medium and high heights, and significantly greater contact velocity, maximum velocity and ROM across almost all landing conditions. Similar results were also observed in the contact and maximum velocities of the hip joint. In our study, NKE-Fem displayed significantly higher hip ROM than ML during stiff landing at medium and high heights. Zhang et al.[29] showed that landing posture became more flexed with increased height and mechanical demand, allowing the hip extensors to have a greater advantage in energy dissipation. The combination of increased knee extension and hip flexion with corresponding contact and maximal velocities of the knee and hip in N-Fem question the neuromuscular deceleration of this group during large mechanical loads.

The belief that increased genu recurvatum might predispose females to an ACL injury has been reported.[11,16] With this understanding there was an underlying interest of how these females would respond to different landing techniques at different heights when compared to a normal (knee extension) female group and a normal male group. It was hypothesized that EKE-Fem would produce significantly greater kinetic and kinematic values than both ML and NKE-Fem. The present study demonstrates more significant differences between NKE-Fem and EKE-Fem in selected VGRF, knee MaxVel, and ankle contact and ROM. On the whole, EKE-Fem performed similar to ML across the variables except for ankle contact and ROM and knee contact velocity.

The biggest surprise of the findings might have been the manner in which NKE-Fem landed from the medium and high heights with increased stiffness. The lower initial contact angle was associated with greater speeds of contact, which could potentially overload the knee and make it difficult to overcome the forces absorbed by the body. In the EKE-Fem group, there appeared to be a tendency to avoid full knee extension through compensatory measures of the ankle and hip. These compensations could decrease stresses applied to the knee. One note to mention is the fact that these forces are only being evaluated in the sagittal plane, and do not even take into consideration what is taking place in the transverse and frontal planes. This would add a whole new dimension to the stresses placed upon the knee that are normally noted for ACL MOI.[14,15]

Future studies should consider the multi-planar mechanisms of the lower extremity in landing using 3-D analysis. This would provide the researcher a more comprehensive view of the effect of subtalar joint pronation and hip internal rotation on the knee joint biomechanics in landing. The lower extremity joints move in three different planes and as such, injuries may be most likely caused by a combination of these motions. Another area of future research should focus on the roles of the gluteals and hamstrings on helping to decelerate the knee while landing. The results from our study display some gender differences in the way the landing was performed and controlled from a neuromuscular and biomechanical standpoint of view. Better understanding of neuromuscular controls in lower extremity mechanics during landing is warranted in order to prescribe preventative ACL programs.

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APPENDIX A

DEFINITIONS OF VARIABLES

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Kinematics

ContAng	Contact joint angle at ground contact
Max	Maximum joint angle
Tmax	Time to maximum joint angle
Min	Minimum joint angle
Tmin	Time to minimum joint angle
ROM	Range of motion of joint
ContVel	Angular joint velocity at ground contact
MaxVel	Angular joint maximum velocity
TmaxVel	Time to angular joint maximum velocity

VGRF

F1	First maximum vertical ground reaction force
T1	Time to first maximum vertical ground reaction force
F2	Second maximum vertical ground reaction force
T2	Time to second maximum vertical ground reaction force
LRF1	Loading rate of the first maximum vertical ground reaction force
LRF2	Loading rate of the second maximum vertical ground reaction force
Imp100ms	Impulse of vertical ground reaction force from contact to 100 ms

APPENDIX B

SUBJECT INFORMATION

Table 6. Subject information

Group	Subject	Age	Body Weight (kg)	Height (cm)
1	1	23	66.5	179.7
	2	25	87.0	185.4
	3	24	65.6	182.9
	4	22	90.5	177.8
	5	23	66.5	165.1
Mean		23.4	75.2	178.2
S.D.		1.1	12.4	7.9
2	1	23	62.7	171.4
	2	23	55.8	157.5
	3	22	57.5	167.6
	4	18	57.4	163.8
	5	25	65	172.7
Mean		22.2	59.7	166.6
S.D.		2.6	3.9	6.2
3	1	24	53.2	167.6
	2	25	56.8	172.7
	3	21	58.7	162.6
	4	21	58.2	162.6
	5	18	60.4	168.3
	6	21	58.6	167.6
Mean		21.7	57.6	166.9
S.D.		2.5	2.5	3.8

APPENDIX C

KINEMATIC DATA

Table 7. Subject means and standard deviations of ankle joint variables (Group 1).

Sub	Cond	ContAng	Max	Tmax	Min	Tmin	ROM	ContVel	MaxVel	TmaxVel
7	1	-10.321	22.583	0.146	-10.321	0.000	32.904	360.941	415.613	0.023
		(2.5)	(2.3)	(0.0)	(2.5)	(0.0)	(2.4)	(51.7)	(47.1)	(0.0)
	2	-14.880	28.449	0.329	-14.880	0.000	43.329	370.759	466.541	0.030
		(2.3)	(1.0)	(0.0)	(2.3)	(0.0)	(1.5)	(30.7)	(49.6)	(0.0)
	3	-9.621	25.196	0.240	-9.621	0.000	34.817	392.241	439.061	0.020
		(2.0)	(1.9)	(0.1)	(2.0)	(0.0)	(3.5)	(49.4)	(65.2)	(0.0)
	4	-10.118	26.218	0.410	-10.118	0.000	36.336	340.674	394.279	0.023
		(6.5)	(0.7)	(0.3)	(6.5)	(0.0)	(6.2)	(76.4)	(90.6)	(0.0)
	5	-14.387	13.813	0.224	-14.387	0.000	28.201	351.604	401.697	0.020
		(3.2)	(1.4)	(0.2)	(3.2)	(0.0)	(2.0)	(20.6)	(16.0)	(0.0)
	6	-11.412	27.865	0.266	-11.412	0.000	39.277	412.501	472.386	0.021
		(4.3)	(1.9)	(0.1)	(4.3)	(0.0)	(5.3)	(42.3)	(27.6)	(0.0)
8	1	-18.086	20.490	0.124	-18.086	0.000	38.576	416.409	520.757	0.027
		(2.1)	(1.7)	(0.0)	(2.1)	(0.0)	(2.5)	(30.7)	(50.7)	(0.0)
	2	-10.970	18.307	0.113	-10.970	0.000	29.277	402.969	443.426	0.018
		(2.5)	(2.9)	(0.0)	(2.5)	(0.0)	(4.0)	(28.6)	(26.6)	(0.0)
	3	-18.665	22.257	0.208	-18.665	0.000	40.922	399.823	465.014	0.025
		(2.6)	(4.0)	(0.1)	(2.6)	(0.0)	(2.8)	(25.1)	(14.9)	(0.0)
	4	-12.852	15.531	0.161	-12.852	0.000	28.383	339.526	387.668	0.022
		(1.8)	(1.3)	(0.0)	(1.8)	(0.0)	(1.0)	(25.9)	(12.9)	(0.0)
	5	-15.422	24.096	0.176	-15.422	0.000	39.518	427.190	501.045	0.023
		(2.6)	(0.8)	(0.1)	(2.6)	(0.0)	(2.4)	(14.3)	(31.8)	(0.0)
	6	-13.870	28.071	0.152	-14.335	0.165	41.941	436.368	527.829	0.027
		(1.5)	(1.2)	(0.0)	(2.3)	(0.4)	(1.7)	(48.0)	(57.8)	(0.0)
9	1	-10.270	26.353	0.228	-10.270	0.000	36.624	386.255	439.305	0.023
		(2.6)	(1.0)	(0.2)	(2.6)	(0.0)	(2.9)	(37.9)	(33.2)	(0.0)
	2	-10.255	21.297	0.569	-10.255	0.000	31.551	285.527	347.900	0.031
		(12.6)	(3.6)	(0.4)	(12.6)	(0.0)	(12.6)	(169.2)	(153.9)	(0.0)
	3	-19.326	14.806	0.155	-19.326	0.000	34.132	408.554	471.556	0.023
		(1.9)	(2.4)	(0.1)	(1.9)	(0.0)	(2.7)	(39.4)	(17.2)	(0.0)
	4	-15.493	22.306	0.147	-15.493	0.000	37.798	434.211	493.754	0.020
		(2.9)	(3.3)	(0.0)	(2.9)	(0.0)	(3.7)	(26.4)	(24.9)	(0.0)
	5	-17.644	22.775	0.124	-17.644	0.000	40.418	434.247	531.451	0.027
		(1.5)	(1.3)	(0.0)	(1.5)	(0.0)	(2.4)	(48.6)	(36.0)	(0.0)
	6	-10.894	21.120	0.133	-10.894	0.000	32.014	373.690	423.093	0.022
		(2.3)	(2.6)	(0.0)	(2.3)	(0.0)	(2.0)	(27.7)	(28.0)	(0.0)

Table 7. (Continued)

Sub	Cond	ContAng	Max	Tmax	Min	Tmin	ROM	ContVel	MaxVel	TmaxVel
13	1	-16.117	25.187	0.183	-16.117	0.000	41.304	390.152	465.972	0.026
		(1.0)	(2.5)	(0.0)	(1.0)	(0.0)	(2.1)	(25.3)	(20.0)	(0.0)
	2	-16.691	16.253	0.169	-16.691	0.000	32.944	358.312	424.564	0.025
		(2.3)	(1.8)	(0.0)	(2.3)	(0.0)	(2.0)	(32.0)	(21.5)	(0.0)
	3	-14.496	25.486	0.140	-14.496	0.000	39.982	462.273	532.381	0.021
		(1.6)	(2.0)	(0.0)	(1.6)	(0.0)	(2.6)	(48.1)	(29.1)	(0.0)
	4	-18.808	24.428	0.130	-18.808	0.000	43.236	435.720	557.782	0.029
		(0.8)	(2.3)	(0.0)	(0.8)	(0.0)	(2.8)	(27.1)	(26.0)	(0.0)
	5	-10.138	23.035	0.146	-10.138	0.000	33.173	420.336	457.031	0.017
		(1.7)	(3.1)	(0.0)	(1.7)	(0.0)	(3.6)	(41.1)	(37.3)	(0.0)
	6	-17.167	24.790	0.168	-17.167	0.000	41.956	440.537	499.059	0.020
		(2.6)	(1.3)	(0.0)	(2.6)	(0.0)	(3.2)	(41.7)	(21.2)	(0.0)
16	1	-14.878	17.836	0.171	-14.878	0.000	32.714	436.086	477.726	0.016
		(1.9)	(1.5)	(0.0)	(1.9)	(0.0)	(3.4)	(35.4)	(28.1)	(0.0)
	2	-13.403	24.319	0.241	-13.403	0.000	37.722	470.157	524.997	0.019
		(1.4)	(1.8)	(0.1)	(1.4)	(0.0)	(2.6)	(21.2)	(19.7)	(0.0)
	3	-18.966	27.583	0.307	-18.966	0.000	46.550	439.616	523.917	0.026
		(1.9)	(2.5)	(0.0)	(1.9)	(0.0)	(4.1)	(51.2)	(49.7)	(0.0)
	4	-8.476	28.827	0.289	-8.476	0.000	37.304	403.973	443.058	0.019
		(3.0)	(1.4)	(0.1)	(3.0)	(0.0)	(3.0)	(51.7)	(43.1)	(0.0)
	5	-20.032	21.768	0.528	-20.032	0.000	41.801	440.066	490.633	0.019
		(3.8)	(1.2)	(0.3)	(3.8)	(0.0)	(4.2)	(31.8)	(37.5)	(0.0)
	6	-16.052	15.425	0.125	-16.052	0.000	31.477	417.186	462.564	0.019
		(1.1)	(1.0)	(0.0)	(1.1)	(0.0)	(1.9)	(19.3)	(21.6)	(0.0)

Note: Angle and ROM units are in degrees and time unit is in seconds.
Velocity unit is in deg/s.
Standard deviation values are in parentheses.
The definitions of variables are in Appendix A.

