Published in final edited form as:

J Orthop Res. 2013 February ; 31(2): 257–267. doi:10.1002/jor.22217.

High Knee Valgus in Female Subjects Does Not Yield Higher Knee Translations During Drop Landings: A Biplane Fluoroscopic Study

Michael R. Torry, PhD⁶, Kevin B. Shelburne, PhD², Casey Myers, MS², J. Erik Giphart, PhD¹, W. Wesley Pennington, MS⁵, Jacob P. Krong, MS¹, Daniel S. Peterson, MS⁴, J. Richard Steadman, MD¹, and Savio L-Y. Woo, Ph.D., D.Sc., D.Eng.³

¹Biomechanics Research Department, Steadman Philippon Research Institute, Vail, CO

²Dept. of Mechanical and Materials Engineering, The University of Denver, Denver, CO

³Musculoskeletal Research Center, Department of Bioengineering, Swanson School of Engineering, University of Pittsburgh, Pittsburgh, PA

⁴Dept. of Biomedical Engineering Program in Physical Therapy, Washington University, St Louis, MO

⁵Dept. of Kinesiology, Indiana University, Bloomington, IN 47401, USA

⁶School of Kinesiology and Recreation, Illinois State University, Normal, IL

Abstract

The goal of this study was to determine the effects of peak knee valgus angle and peak knee abductor moment on the anterior, medial, and lateral tibial translations (ATT, MTT, LTT) in the ‘at risk’ female knee during drop landing. Fifteen female subjects performed drop landings from 40 cm. 3D knee motion was simultaneously recorded using a high speed, biplane fluoroscopy system and a video-based motion analysis system. Valgus knee angles and knee abduction moments were stratified into low, intermediate and high groups and peak ATT, MTT and LTT were compared between these groups with ANOVA ($\alpha = .05$). Significant differences were observed between stratified groups in peak knee valgus angle ($p < .0001$) and peak knee abduction moment ($p < .0001$). However, no corresponding differences in peak ATT, LTT and MTT between groups exhibiting low to high peak knee valgus angles (ATT: $p = .80$; LTT: $p = .25$; MTT: $p = .72$); or, in peak ATT ($p = .61$), LTT ($p = .26$) and MTT ($p = .96$) translations when stratified according to low to high knee abduction moments, were found. We conclude that the healthy female knee is tightly regulated with regard to translations even when motion analysis derived knee valgus angles and abduction moments are high.

Keywords

Knee; ACL; knee valgus; fluoroscopy; knee translation

Introduction

The annual rate of injury to the anterior cruciate ligament (ACL) is reported to be above 1 in 3000 in the United States with a large proportion (~70%) denoted as a non-contact

⁶Correspondence Author: Michael R. Torry, PhD, School of Kinesiology and Recreation, Illinois State University, Normal, IL, mrtorry@gmail.com, Ph: 309-438-7501.

mechanism of injury.^{1, 2} It has been well documented that females exhibit a higher incidence of ACL injury compared to males. Laboratory studies utilizing traditional motion analysis techniques have demonstrated that females exhibit high knee valgus angle and high knee abduction moment profiles more often than males during stop jump and drop landing movements.^{3–8} High knee valgus angles and abduction moments measured during drop landings have been shown to predict ACL injury in female athletes.^{8, 9} These *in vivo* data are supported by human cadaveric and musculoskeletal modeling studies showing that increased knee abduction loads can cause increased forces in the ACL,^{10–13} which can be exacerbated with increased anterior tibial translations.¹¹

The results of current ACL prevention/pre-screening programs^{3–5, 9, 14, 15} in conjunction with the supportive cadaveric studies^{10, 11, 16–18} suggest that those females possessing high knee valgus and/or high abductor moment *in vivo* may also possess correspondingly higher tibial translations, and therefore higher risk for ACL injury. While many of these *in vitro* and computational studies support the notion that ACL load is associated with anterior tibial translation, there is a void in the *in vivo* literature to substantiate that these translations occur during drop landings; or specifically, if tibial translations are higher in those individuals who exhibit high knee valgus angle and abductor moment landing patterns. This is because the determination of knee translations *in vivo* using traditional motion analysis techniques is difficult due to soft tissue artifact.¹⁹ Consequently, investigators utilizing traditional motion analysis techniques have not determined what effects a high knee valgus and/or high abductor moment have on knee translations in these “at risk” populations. Thus, a method to establish the relationship between these parameters *in vivo* is critical to understanding how the knee functions under high demand activities in order to provide better insight as to the mechanism of the non-contact ACL injury.