Table 8. Subject means and standard deviations of ankle joint variables (Group 2).

Sub	Cond	ContAng	Max	Tmax	Min	Tmin	ROM	ContVel	MaxVel	TmaxVel
4	1	-19.337	19.196	0.181	-19.337	0.000	38.534	369.331	458.497	0.027
		(2.5)	(1.1)	(0.0)	(2.5)	(0.0)	(2.1)	(12.7)	(50.5)	(0.0)
	2	8.446	20.629	0.275	4.440	0.495	12.183	210.541	212.746	0.003
		(1.3)	(3.3)	(0.2)	(4.9)	(0.5)	(3.2)	(12.9)	(15.3)	(0.0)
	3	-28.407	25.665	0.220	-28.407	0.000	54.072	541.864	625.643	0.025
		(2.8)	(1.2)	(0.1)	(2.8)	(0.0)	(2.4)	(26.8)	(29.0)	(0.0)
	4	-17.678	24.106	0.166	-17.678	0.000	41.784	501.677	539.457	0.018
		(2.3)	(1.2)	(0.0)	(2.3)	(0.0)	(1.9)	(37.7)	(41.3)	(0.0)
	5	-22.188	20.761	0.188	-22.188	0.000	42.949	276.927	467.255	0.046
		(7.7)	(1.2)	(0.0)	(7.7)	(0.0)	(7.7)	(149.5)	(69.3)	(0.0)
	6	-1.345	21.540	0.102	-1.345	0.000	22.885	378.393	393.886	0.011
		(6.7)	(1.8)	(0.0)	(6.7)	(0.0)	(5.7)	(52.2)	(59.5)	(0.0)
11	1	-26.252	27.049	0.193	-26.252	0.000	53.302	534.929	617.508	0.025
		(1.8)	(1.4)	(0.0)	(1.8)	(0.0)	(1.6)	(20.3)	(20.5)	(0.0)
	2	-26.472	18.692	0.137	-26.472	0.000	45.163	504.428	576.441	0.023
		(1.7)	(2.9)	(0.0)	(1.7)	(0.0)	(2.4)	(39.8)	(26.7)	(0.0)
	3	-16.472	20.482	0.172	-16.472	0.000	36.954	360.353	437.293	0.027
		(4.4)	(0.7)	(0.0)	(4.4)	(0.0)	(4.0)	(38.1)	(43.5)	(0.0)
	4	4.805	24.377	0.307	4.805	0.000	19.572	279.439	288.679	0.010
		(4.3)	(2.0)	(0.3)	(4.3)	(0.0)	(4.4)	(93.6)	(90.3)	(0.0)
	5	-27.831	26.985	0.202	-27.831	0.000	54.816	571.366	659.649	0.024
		(1.6)	(1.5)	(0.0)	(1.6)	(0.0)	(2.3)	(47.5)	(33.6)	(0.0)
	6	-21.783	23.381	0.151	-21.783	0.000	45.165	516.516	579.914	0.022
		(1.5)	(1.5)	(0.0)	(1.5)	(0.0)	(2.5)	(53.1)	(39.3)	(0.0)
12	1	-25.747	19.916	0.148	-25.747	0.000	45.663	455.455	573.845	0.027
		(1.4)	(1.5)	(0.0)	(1.4)	(0.0)	(2.8)	(61.5)	(53.2)	(0.0)
	2	-6.602	22.082	0.111	-6.602	0.000	28.684	404.743	444.056	0.018
		(3.1)	(1.1)	(0.0)	(3.1)	(0.0)	(3.3)	(42.8)	(42.7)	(0.0)
	3	-25.419	24.520	0.188	-25.419	0.000	49.940	571.624	633.971	0.019
		(2.4)	(1.5)	(0.0)	(2.4)	(0.0)	(3.4)	(55.3)	(56.2)	(0.0)
	4	-28.206	19.354	0.126	-28.206	0.000	47.560	573.997	649.580	0.021
		(2.2)	(4.4)	(0.0)	(2.2)	(0.0)	(2.4)	(21.3)	(17.8)	(0.0)
	5	-21.382	17.801	0.173	-21.382	0.000	39.183	420.524	500.482	0.024
		(1.9)	(1.1)	(0.0)	(1.9)	(0.0)	(2.4)	(60.8)	(49.8)	(0.0)
	6	-7.785	21.286	0.122	-7.785	0.000	29.071	433.256	467.957	0.016
		(4.1)	(4.3)	(0.0)	(4.1)	(0.0)	(3.4)	(47.2)	(38.3)	(0.0)

Table 8. (Continued).

Sub	Cond	ContAng	Max	Tmax	Min	Tmin	ROM	ContVel	MaxVel	TmaxVel
14	1	-29.169	17.585	0.152	-29.169	0.000	46.754	577.030	645.100	0.021
		(1.0)	(2.4)	(0.1)	(1.0)	(0.0)	(3.3)	(39.3)	(31.4)	(0.0)
	2	-25.222	19.686	0.140	-25.222	0.000	44.908	541.242	612.599	0.022
		(2.4)	(2.2)	(0.0)	(2.4)	(0.0)	(1.5)	(29.9)	(17.4)	(0.0)
	3	-27.028	17.821	0.175	-27.028	0.000	44.849	447.694	553.213	0.026
		(2.6)	(2.2)	(0.1)	(2.6)	(0.0)	(1.7)	(27.0)	(5.1)	(0.0)
	4	-1.012	21.250	0.261	-1.012	0.000	22.262	367.957	370.020	0.004
		(1.6)	(1.3)	(0.2)	(1.6)	(0.0)	(1.3)	(54.8)	(53.9)	(0.0)
	5	-28.329	22.510	0.260	-28.329	0.000	50.840	576.967	654.364	0.022
		(2.0)	(0.9)	(0.1)	(2.0)	(0.0)	(2.2)	(53.7)	(65.5)	(0.0)
	6	-21.576	23.769	0.149	-21.576	0.000	45.346	573.432	620.489	0.019
		(2.2)	(2.1)	(0.0)	(2.2)	(0.0)	(3.1)	(61.2)	(54.1)	(0.0)

Note: Angle and ROM units are in degrees and time unit is in seconds.
Velocity unit is in deg/s.
Standard deviation values are in parentheses.
The definitions of variables are in Appendix A.

Table 9. Subject means and standard deviations of ankle joint variables (Group 3).

Sub	Cond	ContAng	Max	Tmax	Min	Tmin	ROM	ContVel	MaxVel	TmaxVel
5	1	-23.911	26.777	0.379	-23.911	0.000	50.688	354.514	513.645	0.035
		(2.0)	(1.2)	(0.1)	(2.0)	(0.0)	(0.9)	(43.0)	(51.8)	(0.0)
	2	-25.209	26.102	0.279	-25.209	0.000	51.310	369.365	502.425	0.037
		(2.9)	(1.1)	(0.1)	(2.9)	(0.0)	(3.7)	(12.4)	(36.8)	(0.0)
	3	-22.878	25.689	0.385	-22.878	0.000	48.567	516.617	581.575	0.021
		(2.5)	(2.3)	(0.2)	(2.5)	(0.0)	(2.7)	(61.6)	(54.0)	(0.0)
	4	-22.315	25.780	0.252	-22.315	0.000	48.095	500.631	547.678	0.020
		(1.4)	(1.9)	(0.1)	(1.4)	(0.0)	(1.8)	(33.7)	(36.9)	(0.0)
	5	-29.917	23.635	0.248	-29.917	0.000	53.552	412.398	617.968	0.034
		(3.9)	(2.6)	(0.1)	(3.9)	(0.0)	(5.5)	(65.7)	(59.2)	(0.0)
	6	-30.872	14.527	0.152	-30.872	0.000	45.399	414.385	559.745	0.031
		(1.4)	(3.3)	(0.0)	(1.4)	(0.0)	(3.5)	(22.5)	(42.2)	(0.0)
6	1	-23.074	26.474	0.145	-23.074	0.000	49.547	496.326	595.641	0.028
		(1.4)	(2.6)	(0.0)	(1.4)	(0.0)	(3.6)	(26.1)	(30.0)	(0.0)
	2	-28.044	18.216	0.268	-28.044	0.000	46.260	512.283	576.034	0.023
		(3.8)	(1.0)	(0.2)	(3.8)	(0.0)	(3.5)	(25.9)	(31.5)	(0.0)
	3	-29.316	25.534	0.308	-29.316	0.000	54.851	420.975	563.399	0.032
		(1.9)	(2.9)	(0.0)	(1.9)	(0.0)	(2.4)	(48.2)	(34.7)	(0.0)
	4	-32.840	23.708	0.329	-32.917	0.004	56.548	260.922	578.403	0.063
		(8.4)	(1.7)	(0.1)	(8.5)	(0.0)	(9.6)	(267.0)	(28.6)	(0.0)
	5	-19.105	26.036	0.351	-19.105	0.000	45.141	486.355	571.430	0.024
		(2.4)	(0.7)	(0.3)	(2.4)	(0.0)	(2.1)	(44.6)	(43.6)	(0.0)
	6	-17.232	27.463	0.325	-17.232	0.000	44.694	487.568	529.792	0.019
		(1.5)	(1.8)	(0.1)	(1.5)	(0.0)	(1.5)	(37.4)	(32.0)	(0.0)
10	1	-31.148	22.622	0.224	-31.148	0.000	53.769	450.132	604.250	0.032
		(2.0)	(1.8)	(0.1)	(2.0)	(0.0)	(1.8)	(71.4)	(37.6)	(0.0)
	2	-30.557	17.620	0.134	-30.557	0.000	48.177	505.366	635.815	0.028
		(1.9)	(3.3)	(0.0)	(1.9)	(0.0)	(4.6)	(64.1)	(49.2)	(0.0)
	3	-19.201	25.213	0.175	-19.201	0.000	44.414	509.944	587.916	0.022
		(2.0)	(3.1)	(0.1)	(2.0)	(0.0)	(3.6)	(55.1)	(58.3)	(0.0)
	4	-27.860	21.706	0.155	-27.860	0.000	49.565	561.090	637.690	0.023
		(2.8)	(1.1)	(0.0)	(2.8)	(0.0)	(2.3)	(54.0)	(38.3)	(0.0)
	5	-28.821	23.341	0.253	-28.821	0.000	52.163	447.814	610.149	0.032
		(2.2)	(2.7)	(0.0)	(2.2)	(0.0)	(3.6)	(92.7)	(67.0)	(0.0)
	6	-25.189	18.571	0.184	-25.189	0.000	43.760	478.238	557.869	0.024
		(2.2)	(4.2)	(0.1)	(2.2)	(0.0)	(4.9)	(47.0)	(36.1)	(0.0)

Table 9. (Continued).

Sub	Cond	ContAng	Max	Tmax	Min	Tmin	ROM	ContVel	MaxVel	TmaxVel
15	1	-18.219	24.841	0.183	-18.219	0.000	43.060	503.677	584.921	0.024
		(3.0)	(1.8)	(0.1)	(3.0)	(0.0)	(3.7)	(76.3)	(42.4)	(0.0)
	2	-32.115	20.920	0.125	-32.115	0.000	53.035	557.001	687.885	0.028
		(3.7)	(1.2)	(0.0)	(3.7)	(0.0)	(3.8)	(34.1)	(46.1)	(0.0)
	3	-28.755	24.769	0.323	-28.755	0.000	53.524	466.843	582.049	0.027
		(1.8)	(3.0)	(0.0)	(1.8)	(0.0)	(2.6)	(27.3)	(19.6)	(0.0)
	4	-27.089	24.164	0.259	-27.089	0.000	51.252	476.235	585.002	0.027
		(1.2)	(1.8)	(0.0)	(1.2)	(0.0)	(1.1)	(30.3)	(25.9)	(0.0)
	5	-21.074	25.598	0.327	-21.074	0.000	46.672	512.595	591.661	0.023
		(4.8)	(1.5)	(0.3)	(4.8)	(0.0)	(4.3)	(28.9)	(18.2)	(0.0)
	6	-20.627	25.905	0.300	-20.627	0.000	46.532	540.014	591.609	0.018
		(3.3)	(1.9)	(0.1)	(3.3)	(0.0)	(1.9)	(25.2)	(34.0)	(0.0)

Note: Angle and ROM units are in degrees and time unit is in seconds.
Velocity unit is in deg/s.
Standard deviation values are in parentheses.
The definitions of variables are in Appendix A.

Table 10. Subject means and standard deviations of knee joint variables (Group 1).

Subj	Cond	ContAng	Max	Tmax	Min	Tmin	ContVel	MaxVel	TmaxVel
7	1	26.398	64.458	0.142	26.398	0.000	38.060	307.213	413.828
		(1.7)	(5.5)	(0.0)	(1.7)	(0.0)	(4.7)	(17.3)	(38.9)
	2	36.309	101.577	0.305	12.774	0.825	65.268	324.673	450.885
		(2.8)	(2.4)	(0.0)	(2.3)	(0.0)	(4.1)	(32.8)	(47.0)
	3	32.255	89.036	0.347	27.826	0.330	56.781	339.790	408.685
		(3.8)	(5.8)	(0.1)	(4.1)	(0.5)	(5.3)	(38.4)	(24.0)
	4	27.015	101.748	0.553	27.015	0.000	74.733	356.241	511.131
		(2.3)	(3.1)	(0.1)	(2.3)	(0.0)	(3.3)	(49.3)	(32.3)
	5	32.507	119.978	0.344	32.507	0.000	87.472	372.838	610.079
		(1.4)	(5.6)	(0.1)	(1.4)	(0.0)	(5.7)	(15.6)	(27.5)
	6	31.606	109.109	0.407	31.606	0.000	77.503	332.483	473.419
		(2.5)	(4.7)	(0.0)	(2.5)	(0.0)	(3.0)	(42.3)	(14.2)
8	1	21.801	55.558	0.124	14.805	0.775	33.757	239.422	383.021
		(2.6)	(5.4)	(0.0)	(2.5)	(0.1)	(3.2)	(25.4)	(22.5)
	2	24.857	54.017	0.118	9.102	0.762	29.161	266.772	338.342
		(1.4)	(3.7)	(0.0)	(3.0)	(0.1)	(4.7)	(29.6)	(38.1)
	3	21.522	74.045	0.208	20.913	0.330	52.524	290.562	468.883
		(2.3)	(7.8)	(0.0)	(2.3)	(0.5)	(6.5)	(32.5)	(23.8)
	4	27.941	81.830	0.157	10.090	0.823	53.889	318.411	535.378
		(1.8)	(4.3)	(0.0)	(2.4)	(0.0)	(5.1)	(32.3)	(46.5)
	5	37.258	122.254	0.450	37.258	0.000	84.995	369.213	475.020
		(1.2)	(5.1)	(0.1)	(1.2)	(0.0)	(5.3)	(29.7)	(24.8)
	6	35.497	101.978	0.327	22.505	0.825	66.481	362.645	492.711
		(2.0)	(4.4)	(0.0)	(3.6)	(0.0)	(3.9)	(22.6)	(49.5)
9	1	33.974	95.528	0.401	33.036	0.165	61.553	365.090	477.700
		(2.8)	(7.8)	(0.1)	(3.4)	(0.4)	(5.5)	(32.5)	(8.1)
	2	27.123	100.247	0.690	27.123	0.000	73.125	414.648	551.052
		(4.7)	(3.6)	(0.2)	(4.7)	(0.0)	(5.6)	(36.2)	(18.0)
	3	32.728	116.241	0.270	32.728	0.000	83.513	465.817	649.913
		(2.6)	(7.7)	(0.1)	(2.6)	(0.0)	(7.6)	(45.1)	(40.9)
	4	28.333	73.969	0.156	28.333	0.000	45.636	345.629	466.495
		(3.4)	(3.5)	(0.0)	(3.4)	(0.0)	(3.7)	(22.2)	(34.0)
	5	25.751	64.983	0.127	15.791	0.819	39.232	279.124	452.227
		(0.8)	(2.0)	(0.0)	(2.1)	(0.0)	(1.9)	(19.6)	(12.3)
	6	30.249	68.998	0.129	10.280	0.823	38.749	365.856	460.949
		(1.9)	(4.5)	(0.0)	(2.3)	(0.0)	(3.3)	(21.8)	(20.1)

Table 10. (Continued).

Subj	Cond	ContAng	Max	Tmax	Min	Tmin	ContVel	MaxVel	TmaxVel
13	1	25.248	76.565	0.169	20.853	0.825	51.318	331.194	507.998
		(1.1)	(5.7)	(0.0)	(4.4)	(0.0)	(5.4)	(36.0)	(40.0)
	2	29.363	86.867	0.149	13.719	0.825	57.504	366.868	622.427
		(2.7)	(3.8)	(0.0)	(3.8)	(0.0)	(5.8)	(33.0)	(55.5)
	3	32.749	85.149	0.183	31.738	0.165	52.401	330.100	499.388
		(0.7)	(3.7)	(0.0)	(2.4)	(0.4)	(3.5)	(39.4)	(11.9)
	4	25.842	69.485	0.132	15.913	0.727	43.644	307.164	493.268
		(1.1)	(1.8)	(0.0)	(2.3)	(0.1)	(1.4)	(27.8)	(17.5)
	5	33.273	73.358	0.130	12.260	0.732	40.085	378.498	474.598
		(1.7)	(8.4)	(0.0)	(2.0)	(0.2)	(8.7)	(36.0)	(60.7)
	6	24.834	81.728	0.183	9.621	0.825	56.895	372.521	540.854
		(2.2)	(5.4)	(0.0)	(1.2)	(0.0)	(7.1)	(27.2)	(31.5)
16	1	28.339	93.588	0.164	15.623	0.825	65.249	413.440	686.051
		(1.7)	(3.5)	(0.0)	(6.6)	(0.0)	(3.3)	(29.2)	(53.2)
	2	34.978	122.320	0.404	34.978	0.000	87.342	348.958	545.192
		(2.4)	(5.1)	(0.0)	(2.4)	(0.0)	(5.0)	(25.3)	(13.1)
	3	35.825	105.857	0.303	13.275	0.825	70.032	365.252	525.846
		(3.5)	(4.3)	(0.0)	(5.1)	(0.0)	(6.7)	(22.6)	(22.3)
	4	34.777	111.066	0.413	34.777	0.000	76.289	392.265	575.344
		(1.6)	(6.8)	(0.0)	(1.6)	(0.0)	(6.9)	(21.9)	(37.8)
	5	24.677	97.950	0.427	24.677	0.000	73.274	384.106	564.592
		(2.8)	(2.2)	(0.1)	(2.8)	(0.0)	(1.4)	(35.4)	(37.7)
	6	34.016	120.845	0.261	27.997	0.660	86.829	460.767	687.347
		(2.1)	(7.9)	(0.0)	(5.6)	(0.4)	(6.5)	(20.9)	(32.2)

Note: Angle and ROM units are in degrees and time unit is in seconds.
Velocity unit is in deg/s.
Standard deviation values are in parentheses.
The definitions of variables are in Appendix A.

Table 11. Subject means and standard deviations of knee joint variables (Group 2).