The objectives of this study were: 1) to utilize a traditional motion analysis system in combination with high-speed, biplane fluoroscopy imaging to measure the differences in knee kinematics between females exhibiting low, intermediate and high knee valgus angles and abduction moments during drop landings; and, 2) to describe the relationships between anterior, medial and lateral tibial translations (ATT, MTT, LTT), and knee valgus angles and knee abductor moments in “at risk” female athletes. Based on the previous literature and the association between high knee valgus angle and/or abductor moment and ACL injury^{9, 14}, we hypothesized that females exhibiting higher knee valgus and/or higher knee abductor moments would also experience larger ATT, MTT and/or LTT values during the drop landing.

Methods

Subject Recruitment and Subject Testing Protocol

Recruitment of the subjects consisted of an initial visual pre-screening (by author MRT) in which potential participants were asked to jump and land from a 40 cm box. No verbal or visual instruction as to landing style was given prior to or during this initial screening. Fifty three subjects were screened in this manner. No exclusion criteria or cutoff based on the degree or severity of knee valgus motion was utilized in this pre-screening and selection process. If the subject consistently exhibited a knee valgus landing pattern, they were asked to participate further in the study. From the 53 initially screened, 18 were identified visually as possessing valgus knee motions during the landing. Once identified, these individuals were introduced to the testing apparatus and the risks of the study were disclosed and discussed via the informed consent documents. The radiation dose and the risk to the subject's health were also presented and discussed with the subjects at this time. With the combined biplane fluoroscopy testing and the CT scan of the knee (as detailed below), the effective dose to the subject was estimated to be 15.4 mrem, which is 0.3% of the FDA limit

of 5000 mrem for research studies on participants over age 18 years of age.²⁰ Of the 18 individuals identified in the pre-screening, 15 agreed to participate further. These 15 recreational athletes (Age 26.1 ± 6.3 yrs.; Height 167.9 ± 6.3 cm, Body Mass 58.2 ± 5.2 Kg) self-reported at least two days a week of active court sport activities and had no history of lower extremity injury. Prior to participation, each subject was provided and signed informed consent that was approved by the local Institutional Review Board.

The methods for the bone pose estimation from the fluoroscopy images require 3D geometry of the test knee of each subject. To obtain this geometry, each subject completed a supine, high-resolution (voxels: $0.7 \times 0.7 \times 0.5$ mm), static bone CT scan (12 cm above and below the joint line) in an Aquilion 64 (Toshiba America Medical Systems, Tustin, CA), with scan technique factors of 120 kVp and 200 mA, which are used routinely to image these bones for clinical detail. The subjects wore a lead apron and genital shield to help screen radiation from sensitive organs during the CT scan. Three dimensional models of the femur and tibia/fibula were reconstructed from the bone contours extracted from the CT images using commercial software (Mimics, Materialize, Inc., Ann Arbor, MI).

Following CT, simultaneous motion capture and biplane fluoroscopy data were collected while each subject performed a drop landing. The subjects were provided spandex-like, tight-fitting clothing and standardized court shoes (Turntec, model no. TM08061). The dominant kicking leg for all subjects was the focus of the data collection and analysis. All subjects self identified as right leg dominant. Each subject performed a drop-landing from a 40 cm platform while simultaneous motion analysis and biplane fluoroscopy system data were collected. The subjects were instructed to step off the platform and land onto a force plate (Bertec Corp., Columbus, Ohio) fixed to the laboratory floor and then immediately perform a peak height vertical jump as previously described.⁸ As in the initial pre-screening session, no verbal or visual instruction as to the landing style was given.