Subj	Cond	ContAng	Max	Tmax	Min	Tmin	ROM	ContVel	MaxVel	TmaxVel
4	1	23.839	90.692	0.322	23.305	0.165	66.853	268.978	519.450	0.063
		(4.5)	(8.1)	(0.1)	(5.2)	(0.4)	(7.4)	(43.1)	(44.6)	(0.0)
	2	20.176	77.043	0.219	6.935	0.827	56.866	192.093	495.615	0.080
		(4.5)	(1.3)	(0.0)	(5.5)	(0.0)	(3.3)	(97.6)	(23.9)	(0.0)
	3	25.643	100.160	0.383	25.643	0.000	74.516	256.328	585.440	0.064
		(2.6)	(4.0)	(0.1)	(2.6)	(0.0)	(3.6)	(26.2)	(31.5)	(0.0)
	4	18.831	81.030	0.203	16.036	0.330	62.200	275.010	543.960	0.065
		(1.3)	(5.0)	(0.0)	(3.2)	(0.5)	(3.8)	(38.7)	(32.6)	(0.0)
	5	22.119	107.076	0.471	22.119	0.000	84.957	345.238	616.374	0.053
		(2.2)	(8.9)	(0.2)	(2.2)	(0.0)	(7.5)	(19.5)	(54.5)	(0.0)
	6	17.782	82.745	0.252	17.782	0.000	64.963	310.622	599.534	0.058
		(3.1)	(4.3)	(0.1)	(3.1)	(0.0)	(5.1)	(38.7)	(42.3)	(0.0)
11	1	41.550	112.092	0.393	40.397	0.165	70.543	490.386	556.590	0.023
		(1.9)	(5.5)	(0.0)	(2.5)	(0.4)	(5.9)	(36.3)	(27.8)	(0.0)
	2	26.072	57.031	0.105	24.250	0.172	30.959	367.501	452.017	0.028
		(4.6)	(6.5)	(0.0)	(6.7)	(0.3)	(3.4)	(48.9)	(50.2)	(0.0)
	3	40.783	117.125	0.438	40.783	0.000	76.342	474.943	564.748	0.027
		(2.8)	(4.3)	(0.1)	(2.8)	(0.0)	(3.9)	(42.9)	(42.6)	(0.0)
	4	28.613	67.506	0.120	27.156	0.070	38.893	408.592	511.050	0.030
		(1.4)	(6.8)	(0.0)	(4.5)	(0.2)	(6.1)	(32.0)	(45.0)	(0.0)
	5	28.356	72.723	0.128	28.356	0.000	44.368	416.286	554.263	0.033
		(1.9)	(3.8)	(0.0)	(1.9)	(0.0)	(3.2)	(33.3)	(39.3)	(0.0)
	6	38.297	118.666	0.375	38.297	0.000	80.369	535.584	648.937	0.029
		(0.9)	(5.2)	(0.1)	(0.9)	(0.0)	(5.3)	(56.3)	(56.2)	(0.0)
12	1	24.051	112.595	0.405	24.051	0.000	88.545	439.009	603.909	0.050
		(1.4)	(6.6)	(0.1)	(1.4)	(0.0)	(7.0)	(30.7)	(11.1)	(0.0)
	2	25.273	90.871	0.197	9.978	0.805	65.598	408.868	543.537	0.048
		(2.5)	(3.2)	(0.0)	(1.8)	(0.1)	(4.7)	(17.2)	(22.5)	(0.0)
	3	27.656	115.301	0.325	26.888	0.165	87.645	493.434	637.167	0.046
		(2.1)	(5.7)	(0.0)	(1.7)	(0.4)	(5.2)	(29.0)	(39.4)	(0.0)
	4	27.466	90.840	0.178	9.165	0.777	63.373	447.650	570.629	0.045
		(2.1)	(3.1)	(0.0)	(2.1)	(0.1)	(2.2)	(21.8)	(7.8)	(0.0)
	5	25.226	86.705	0.159	12.719	0.539	61.479	479.343	602.584	0.038
		(2.1)	(1.6)	(0.0)	(5.4)	(0.3)	(3.2)	(19.1)	(16.1)	(0.0)
	6	28.346	111.959	0.403	28.346	0.000	83.613	512.615	660.066	0.046
		(2.7)	(5.4)	(0.1)	(2.7)	(0.0)	(3.1)	(12.3)	(32.4)	(0.0)

Table 11. (Continued).

Subj	Cond	ContAng	Max	Tmax	Min	Tmin	ROM	ContVel	MaxVel	TmaxVel
14	1	26.292	106.328	0.272	26.292	0.000	80.035	480.003	641.147	0.045
		(2.1)	(2.6)	(0.0)	(2.1)	(0.0)	(2.9)	(19.4)	(22.0)	(0.0)
	2	19.006	70.050	0.141	19.006	0.000	51.044	402.754	537.457	0.044
		(3.1)	(5.2)	(0.0)	(3.1)	(0.0)	(2.7)	(24.3)	(30.3)	(0.0)
	3	23.835	106.807	0.258	23.835	0.000	82.972	521.023	693.056	0.043
		(2.5)	(1.8)	(0.0)	(2.5)	(0.0)	(2.0)	(31.0)	(30.7)	(0.0)
	4	19.421	75.260	0.144	19.421	0.000	55.839	439.739	586.061	0.044
		(4.2)	(10.4)	(0.0)	(4.2)	(0.0)	(6.5)	(23.4)	(23.2)	(0.0)
	5	21.759	85.096	0.153	21.016	0.165	63.337	490.027	655.329	0.042
		(1.7)	(5.9)	(0.0)	(2.8)	(0.4)	(6.2)	(19.7)	(26.7)	(0.0)
	6	28.482	107.282	0.246	28.482	0.000	78.800	558.430	701.659	0.039
		(1.1)	(3.8)	(0.0)	(1.1)	(0.0)	(4.2)	(44.6)	(38.7)	(0.0)

Note: Angle and ROM units are in degrees and time unit is in seconds.
Velocity unit is in deg/s.
Standard deviation values are in parentheses.
The definitions of variables are in Appendix A.

Table 12. Subject means and standard deviations of knee joint variables (Group 3).

Subj	Cond	ContAng	Max	Tmax	Min	Tmin	ROM	ContVel	MaxVel	TmaxVel
5	1	36.590	119.426	0.318	36.590	0.000	82.837	375.848	529.901	0.077
		(3.1)	(3.8)	(0.0)	(3.1)	(0.0)	(6.4)	(28.6)	(31.7)	(0.0)
	2	21.919	82.528	0.176	15.766	0.653	60.609	325.998	494.344	0.073
		(4.2)	(4.5)	(0.0)	(3.9)	(0.4)	(8.2)	(32.4)	(52.8)	(0.0)
	3	31.592	118.480	0.283	30.249	0.330	86.887	416.376	614.154	0.068
		(5.2)	(6.5)	(0.0)	(5.2)	(0.5)	(3.3)	(23.4)	(39.2)	(0.0)
	4	25.761	93.765	0.182	19.290	0.660	68.003	411.563	564.639	0.059
		(2.8)	(2.5)	(0.0)	(3.9)	(0.4)	(3.8)	(20.4)	(35.8)	(0.0)
	5	27.725	97.602	0.178	16.109	0.825	69.877	370.876	621.377	0.067
		(1.9)	(5.0)	(0.0)	(5.2)	(0.0)	(5.7)	(24.6)	(29.9)	(0.0)
	6	33.865	118.503	0.275	31.890	0.330	84.638	455.326	628.724	0.064
		(4.1)	(4.9)	(0.0)	(4.8)	(0.5)	(2.2)	(29.9)	(17.6)	(0.0)
6	1	28.044	105.790	0.335	26.808	0.165	77.746	332.917	529.962	0.069
		(3.0)	(3.0)	(0.0)	(2.3)	(0.4)	(4.1)	(20.9)	(33.6)	(0.0)
	2	22.305	62.805	0.144	9.127	0.795	40.501	305.517	379.027	0.047
		(2.4)	(6.9)	(0.0)	(5.9)	(0.1)	(4.8)	(22.6)	(22.2)	(0.0)
	3	21.650	98.696	0.327	14.233	0.495	77.046	243.675	541.599	0.093
		(8.8)	(4.5)	(0.1)	(5.2)	(0.5)	(11.9)	(177.3)	(15.0)	(0.0)
	4	23.950	68.984	0.137	7.111	0.785	45.034	347.566	474.380	0.049
		(1.6)	(5.3)	(0.0)	(9.4)	(0.1)	(3.9)	(31.4)	(29.5)	(0.0)
	5	31.785	85.119	0.162	3.767	0.802	53.335	387.555	511.588	0.046
		(2.6)	(10.4)	(0.0)	(7.0)	(0.0)	(10.8)	(43.8)	(48.0)	(0.0)
	6	29.996	106.642	0.265	4.920	0.825	76.646	442.949	601.785	0.050
		(2.9)	(3.1)	(0.0)	(5.3)	(0.0)	(4.7)	(17.9)	(37.6)	(0.0)
10	1	31.207	117.542	0.641	31.207	0.000	86.336	428.698	525.387	0.043
		(3.1)	(6.1)	(0.1)	(3.1)	(0.0)	(4.2)	(18.8)	(38.1)	(0.0)
	2	28.486	77.521	0.163	6.211	0.801	49.035	362.591	456.185	0.042
		(2.7)	(8.2)	(0.0)	(8.8)	(0.0)	(5.8)	(23.7)	(22.9)	(0.0)
	3	34.828	111.453	0.462	34.828	0.000	76.625	438.126	542.721	0.045
		(2.0)	(7.4)	(0.1)	(2.0)	(0.0)	(6.1)	(22.3)	(40.4)	(0.0)
	4	30.769	88.489	0.173	14.607	0.660	57.720	386.556	534.306	0.048
		(3.0)	(4.3)	(0.0)	(9.8)	(0.4)	(5.6)	(16.6)	(59.4)	(0.0)
	5	34.170	92.341	0.174	4.560	0.825	58.172	405.926	537.631	0.048
		(2.4)	(8.2)	(0.0)	(4.2)	(0.0)	(6.5)	(26.0)	(33.7)	(0.0)
	6	35.290	114.131	0.546	35.290	0.000	78.840	478.239	594.578	0.044
		(3.2)	(3.7)	(0.3)	(3.2)	(0.0)	(5.0)	(19.0)	(33.3)	(0.0)

Table 12. (Continued).