After completion of a successful landing trial, the subjects entered the capture volume area and completed a slow (2 sec), unloaded knee extension motion from a seated position (hip angle at 90°) starting at a knee flexion of 90° to the fully extended position while simultaneous motion capture and biplane fluoroscopy data was collected. This motion was used to define the zero reference position of the tibia relative to the femur at full extension for the biplane data set.²¹

Testing Equipment

Traditional motion analysis techniques were used to collect kinematic data of the test limb at 240 Hz using a 10-camera system (Motion Analysis Corporation, Santa Rosa, CA) positioned around the landing. The 3D kinematics of each trial were captured by securing retro reflective spherical markers to anatomical landmarks on each subject as reported previously^{7, 22, 23} to produce a standard, 3-marker-per-segment configuration. Subjects were allowed to practice their landings until they could land consistently within the calibrated area. To facilitate this, additional motion analysis surface markers were placed around the circular edges of the two fluoroscopy image intensifiers. These markers in conjunction with the markers on the subjects' knee, could be viewed in 3D within the motion analysis software in real time and thus allowed us to observe the subject's knee during the landing in relation to the capture volume. This process assisted in minimizing radiation exposure to the subject as it provided us with immediate feedback as to when the subject was landing consistently within the biplane calibrated area before any radiation trials were attempted. We then collected a single 'good, radiation live' landing trial for the subject; with a 'good trial' defined as one where the knee remained within the fluoroscopy 3D field of view for the complete landing phase.

The force plate data were sampled at 1200Hz and synchronized with the fluoroscopy video images such that the specific landing contact fluoroscopy frame could be determined.^{24, 25} This was accomplished by simultaneously recording the ground reaction force data with the triggering signal of the high speed cameras of the fluoroscopy system.

The biplane fluoroscopy system was constructed from two BV Pulsera c-arms with 30 cm image intensifiers (Philips Medical Systems, Best, The Netherlands), which were modified and coupled to two high-speed, gen-locked (500 Hz), high-resolution (1024×1024) digital cameras (Phantom V5.1, Vision Research, Wayne, NJ) using a custom built interface. The biplane fluoroscopy system was corrected for image distortion and calibrated using techniques that have been previously described.^{26, 27} The focus position of each fluoroscopy system and the 3D relationship between the two fluoroscopy systems were calculated by imaging a 15 cm^3 calibration cube enclosing 15 tantalum markers positioned inside the 24 cm^3 3D capture volume of the biplane fluoroscopy system. A combination of custom software (Matlab, The MathWorks, Natick, MA) and MBRSA (Medis Specials, Leiden, The Netherlands) were used to complete the calibration process.²⁷

Motion Analysis System Data Processing

Motion analysis marker trajectories and force data were filtered at 20Hz using a fourth order Butterworth filter.²⁸ Euler joint angular positions as well as velocities, and accelerations were calculated from the filtered 3D marker coordinate data using the flexion-extension, varus-valgus, internal-external rotation (YXZ) sequence and estimates of the (external) knee joint moments were subsequently calculated.^{7, 22, 25, 29}

Biplane Fluoroscopy Data Processing

For each trial, determination of the bone poses from the biplane fluoroscopy data were performed using commercial software (MBRSA, Medis Specials, Leiden, The Netherlands). For each frame, both inner and outer contours of the femur and tibia/fibula were semi-automatically extracted from the biplane fluoroscopy images (automatically detected, manually assigned to each bone). A fully-automatic 6 degree-of-freedom optimization algorithm was used to determine the pose, which matched the detected contours with the projected contours from the imported bone geometries.^{30, 31} Using these pose sequences, knee kinematics were calculated using methods described by Grood & Suntay³² where knee rotations and translations indicate motions of the tibia with respect to the femur. Specifically, the origin of the femoral coordinate system was placed at the midpoint between the medial and lateral femoral condyles on the center line of a cylinder fitted to the medial and lateral posterior condyles. The medio-lateral (ML) axis of the femur was assigned as the line through the long axis of this cylinder. The superior-inferior (SI) axis was placed along the posterior line of the femur, and an anterior-posterior axis was determined as the cross product of the ML and SI axes.³³ The coordinate system assigned to the tibia was co-located with the femoral coordinate system at full extension (i.e., 0 deg) in the knee extension trial and produced positive values for the flexion, valgus and external tibial rotations.^{24, 29, 34–36}

The kinematic accuracies for tracking bones during a dynamic land/impact using MBRSA and this biplane fluoroscopy system were determined and have been described previously^{24, 35}. In brief, the accuracy and precision (mean and SD) of tracking 1.0 mm diameter tantalum markers were $-5.5 \mu\text{m}$ and $32.5 \mu\text{m}$, respectively.³⁵ The mean and ± 1 standard deviation of the differences in joint kinematics between those determined by tracking the tantalum markers and those by MBRSA bone tracking methods were 0.07 ± 0.7 mm, 0.1 ± 0.6 mm in medial-lateral and anterior-posterior translations; and $0.09 \pm 0.4^\circ$, $0.1 \pm 0.05^\circ$, $0.1 \pm 0.8^\circ$ in flexion, varus-valgus, internal-external rotations, respectively.³⁵ These

values are consistent with previous work by others using the MBRSA software^{24, 30, 37} and researchers using other bone-contour tracking software.³⁸

Following the examples of precision motion measurements taken from cadaver experiments³⁹ tibial translations and rotations measured from each subject during landing were referenced to the unloaded knee extension motion. Referencing to the motion of the knee collected during a low load activity, such as knee extension in which the ACL is generally taut but not overly strained^{40, 41}, is important for studying the change in kinematics when loads are applied^{42, 43} and reduces intra-subject variability.^{24, 25, 29}

Data Analysis

Linear regression analyses were conducted to determine if peak knee valgus angles and abductor moments were significant predictors of peak anterior tibial translation (ATT), peak medial (MTT) and/or peak lateral tibial translations (LTT) during the drop landing. In order to investigate differences in translations across individuals exhibiting low, intermediate and high peak knee valgus angle and peak knee abductor moments, the subject pool (N = 15) was stratified into groups of five according to each of these variables as measured by the motion analysis system. The peak ATT, MTT and LTT between these subdivisions were then compared utilizing an ANOVA ($\alpha = .05$) with a Bonferroni adjusted p-value (.016) for post-hoc comparisons. A paired sample t-test was utilized to compare differences in the peak knee valgus and peak internal-external tibial rotation angles measured via the biplane fluoroscopy and the motion analysis systems.

Results

As measured by the motion analysis system, the subjects landed with knee flexion angles of $14.6^\circ \pm 4.6^\circ$ at ground contact, experienced peak flexion angles of $83.6^\circ \pm 9.2^\circ$ and demonstrated peak resultant ground reaction forces of $1,315.2 \pm 629.3$ N (22.3 ± 9.4 N/Kg), which is consistent with previous landing studies.^{7, 22, 44} The subjects generally contacted the floor with the tibia internally rotated (mean peak, $6.4^\circ \pm 6.7^\circ$) and then proceeded into an externally rotated position (mean peak, $-5.2^\circ \pm 7.3^\circ$). The motion analysis system yielded larger peak knee valgus angles ($6.2^\circ \pm 7.0^\circ$) compared to those obtained via fluoroscopy ($1.8^\circ \pm 1.4^\circ$; $p = .03$). The motion analysis system also yielded larger values for peak external tibial rotation (mocap: $5.3^\circ \pm 7.4^\circ$ vs. fluoroscopy: $0.4^\circ \pm 4.3^\circ$; $p = .01$), but similar values for peak internal rotation (mocap: $6.4^\circ \pm 6.7^\circ$ vs. fluoroscopy: $5.2^\circ \pm 4.5^\circ$; $p = .58$; $\beta = .08$). The motion analysis system produced larger values ($\sim 7^\circ$) of the internal-external tibial rotation range of motion (mocap: $11.7^\circ \pm 5.9^\circ$ vs. fluoroscopy: $4.7^\circ \pm 2.7^\circ$; $p = .03$).

The subjects demonstrated average peak knee abductor moments of 16.6 ± 10.8 Nm as derived by the kinematic and ground reaction force data obtained from the motion analysis system.

As measured by the biplane fluoroscopy system, subjects experienced average peak ATT, LTT and MTT values of 4.6 ± 1.7 mm, 0.3 ± 1.0 mm and -2.4 ± 1.1 mm, respectively (Table 1). The average ATT excursions across the subjects were 3.8 ± 2.1 mm and the range of lateral to medial tibial excursions were 2.8 ± 1.2 mm. The time to peak ATT occurred at 48 ± 19 msec after ground contact.

No significant relationships were observed between peak knee valgus angles as measured by the motion analysis system and peak ATT, LTT and MTT values as measured by the biplane fluoroscopy system ($p = .80$, $p = .25$, and $p = .71$, respectively; Table 1, Figures 1A and 1B). Similarly, no significant relationships were observed between peak knee abductor moments

and peak ATT, LTT and MTT values ($p = .61$, $p = .26$, and $p = .96$, respectively; Table 2, Figures 2A and 2B).