Subj	Cond	ContAng	Max	Tmax	Min	Tmin	ROM	ContVel	MaxVel	TmaxVel
15	1	23.794	90.649	0.288	23.794	0.000	66.855	412.240	506.618	0.048
		(2.5)	(4.1)	(0.1)	(2.5)	(0.0)	(3.5)	(19.3)	(22.5)	(0.0)
	2	20.082	68.060	0.160	20.082	0.000	47.978	349.533	426.235	0.046
		(2.3)	(4.5)	(0.0)	(2.3)	(0.0)	(3.1)	(18.8)	(24.6)	(0.0)
	3	30.588	105.605	0.307	30.588	0.000	75.018	455.184	555.610	0.052
		(2.9)	(4.0)	(0.1)	(2.9)	(0.0)	(2.8)	(30.4)	(36.8)	(0.0)
	4	22.987	75.162	0.166	22.606	0.165	52.175	410.282	480.938	0.034
		(2.3)	(3.8)	(0.0)	(2.1)	(0.4)	(4.2)	(16.5)	(19.8)	(0.0)
	5	27.415	71.678	0.134	8.058	0.743	44.263	436.949	482.993	0.023
		(1.0)	(4.7)	(0.0)	(5.6)	(0.2)	(4.3)	(26.5)	(30.0)	(0.0)
	6	30.179	105.803	0.279	23.394	0.483	75.624	457.701	587.889	0.054
		(2.9)	(4.9)	(0.1)	(12.1)	(0.4)	(5.0)	(35.1)	(36.7)	(0.0)

Note: Angle and ROM units are in degrees and time unit is in s.
Velocity unit is in deg/s.
Standard deviation values are in parentheses.
The definitions of variables are in Appendix A.

Table 13. Subject means and standard deviations of hip joint variables (Group 1).

Subj	Cond	ContAng	Max	Tmax	Min	Tmin	ROM	ContVel	MaxVel	Tmaxvel
7	1	11.825	27.343	0.150	11.825	0.000	15.518	123.189	179.721	0.041
		(2.0)	(2.5)	(0.0)	(2.0)	(0.0)	(1.7)	(24.7)	(26.3)	(0.0)
	2	22.239	77.100	0.411	22.239	0.000	54.860	149.973	252.746	0.062
		(1.8)	(1.8)	(0.0)	(1.8)	(0.0)	(2.2)	(21.3)	(10.3)	(0.0)
	3	24.830	82.573	0.390	24.830	0.000	57.743	188.400	303.616	0.061
		(2.3)	(2.4)	(0.0)	(2.3)	(0.0)	(2.6)	(11.7)	(13.1)	(0.0)
	4	15.932	34.728	0.170	15.932	0.000	18.796	142.233	209.636	0.044
		(3.7)	(4.8)	(0.1)	(3.7)	(0.0)	(2.8)	(29.2)	(32.5)	(0.0)
	5	17.410	41.493	0.220	17.410	0.000	24.083	139.912	219.988	0.049
		(2.4)	(1.3)	(0.0)	(2.4)	(0.0)	(3.4)	(25.6)	(14.2)	(0.0)
	6	20.314	83.117	0.397	20.314	0.000	62.803	182.560	307.269	0.062
		(2.2)	(4.9)	(0.0)	(2.2)	(0.0)	(3.6)	(23.6)	(17.3)	(0.0)
8	1	24.234	90.145	0.309	18.217	0.825	65.912	190.629	391.709	0.087
		(2.8)	(2.6)	(0.0)	(2.9)	(0.0)	(4.9)	(23.9)	(22.5)	(0.0)
	2	16.159	30.312	0.212	12.186	0.653	14.153	94.461	156.826	0.050
		(2.1)	(5.6)	(0.2)	(1.3)	(0.4)	(4.7)	(33.4)	(35.7)	(0.0)
	3	28.393	101.336	0.338	25.165	0.660	72.943	224.840	379.851	0.065
		(2.3)	(3.4)	(0.0)	(3.3)	(0.4)	(5.1)	(13.5)	(25.3)	(0.0)
	4	17.672	31.823	0.116	11.802	0.819	14.151	109.153	180.335	0.048
		(1.8)	(2.7)	(0.0)	(2.5)	(0.0)	(1.0)	(7.1)	(8.2)	(0.0)
	5	18.964	35.812	0.143	12.276	0.810	16.847	127.866	206.943	0.047
		(1.7)	(2.2)	(0.0)	(1.0)	(0.0)	(1.4)	(12.0)	(9.2)	(0.0)
	6	31.298	98.299	0.314	15.321	0.825	67.002	231.679	389.359	0.060
		(2.1)	(3.1)	(0.0)	(1.8)	(0.0)	(3.3)	(21.1)	(23.1)	(0.0)
9	1	24.203	81.216	0.358	23.594	0.165	57.012	190.997	288.594	0.070
		(4.7)	(10.3)	(0.0)	(3.5)	(0.4)	(8.7)	(45.7)	(46.4)	(0.0)
	2	17.093	32.911	0.122	11.744	0.758	15.818	141.815	189.546	0.042
		(0.7)	(1.2)	(0.0)	(3.3)	(0.1)	(1.0)	(36.9)	(18.0)	(0.0)
	3	27.648	93.032	0.428	26.839	0.165	65.384	217.780	318.071	0.052
		(2.8)	(11.2)	(0.1)	(3.7)	(0.4)	(10.1)	(31.9)	(30.3)	(0.0)
	4	23.183	50.927	0.182	12.758	0.807	27.744	199.747	278.543	0.043
		(1.3)	(5.1)	(0.0)	(2.6)	(0.0)	(4.3)	(13.4)	(17.8)	(0.0)
	5	23.344	51.579	0.173	11.534	0.825	28.235	222.201	297.426	0.038
		(2.9)	(9.1)	(0.0)	(6.3)	(0.0)	(7.8)	(30.2)	(48.5)	(0.0)
	6	26.405	104.737	0.438	26.405	0.000	78.332	246.027	366.494	0.049
		(1.9)	(2.0)	(0.0)	(1.9)	(0.0)	(2.8)	(16.3)	(23.3)	(0.0)

Table 13. (Continued).

Subj	Cond	ContAng	Max	Tmax	Min	Tmin	ROM	ContVel	MaxVel	Tmaxvel
13	1	28.980	109.948	0.713	28.980	0.000	80.968	257.421	401.974	0.056
		(5.6)	(3.1)	(0.1)	(5.6)	(0.0)	(6.8)	(38.4)	(32.0)	(0.0)
	2	23.328	58.705	0.223	21.228	0.493	35.377	179.076	322.289	0.054
		(3.3)	(7.4)	(0.1)	(5.0)	(0.5)	(6.1)	(20.1)	(31.2)	(0.0)
	3	34.575	114.416	0.729	34.575	0.000	79.841	312.294	423.271	0.041
		(4.3)	(4.1)	(0.2)	(4.3)	(0.0)	(5.6)	(55.0)	(31.2)	(0.0)
	4	25.691	56.739	0.186	18.895	0.825	31.048	201.319	317.103	0.047
		(1.3)	(2.5)	(0.0)	(3.2)	(0.0)	(3.3)	(23.8)	(27.4)	(0.0)
	5	27.285	65.251	0.191	11.879	0.825	37.966	259.530	369.343	0.044
		(1.8)	(4.1)	(0.0)	(1.7)	(0.0)	(5.0)	(12.2)	(27.0)	(0.0)
	6	33.477	111.358	0.514	33.477	0.000	77.881	280.768	437.939	0.053
		(3.1)	(2.3)	(0.2)	(3.1)	(0.0)	(2.0)	(30.0)	(49.6)	(0.0)
16	1	27.139	108.863	0.401	27.139	0.000	81.724	237.418	444.013	0.070
		(4.1)	(6.4)	(0.0)	(4.1)	(0.0)	(4.9)	(17.3)	(29.3)	(0.0)
	2	20.677	50.027	0.185	8.652	0.825	29.350	177.241	286.944	0.051
		(1.6)	(4.9)	(0.0)	(2.4)	(0.0)	(4.5)	(16.2)	(26.5)	(0.0)
	3	29.626	118.567	0.372	29.626	0.000	88.942	276.073	467.789	0.069
		(4.6)	(8.2)	(0.1)	(4.6)	(0.0)	(6.7)	(22.2)	(14.5)	(0.0)
	4	25.296	63.479	0.222	14.876	0.825	38.184	229.114	369.323	0.045
		(0.6)	(3.1)	(0.0)	(2.9)	(0.0)	(3.1)	(29.7)	(26.1)	(0.0)
	5	24.231	67.324	0.236	13.693	0.825	43.093	249.390	392.686	0.044
		(1.9)	(7.8)	(0.0)	(4.1)	(0.0)	(6.4)	(17.1)	(32.3)	(0.0)
	6	32.407	123.981	0.349	31.475	0.165	91.574	304.270	497.270	0.059
		(2.4)	(4.5)	(0.0)	(0.8)	(0.4)	(5.0)	(22.2)	(20.7)	(0.0)

Note: Angle and ROM units are in degrees and time unit is in seconds.
Velocity unit is in deg/s.
Standard deviation values are in parentheses.
The definitions of variables are in Appendix A.

Table 14. Subject means and standard deviations of hip joint variables (Group 2).

Subj	Cond	ContAng	Max	Tmax	Min	Tmin	ROM	ContVel	MaxVel	Tmaxvel
4	1	17.949	90.342	0.361	17.949	0.000	72.393	185.675	416.112	0.072
		(5.1)	(12.7)	(0.1)	(5.1)	(0.0)	(11.8)	(42.7)	(10.2)	(0.0)
	2	12.229	52.408	0.232	8.715	0.677	40.179	126.143	335.547	0.082
		(3.0)	(9.9)	(0.1)	(2.6)	(0.4)	(9.6)	(57.4)	(37.3)	(0.0)
	3	18.180	108.470	0.438	18.180	0.000	90.290	181.693	458.762	0.069
		(2.1)	(3.8)	(0.0)	(2.1)	(0.0)	(3.4)	(13.6)	(23.0)	(0.0)
	4	13.145	55.973	0.286	13.145	0.000	42.828	166.004	357.860	0.069
		(1.6)	(6.8)	(0.1)	(1.6)	(0.0)	(6.1)	(16.2)	(17.7)	(0.0)
	5	15.002	111.037	0.508	15.002	0.000	96.036	235.078	474.725	0.081
		(2.9)	(3.8)	(0.1)	(2.9)	(0.0)	(5.1)	(14.7)	(54.1)	(0.0)
	6	9.662	59.935	0.369	9.662	0.000	50.273	183.356	365.517	0.059
		(1.6)	(7.4)	(0.1)	(1.6)	(0.0)	(6.5)	(23.2)	(19.1)	(0.0)
11	1	50.488	140.025	0.356	50.488	0.000	89.538	439.026	510.963	0.029
		(4.8)	(4.7)	(0.0)	(4.8)	(0.0)	(8.7)	(25.6)	(43.1)	(0.0)
	2	28.326	45.541	0.103	24.929	0.601	17.215	220.142	263.404	0.025
		(4.1)	(8.3)	(0.0)	(6.6)	(0.4)	(4.8)	(44.9)	(44.5)	(0.0)
	3	47.908	143.045	0.434	47.908	0.000	95.136	425.628	527.419	0.035
		(5.1)	(1.2)	(0.1)	(5.1)	(0.0)	(4.5)	(28.1)	(19.4)	(0.0)
	4	31.451	55.715	0.159	31.451	0.000	24.264	245.784	295.123	0.027
		(2.3)	(7.0)	(0.0)	(2.3)	(0.0)	(6.3)	(17.9)	(36.5)	(0.0)
	5	33.759	63.705	0.183	33.759	0.000	29.946	270.446	336.452	0.033
		(3.6)	(7.1)	(0.0)	(3.6)	(0.0)	(4.6)	(27.7)	(40.0)	(0.0)
	6	47.877	145.090	0.391	47.877	0.000	97.213	438.136	553.125	0.033
		(6.0)	(2.4)	(0.0)	(6.0)	(0.0)	(5.5)	(45.8)	(30.8)	(0.0)
12	1	21.240	102.493	0.410	21.240	0.000	81.253	244.314	441.987	0.074
		(2.1)	(7.1)	(0.0)	(2.1)	(0.0)	(7.3)	(26.3)	(15.0)	(0.0)
	2	21.685	63.636	0.194	10.464	0.828	41.952	229.911	354.987	0.061
		(1.8)	(4.0)	(0.0)	(3.2)	(0.0)	(3.9)	(20.4)	(20.2)	(0.0)
	3	23.953	98.200	0.357	23.953	0.000	74.247	263.537	448.211	0.068
		(2.2)	(7.2)	(0.0)	(2.2)	(0.0)	(6.4)	(14.4)	(50.9)	(0.0)
	4	22.312	66.223	0.193	8.968	0.812	43.910	256.818	391.229	0.057
		(0.8)	(3.1)	(0.0)	(2.8)	(0.0)	(3.3)	(17.4)	(20.9)	(0.0)
	5	23.397	63.929	0.181	13.999	0.660	40.532	272.811	397.521	0.049
		(4.0)	(1.5)	(0.0)	(3.2)	(0.4)	(2.7)	(41.8)	(4.4)	(0.0)
	6	26.153	114.731	0.431	26.153	0.000	88.578	308.624	483.171	0.060
		(3.7)	(6.5)	(0.0)	(3.7)	(0.0)	(7.6)	(14.7)	(29.3)	(0.0)

Table 14. (Continued).

Subj	Cond	ContAng	Max	Tmax	Min	Tmin	ROM	ContVel	MaxVel	Tmaxvel
14	1	13.794	70.156	0.353	13.794	0.000	56.362	255.596	359.467	0.050
		(1.3)	(2.2)	(0.0)	(1.3)	(0.0)	(3.2)	(8.0)	(14.4)	(0.0)
	2	7.477	30.292	0.146	7.477	0.000	22.815	173.359	237.294	0.045
		(0.9)	(2.0)	(0.0)	(0.9)	(0.0)	(1.3)	(11.9)	(15.5)	(0.0)
	3	13.381	79.532	0.391	13.381	0.000	66.150	277.866	403.138	0.048
		(3.0)	(2.7)	(0.0)	(3.0)	(0.0)	(2.9)	(19.8)	(19.8)	(0.0)
	4	8.891	33.605	0.161	8.579	0.165	24.714	184.373	253.980	0.048
		(2.2)	(4.6)	(0.0)	(1.9)	(0.4)	(2.5)	(18.2)	(15.2)	(0.0)
	5	12.314	42.844	0.167	10.593	0.330	30.531	237.952	313.451	0.042
		(1.0)	(2.1)	(0.0)	(3.3)	(0.5)	(2.1)	(13.1)	(15.3)	(0.0)
	6	16.621	77.024	0.373	16.621	0.000	60.404	302.761	407.885	0.042
		(3.4)	(2.4)	(0.0)	(3.4)	(0.0)	(4.9)	(22.8)	(31.5)	(0.0)

Note: Angle and ROM units are in degrees and time unit is in seconds.
Velocity unit is in deg/s.
Standard deviation values are in parentheses.
The definitions of variables are in Appendix A.

Table 15. Subject means and standard deviations of hip joint variables (Group 3).

Subj	Cond	ContAng	Max	Tmax	Min	Tmin	ROM	ContVel	MaxVel	Tmaxvel
5	1	32.791	101.641	0.377	32.532	0.165	68.850	192.502	379.607	0.090
		(1.8)	(5.3)	(0.0)	(1.4)	(0.4)	(5.5)	(13.3)	(19.3)	(0.0)
	2	13.249	41.393	0.218	11.386	0.330	28.144	111.731	242.459	0.083
		(6.4)	(1.8)	(0.1)	(5.7)	(0.5)	(5.4)	(32.3)	(32.3)	(0.0)
	3	24.536	95.798	0.342	24.536	0.000	71.262	195.553	435.229	0.083
		(2.5)	(5.8)	(0.0)	(2.5)	(0.0)	(5.0)	(22.8)	(29.1)	(0.0)
	4	17.993	56.929	0.234	12.566	0.330	38.935	172.372	310.556	0.072
		(5.6)	(5.9)	(0.0)	(6.2)	(0.5)	(3.9)	(21.5)	(21.1)	(0.0)
	5	20.511	56.301	0.190	16.166	0.637	35.791	169.215	335.006	0.068
		(0.7)	(6.5)	(0.0)	(2.2)	(0.4)	(6.8)	(19.1)	(37.2)	(0.0)
	6	27.325	96.514	0.337	27.325	0.000	69.189	201.081	432.956	0.076
		(3.2)	(4.4)	(0.0)	(3.2)	(0.0)	(1.9)	(19.2)	(5.6)	(0.0)
6	1	28.027	86.016	0.365	27.171	0.165	57.989	187.229	342.713	0.077
		(4.0)	(6.0)	(0.0)	(2.9)	(0.4)	(9.5)	(14.7)	(39.1)	(0.0)
	2	17.431	34.226	0.152	5.991	0.825	16.794	96.950	159.183	0.074
		(1.5)	(3.0)	(0.0)	(4.7)	(0.0)	(2.0)	(12.0)	(19.1)	(0.0)
	3	22.494	74.135	0.322	16.583	0.495	51.641	133.597	339.835	0.103
		(6.4)	(3.5)	(0.1)	(3.5)	(0.5)	(4.3)	(79.5)	(12.8)	(0.0)
	4	19.918	44.969	0.164	6.962	0.660	25.051	141.375	254.260	0.058
		(2.7)	(8.6)	(0.0)	(7.5)	(0.4)	(6.4)	(38.2)	(52.8)	(0.0)
	5	24.402	57.096	0.195	8.230	0.825	32.694	199.136	291.173	0.052
		(2.8)	(12.6)	(0.1)	(4.0)	(0.0)	(12.0)	(24.4)	(50.3)	(0.0)
	6	26.646	80.333	0.288	10.104	0.825	53.686	237.177	377.300	0.063
		(1.5)	(2.1)	(0.0)	(5.2)	(0.0)	(3.3)	(15.5)	(19.6)	(0.0)
10	1	25.162	122.858	0.623	-18.722	0.118	97.696	273.646	368.742	0.161
		(3.6)	(24.2)	(0.1)	(97.5)	(0.3)	(23.5)	(15.3)	(12.4)	(0.2)
	2	14.986	44.174	0.190	5.138	0.753	29.188	177.168	242.928	0.049
		(3.3)	(13.2)	(0.1)	(8.3)	(0.1)	(10.1)	(39.3)	(43.1)	(0.0)
	3	26.311	105.920	0.440	26.311	0.000	79.609	292.760	425.383	0.064
		(2.8)	(8.3)	(0.1)	(2.8)	(0.0)	(8.0)	(12.3)	(23.8)	(0.0)
	4	18.735	60.283	0.237	9.133	0.625	41.548	222.104	349.518	0.056
		(3.1)	(5.6)	(0.1)	(7.2)	(0.4)	(6.2)	(23.2)	(37.7)	(0.0)
	5	25.604	64.823	0.211	3.939	0.773	39.219	267.271	348.384	0.046
		(1.9)	(5.6)	(0.0)	(5.2)	(0.1)	(4.3)	(8.1)	(23.5)	(0.0)
	6	31.325	114.439	0.426	31.325	0.000	83.114	339.824	450.179	0.049
		(4.2)	(4.8)	(0.1)	(4.2)	(0.0)	(3.7)	(24.1)	(32.2)	(0.0)

Table 15. (Continued).

Subj	Cond	ContAng	Max	Tmax	Min	Tmin	ROM	ContVel	MaxVel	Tmaxvel
15	1	15.272	73.717	0.358	15.272	0.000	58.445	245.413	356.837	0.061
		(3.7)	(4.4)	(0.0)	(3.7)	(0.0)	(5.8)	(21.3)	(21.3)	(0.0)
	2	9.474	38.052	0.183	8.775	0.165	28.578	178.594	254.479	0.054
		(2.3)	(6.1)	(0.1)	(3.5)	(0.4)	(4.2)	(14.9)	(20.7)	(0.0)
	3	23.340	92.333	0.357	23.340	0.000	68.993	297.961	408.108	0.062
		(3.6)	(4.8)	(0.1)	(3.6)	(0.0)	(5.3)	(21.6)	(24.7)	(0.0)
	4	13.346	48.707	0.209	13.346	0.000	35.361	211.581	294.294	0.051
		(2.2)	(3.8)	(0.1)	(2.2)	(0.0)	(2.3)	(11.5)	(11.6)	(0.0)
	5	16.460	46.489	0.309	-1.045	0.694	30.028	200.374	232.445	0.152
		(1.1)	(10.0)	(0.3)	(28.1)	(0.2)	(9.4)	(18.1)	(22.7)	(0.3)
	6	20.942	92.907	0.338	14.700	0.165	71.965	280.281	416.925	0.063
		(2.9)	(5.7)	(0.0)	(15.2)	(0.4)	(5.4)	(25.9)	(24.9)	(0.0)

Note: Angle and ROM units are in degrees and time unit is in seconds.
Velocity unit is in deg/s.
Standard deviation values are in parentheses.
The definitions of variables are in Appendix A.