After stratifying the original subject pool ($N = 15$) into three groups ($n = 5$) according to low, intermediate and high peak knee valgus angle as measured by using the motion analysis system, differences were noted in the peak knee valgus angles ($p < 0.0001$; Table 1) with the 'High' peak knee valgus group exhibiting significantly larger valgus angles than the 'Intermediate' or 'Low' knee valgus angle groups (both $p = 0.013$). However, ANOVA failed to yield significant differences in peak ATT, LTT and MTT between these groups (Table 1).

Differences were also noted between all three groups when stratified by each person's peak knee abductor moment (Table 2, $p < .0001$) with the High knee abductor moment group exhibiting significantly greater peak knee abductor torque values compared to the Intermediate and Low groups (both $p = .01$). However, no differences in peak ATT, LTT and MTT were observed in conjunction with this stratification (Table 2). This result remained after the knee abductor moment values (Nm) were normalized to subject body mass (Nm/Kg).

To investigate the effect of peak internal and external rotation on ATT, LTT and MTT, an additional statistical comparison was performed based on these rotations. Significant differences were observed in both peak internal and peak external tibial rotation angles when stratified into groups by each subject's peak internal-external values as measured by the motion analysis system (Tables 3 and 4, $p < .0001$). However, no differences were noted in peak ATT, LTT and MTT in conjunction with this stratification (all $p = .25$).

There was no difference in the time to maximal ATT when the group was stratified by valgus knee angles or knee abduction moment (all $p = .26$).

Discussion

Gender based landing studies have identified females who are "at risk" for non-contact ACL injury as those who exhibit a high knee valgus angle and/or a high knee abductor moment profile during drop landing. These landing mechanics are thought to provoke high ACL loads⁷⁻⁹ and have been shown to be predictive of non-contact ACL injury.⁹ The current study used motion analysis techniques that have traditionally been used to identify females who are "at risk" in these ACL prevention/prescreening motion analysis programs. In addition, the present study used a high speed biplane fluoroscopy system simultaneously to determine the bone movements directly and to determine the relationship of high knee valgus angles and knee abduction moments with anterior and medial-lateral knee translations occurring during this motion. The results indicate that neither high knee valgus angles nor high knee abduction moments are predictive of peak ATT, LTT and MTT translations during drop landing. To reinforce this finding, the data were stratified into High, Intermediate and Low peak knee valgus angle and peak knee abductor moment groups as measured with the motion analysis system. The result of this stratification suggests that even when *large* differences are detected in valgus knee angles or in knee abductor moments as measured by the traditional motion analysis technique; there is no difference in peak ATT, LTT or MTT values among these groups.

The lack of association in the knee translations with the peak knee valgus angle was unexpected as literature has suggested valgus angle to be one of the major predictors for ACL injury.⁹ The reason for this lack of association may be related to several factors. Recently, a cadaveric study has shown that intact knee specimens exhibit valgus angles below 5° with 10 Nm applied valgus load. The valgus angle can increase to 9° (at 0° flexion)

and to 20° (at ~30° flexion) after the medial ligaments are fully sectioned.⁴⁵ Thus, it is unlikely that healthy individuals are achieving the peak knee valgus angle values reported by many of the *in vivo* ACL gender laboratory landing studies.^{4, 9, 22, 46, 47} This is coupled with results suggesting the majority of non-contact ACL injuries do not involve the MCL¹⁵. In our measurements, the motion analysis system consistently yielded larger peak knee valgus and internal-external rotation angles (by ~5°) when compared to those values derived from the biplane fluoroscopy system. These findings are in agreement with a recent study using static biplane fluoroscopy technology.⁴⁸ Iribarra et al.,⁴⁸ showed that ACL length does not increase when the knee is “forced” into a ‘valgus’ position during single limb stance. The present study and the data provided by Iribarra et al. suggests large valgus knee angles observed visually or measured via traditional motion analysis techniques may not necessarily equate to high valgus angles at the joint level, which are thought to occur after ground contact during the drop landing. Because we did not capture data with significant lead time prior to ground contact, nor did we collect soft tissue images, we cannot make direct comparisons to previous studies^{49–51} with regard to the valgus angle and ACL length or strain values from our data set.