APPENDIX D

VERTICAL GROUND REACTION FORCE DATA

Table 16. Subject means and standard deviations of VGRF variables (Group 1).

Subj	Cond	F1	T1	F2	T2	LRF1	LRF2	Imp 100ms
7	1	19.276	0.011	38.220	0.058	1740.190	715.080	2.257
		(0.5)	(0.0)	(11.1)	(0.0)	(154.5)	(407.6)	(0.1)
	2	24.602	0.010	63.395	0.046	2530.498	1515.850	3.241
		(3.6)	(0.0)	(7.3)	(0.0)	(393.2)	(397.0)	(0.1)
	3	28.938	0.013	30.257	0.062	2289.456	421.165	2.368
		(1.8)	(0.0)	(3.4)	(0.0)	(96.0)	(138.0)	(0.2)
	4	35.104	0.011	63.527	0.048	3311.855	1496.597	3.377
		(3.6)	(0.0)	(8.1)	(0.0)	(371.3)	(406.2)	(0.2)
	5	51.772	0.011	71.737	0.047	4572.610	2028.878	3.414
		(4.4)	(0.0)	(14.9)	(0.0)	(401.7)	(873.9)	(0.1)
	6	42.622	0.011	59.636	0.047	3885.971	1623.687	2.928
		(2.9)	(0.0)	(7.3)	(0.0)	(339.7)	(351.0)	(0.2)
8	1	12.524	0.014	32.410	0.074	882.324	455.437	1.917
		(1.8)	(0.0)	(6.6)	(0.0)	(97.1)	(138.0)	(0.1)
	2	21.836	0.015	64.132	0.062	1471.383	1217.338	3.311
		(1.5)	(0.0)	(8.8)	(0.0)	(80.1)	(233.5)	(0.2)
	3	18.846	0.014	45.167	0.058	1403.531	878.866	2.423
		(1.9)	(0.0)	(5.2)	(0.0)	(94.4)	(163.3)	(0.2)
	4	33.621	0.010	46.328	0.048	3430.977	858.607	2.735
		(1.7)	(0.0)	(3.8)	(0.0)	(177.4)	(141.4)	(0.1)
	5	38.946	0.015	89.216	0.052	2656.642	2260.770	4.228
		(3.7)	(0.0)	(14.5)	(0.0)	(251.8)	(825.6)	(0.1)
	6	27.047	0.015	53.104	0.055	1824.984	1215.608	2.825
		(3.4)	(0.0)	(11.5)	(0.0)	(240.5)	(437.3)	(0.2)
9	1	23.287	0.013	39.776	0.066	1813.967	521.612	2.593
		(2.7)	(0.0)	(10.1)	(0.0)	(347.7)	(254.3)	(0.4)
	2	31.297	0.011	63.037	0.053	2748.156	1315.686	3.395
		(1.2)	(0.0)	(12.1)	(0.0)	(292.3)	(733.3)	(0.3)
	3	34.224	0.010	46.147	0.048	3608.708	890.724	2.612
		(4.7)	(0.0)	(8.0)	(0.0)	(421.6)	(345.1)	(0.2)
	4	37.000	0.009	72.602	0.041	4119.916	1953.880	3.219
		(2.9)	(0.0)	(5.9)	(0.0)	(403.4)	(434.9)	(0.3)
	5	50.595	0.010	78.940	0.040	5393.501	2165.747	3.525
		(5.0)	(0.0)	(10.3)	(0.0)	(976.0)	(578.9)	(0.3)
	6	45.495	0.009	67.435	0.041	5102.064	1966.051	2.854
		(5.1)	(0.0)	(8.0)	(0.0)	(806.4)	(518.5)	(0.1)

Table 16. (Continued).

Subj	Cond	F1	T1	F2	T2	LRF1	LRF2	Imp 100ms
13	1	19.248	0.008	43.525	0.042	2501.691	1020.467	2.113
		(1.4)	(0.0)	(5.8)	(0.0)	(214.7)	(309.2)	(0.0)
	2	25.358	0.010	46.719	0.050	2537.926	900.140	2.777
		(1.9)	(0.0)	(3.4)	(0.0)	(164.6)	(200.3)	(0.3)
	3	24.370	0.008	77.379	0.026	2944.809	4371.194	2.489
		(1.1)	(0.0)	(23.3)	(0.0)	(287.9)	(3088.7)	(0.2)
	4	29.202	0.010	57.036	0.041	3024.640	1340.928	3.151
		(1.0)	(0.0)	(5.2)	(0.0)	(133.1)	(364.2)	(0.2)
	5	35.748	0.010	74.239	0.038	3709.197	2275.975	3.376
		(2.0)	(0.0)	(7.2)	(0.0)	(262.5)	(421.4)	(0.2)
16	1	18.571	0.010	34.778	0.059	1826.474	619.097	2.108
		(1.9)	(0.0)	(4.9)	(0.0)	(363.8)	(183.9)	(0.1)
	2	22.017	0.009	46.270	0.049	2364.711	981.043	2.676
		(1.3)	(0.0)	(5.0)	(0.0)	(153.5)	(173.2)	(0.0)
	3	18.769	0.011	32.536	0.065	1670.158	492.422	2.239
		(4.0)	(0.0)	(4.1)	(0.0)	(407.3)	(127.4)	(0.2)
	4	34.316	0.010	51.925	0.041	3383.513	1470.932	3.026
		(2.3)	(0.0)	(6.8)	(0.0)	(318.0)	(302.1)	(0.1)
	5	37.663	0.009	74.793	0.038	4286.781	2605.487	3.097
		(2.2)	(0.0)	(2.6)	(0.0)	(307.4)	(125.7)	(0.1)
	6	29.865	0.010	43.978	0.052	2951.538	958.488	2.555
		(1.3)	(0.0)	(8.2)	(0.0)	(285.8)	(349.5)	(0.1)

Note: Force unit is in N/kg and time unit is in s.
Loading rate unit is in N/kg/s and Impulse unit is in (N/kg)·s
Standard deviation values are in parentheses.
The definitions of variables are in Appendix A.

Table 17. Subject means and standard deviations of VGRF variables (Group 2).

Subj	Cond	F1	T1	F2	T2	LRF1	LRF2	Imp 100ms
4	1	26.476	0.01	50.630	0.055	2800.397	1211.003	3.242
		(2.5)	(0.0)	(8.7)	(0.0)	(476.6)	(437.0)	(0.1)
	2	31.154	0.011	59.616	0.059	2943.219	1201.394	2.418
		(2.4)	(0.0)	(4.2)	(0.0)	(252.2)	(188.8)	(0.1)
	3	35.358	0.013	51.479	0.061	2729.246	981.100	2.819
		(1.6)	(0.0)	(5.6)	(0.0)	(154.7)	(259.2)	(0.1)
	4	39.15	0.010	67.406	0.050	4187.597	1919.254	2.983
		(3.5)	(0.0)	(13.6)	(0.0)	(868.9)	(747.1)	(0.2)
	5	49.599	0.010	90.756	0.042	5152.791	3272.736	3.263
		(7.5)	(0.0)	(25.2)	(0.0)	(568.5)	(1450.4)	(0.3)
	6	59.842	0.012	92.652	0.047	5055.127	2540.656	4.262
		(4.5)	(0.0)	(7.2)	(0.0)	(199.1)	(418.4)	(0.3)
11	1	72.434	0.008	47.567	0.029	9657.904	3382.876	2.016
				(16.6)	(0.0)		(2160.0)	(0.1)
	2	17.295	0.013	88.966	0.034	1405.190	3334.419	3.654
		(3.1)	(0.0)	(12.4)	(0.0)	(152.8)	(1062.1)	(0.5)
	3	23.356	0.009	74.734	0.019	2824.957	6362.849	2.447
		(5.5)	(0.0)	(9.3)	(0.0)	(776.3)	(3592.2)	(0.3)
	4	27.025	0.008	92.446	0.032	3831.003	3802.032	3.719
		(1.7)	(0.0)	(13.6)	(0.0)	(1402.4)	(1552.7)	(0.3)
	5	32.284	0.006	100.859	0.034	5352.177	3788.936	3.798
		(4.8)	(0.0)	(18.7)	(0.0)	(815.1)	(1760.9)	(0.2)
	6	32.333	0.006	84.897	0.023	5542.765	3972.875	2.566
				(15.2)	(0.0)		(1049.4)	(0.3)
12	1	19.895	0.011	26.779	0.067	1815.815	404.109	1.853
		(1.0)	(0.0)	(5.2)	(0.0)	(135.8)	(97.1)	(0.2)
	2	19.933	0.012	35.817	0.071	1693.305	525.448	2.176
		(1.1)	(0.0)	(5.1)	(0.0)	(170.4)	(119.0)	(0.1)
	3	29.563	0.012	33.529	0.062	2507.088	553.221	2.183
		(1.1)	(0.0)	(9.6)	(0.0)	(162.5)	(204.2)	(0.3)
	4	29.544	0.012	46.210	0.064	2524.789	770.343	2.712
		(1.3)	(0.0)	(3.8)	(0.0)	(299.8)	(107.7)	(0.1)
	5	47.735	0.012	59.233	0.060	4059.254	1034.311	3.671
		(2.0)	(0.0)	(2.3)	(0.0)	(425.0)	(130.6)	(0.1)
	6	41.843	0.011	49.642	0.057	3660.036	953.232	2.810
		(4.5)	(0.0)	(3.5)	(0.0)	(525.4)	(126.5)	(0.1)

Table 17. (Continued).

Subj	Cond	F1	T1	F2	T2	LRF1	LRF2	Imp 100ms
14	1	19.260	0.009	37.026	0.048	2208.116	895.880	2.100
		(0.8)	(0.0)	(3.4)	(0.0)	(324.3)	(109.8)	(0.2)
	2	34.524	0.012	54.373	0.059	2884.456	1072.903	3.157
		(2.8)	(0.0)	(9.7)	(0.0)	(521.8)	(366.5)	(0.2)
	3	32.996	0.009	46.770	0.046	3560.599	1264.058	2.439
		(2.4)	(0.0)	(2.4)	(0.0)	(424.8)	(103.6)	(0.1)
	4	53.227	0.013	67.975	0.054	4078.167	1596.398	3.653
		(9.7)	(0.0)	(8.6)	(0.0)	(574.0)	(293.4)	(0.4)
	5	59.435	0.011	71.452	0.048	5329.045	1914.720	3.824
		(6.7)	(0.0)	(14.1)	(0.0)	(587.9)	(663.3)	(0.2)
	6	47.591	0.010	62.893	0.044	4923.751	2054.030	2.818
		(2.5)	(0.0)	(8.4)	(0.0)	(140.5)	(420.9)	(0.1)

Note: Force unit is in N/kg and time unit is in s.
Loading rate unit is in N/kg/s and Impulse unit is in (N/kg)· s
Standard deviation values are in parentheses.
The definitions of variables are in Appendix A.

Table 18. Subject means and standard deviations of VGRF variables (Group 3).

Subj	Cond	F1	T1	F2	T2	LRF1	LRF2	Imp 100ms
5	1	15.818	0.015	27.905	0.072	1064.770	384.412	1.863
		(4.8)	(0.0)	(3.3)	(0.0)	(311.7)	(110.5)	(0.2)
	2	17.240	0.015	36.343	0.072	1162.209	445.351	2.460
		(0.9)	(0.0)	(6.1)	(0.0)	(49.5)	(151.6)	(0.2)
	3	25.286	0.015	28.279	0.059	1706.856	498.170	2.697
		(2.1)	(0.0)	(3.1)	(0.0)	(167.6)	(165.4)	(0.0)
	4	31.048	0.015	36.643	0.067	2069.899	480.253	2.125
		(2.5)	(0.0)	(2.2)	(0.0)	(168.6)	(82.1)	(0.1)
	5	47.831	0.015	50.451	0.057	3261.489	915.203	2.513
		(2.9)	(0.0)	(4.8)	(0.0)	(173.8)	(186.1)	(0.1)
	6	33.500	0.015	34.173	0.058	2260.896	622.862	3.224
		(4.3)	(0.0)	(3.3)	(0.0)	(309.0)	(147.2)	(0.1)
6	1	19.948	0.010	33.618	0.061	2068.525	459.434	2.084
		(1.5)	(0.0)	(7.2)	(0.0)	(97.3)	(228.4)	(0.1)
	2	31.203	0.014	54.350	0.069	2230.665	726.418	3.266
		(1.8)	(0.0)	(3.0)	(0.0)	(153.2)	(104.6)	(0.2)
	3	33.981	0.012	44.964	0.063	2879.151	642.918	2.747
		(3.4)	(0.0)	(6.0)	(0.0)	(282.6)	(188.9)	(0.1)
	4	47.941	0.012	70.779	0.055	3995.455	1535.843	3.531
		(3.0)	(0.0)	(13.1)	(0.0)	(208.6)	(798.6)	(0.4)
	5	48.722	0.012	59.524	0.058	4033.667	939.908	2.948
		(3.6)	(0.0)	(7.5)	(0.0)	(522.1)	(249.8)	(0.1)
	6	48.359	0.010	47.960	0.055	4675.151	813.812	3.628
		(6.1)	(0.0)	(9.3)	(0.0)	(435.1)	(389.5)	(0.1)
10	1	18.415	0.011	31.917	0.072	1672.661	416.979	2.039
		(1.2)	(0.0)	(5.4)	(0.0)	(273.5)	(151.7)	(0.1)
	2	20.753	0.013	40.969	0.079	1670.489	465.153	2.589
		(1.6)	(0.0)	(5.1)	(0.0)	(167.1)	(96.2)	(0.3)
	3	31.276	0.012	48.040	0.056	2692.098	847.982	2.716
		(2.3)	(0.0)	(3.8)	(0.0)	(201.9)	(146.4)	(0.1)
	4	36.367	0.013	59.751	0.057	2787.492	1043.196	3.442
		(2.9)	(0.0)	(8.2)	(0.0)	(411.3)	(294.1)	(0.1)
	5	46.588	0.012	67.520	0.052	3823.665	1382.400	3.691
		(4.5)	(0.0)	(6.4)	(0.0)	(258.7)	(444.4)	(0.3)
	6	33.520	0.011	45.042	0.053	3123.183	952.892	2.619
		(4.1)	(0.0)	(10.1)	(0.0)	(526.1)	(636.3)	(0.3)

Table 18. (Continued).

Subj	Cond	F1	T1	F2	T2	LRF1	LRF2	Imp 100ms
15	1	22.253	0.014	32.753	0.072	1569.045	348.398	2.232
		(2.0)	(0.0)	(4.7)	(0.0)	(98.3)	(99.2)	(0.2)
	2	29.913	0.015	37.611	0.076	2016.872	361.878	2.685
		(3.4)	(0.0)	(4.5)	(0.0)	(222.2)	(105.1)	(0.1)
	3	26.681	0.013	37.350	0.061	2109.419	525.249	2.395
		(2.5)	(0.0)	(5.6)	(0.0)	(109.6)	(169.1)	(0.2)
	4	43.702	0.015	38.228	0.075	2913.474	347.821	2.958
		(6.6)	(0.0)	(2.6)	(0.0)	(437.9)	(62.1)	(0.1)
	5	62.330	0.015	73.653	0.039	4155.343	3279.291	3.559
		(14.1)	(0.0)	(17.9)	(0.0)	(942.4)	(2256.3)	(0.2)
	6	51.325	0.013	50.559	0.058	3809.955	919.911	2.965
		(5.8)	(0.0)	(5.5)	(0.0)	(459.6)	(272.1)	(0.1)

Note: Force unit is in N/kg and time unit is in s.
Loading rate unit is in N/kg/s and Impulse unit is in (N/kg)·s
Standard deviation values are in parentheses.
The definitions of variables are in Appendix A.

APPENDIX E

INFORMED CONSENT FORM

INFORMED CONSENT FORM

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You are invited to participate in a research study entitled "The relationship between static postural measurements and biomechanical characteristics during landing" which examines the effects of various landing heights and techniques during a landing activity.

You should be healthy, physically active, and between the ages of 18 and 25 with no history of impairments to your lower extremity. If you choose to participate, you will be asked to attend one screening session (10 minutes) and one testing session (one and one-half hours). Please wear loose shorts and a comfortable short-sleeved shirt or tank top when you report to the lab. The screening session will involve measurements of standing knee extension, pelvic angle, and foot or arch height. At the beginning of the test session, you will need to fill out an information sheet about your age, height, and recreational sport activities. You will be asked to warm-up on a stationary bike for 5 minutes. Following this, we will take anthropometric measurements of the lengths and girths of your thigh, leg, and foot. Before the actual testing, you will become familiar with the testing protocol by performing 3 landing trials on a force platform. You will then be asked to perform 5 step-off landings in each of six test conditions from a raised platform onto the force platform. The six test conditions include combinations of three landing heights and two different landing techniques (stiff and soft). The three landing heights will be set at 45, 60, and 75cm. The order of the condition presentations will be randomized. During the test, biomechanics instruments will be used to make measurements. Some of these instruments will be placed/fixed on your body. None of the instruments will impede your ability to engage in normal and effective movements during the test. If you have any further questions, interests, or concerns about any instrumentation, please feel free to contact the investigator.

The potential risks include an ankle sprain from landing in an unbalanced manner and muscular strains to the lower extremity. Every effort will be made to reduce these risks through proper warm-up, sufficient practice before the test, and use of spotters. All tests will be conducted and qualified research personnel in the Biomechanics/Sports Medicine Lab will handle the equipment. However, the physical requirements of the study are not beyond those observed in normal weight training or recreational sport activities. The Biomechanics/Sports Medicine Lab has tested more than 130 subjects in several jumping/landing related studies over the past five years. None of them was injured in any fashion during the test sessions. You will be encouraged to warm-up actively prior to each testing session so that you feel physically prepared to perform effectively and thus minimize any chance for injury. Should any injury occur during the course of testing, standard first aid procedures would be administered as necessary. At least one researcher with a basic knowledge of athletic training and/or first aid procedures will be present at each test session. In the event that a physical injury is suffered as a result of participation in this study, the University of Tennessee does not automatically provide reimbursement for medical care or other compensation. Your benefits include assessment of your performance and biomechanics of the landing tasks. You are welcome to make an appointment to review the data from your tests. In addition, if you wish to have a copy of the results, please let me know.

Your participation is entirely voluntary and your decision of whether or not to participate will involve no penalty or loss of benefits to which you are otherwise entitled. Your identity as a subject will be held in strict confidence and only a subject number will be used to refer to any description of your data. Any information that is obtained in connection with this study and that can be identified with you will remain confidential and will be disclosed only with your permission.

After you have read this informed consent form and all of your questions have been answered, you are requested to sign and date the form below and the attached form that lists individual subject requirement. Your signature indicates that you have read and understand the information provided above, that you willingly agree to participate, that you may withdraw your consent at any time, and discontinue participation at any time without penalty or loss of benefits to which you are otherwise entitled.

Subject Name: ____________________ Signature: ____________________ Date: ____________________

Investigator: ____________________ Date: ____________________

VITA

James Craig Garrison was born in Guymon, OK on September 14, 1969. After graduating from Guymon Highschool, he attended Texas Christian University in Ft. Worth, TX where he received his Bachelor of Science in Movement Science in 1992. In August of 1993, he began Physical Therapy school at the University of Oklahoma, completing his degree in May of 1995. Upon completion of his physical therapy training, Craig accepted a job with a sports medicine clinic in Ft. Worth, TX. He worked for two years as a staff physical therapist and then progressed into management. In the fall of 2000, he began his graduate studies in Exercise Science with a concentration in Sports Medicine and Biomechanics at the University of Tennessee, Knoxville. As a graduate student, he was involved with several research projects, worked as a graduate teaching associate, and also worked part-time in the physical therapy clinic. He received an A.W. Hobt Teaching Award for the 2001-2002 school year. After completing his thesis, he received the Master of Science degree in Human Performance and Sport Studies in August of 2002.