Several studies demonstrate that neuromuscular training reduces high knee abductor moments, increases performance and decreases knee injury incidence in female athletes.^{5, 8, 52, 53} The current study found no differences in peak knee translations as measured via biplane fluoroscopy when considering the higher knee abductor moment profiles. The lack of association in knee translations with the peak knee abductor moment, however, is not mechanically unreasonable. Inverse dynamic calculations and the net moments they estimate do not account for co-contractions, which may serve to limit or negate joint translations. This is supported by a recent modelling and simulation study in which high knee abductor moments alone, while slightly increasing ACL load, did not achieve rupture strain values.⁵⁴ Previous studies have shown that the soft tissues of the knee joint are able to regulate and tightly control translations of the knee in response to external loading conditions applied *in vivo*. Specifically, Myers et al.²⁴ showed that there was no difference in peak ATT (4.7 ± 1.6 mm vs. 4.4 ± 0.8 mm) and varus-valgus peak knee angles ($1.7^\circ \pm 1.2^\circ$ vs $1.6^\circ \pm 0.9^\circ$) between soft and stiff landings. Myers et al.²⁵ also reported that stiff drop landings produced significantly greater peak ATT (5.6 ± 1.9 mm) than walking (3.1 ± 2.2 mm) and an unweighted full knee extension (2.6 ± 2.1 mm), but no difference between stiff landings and a maximum isometric quadriceps contraction (5.0 ± 1.9 mm). With peak ATT differences of approximately 2–3 mm observed between these motions, the findings of these studies^{24, 25} emphasize that in healthy female knees, the musculature and soft tissues are able to maintain knee translations within a relatively small, tightly regulated range during controlled tasks of differing load demands. The ability of the knee soft tissues to closely regulate tibio-femoral translations even when demand of the task is greatly increased, may also serve to explain why we did not observe differences between our knee valgus angle and knee abductor moment stratified groups. Despite the implications for ACL injury, high knee valgus angle and high knee abductor moment may not indicate large enough variations in knee joint loads to result in greater translations in these high valgus angle and/or high knee abductor moment female landers during laboratory-controlled drop landings.

Internal and external tibial rotations in conjunction with tibial torques have been shown to increase ACL strain *in vitro*, with internal tibial torques exerting a greater influence.¹¹ The present study found no differences in those who exhibit low or high peak internal and external tibial rotations. Moreover, the subjects in this study exhibited similar internal-external rotation values as in a previous study evaluating a small group of male landers,³⁶ suggesting that high knee valgus angles do not necessarily equate to simultaneously high internal-external tibial rotations. Analysis of Coefficient of Variation [defined as (SD/

Mean}*100] for internal-external and varus-valgus knee rotations during landings suggests that internal-external rotations angles display greater variability^{55, 56} and this variability can increase with factors such as increased fatigue of select hip and ankle musculature⁵⁶. This suggests a lack of neuromuscular control in this plane. The present study has shown that biplane fluoroscopy can yield lower Coefficient of Variation in the internal-external tibial rotation assessment of dynamic knee function. Since ACL load is sensitive to these tibio-femoral rotations and their associated torques¹¹, understanding how knee rotation, and the neuromuscular control of its variability affects the ACL *in vivo* may help predict ACL injury. With increased sensitivity to internal-external rotations, biplane fluoroscopy offers a unique opportunity to investigate this relationship with greater insight in future studies.

The time required to reach peak ATT after initial ground contact is important for the design of neuromuscular training programs to prevent noncontact ACL injury. This study shows that peak ATT occurs within 48 ± 19 msec after ground contact in females performing a drop landing. This time frame was not altered by the degree of knee valgus or magnitude of knee abduction torque. This was also within the 40–50 msec ranges exhibited by males performing a stiff landing³⁶ and ‘non-at risk’ females and males performing soft and stiff landings.²⁴

Limitations of the present study are recognized. A cohort of 15 female subjects is low in comparison to the notable predictive studies employing larger number of subjects.^{9, 57} A larger sample size may have elicited stronger trends in the regression analysis. Yet, we were able to effectively delineate the cohort into high, intermediate and low knee valgus angle and knee abductor moment groups. Thus, adequate statistical power was achieved via our stratification scheme to detect differences in the currently accepted variables as measured by the motion analysis system and the derived kinematic and kinetic variables it yields are considered to be associated with increase risk of suffering non-contact ACL injury in females. However, this stratification design did not yield correspondingly similar statistical differences with regard to the peak knee translations as measured by the biplane system. This could be due to a smaller effect size in the biplane fluoroscopy derived measures. However, the data demonstrated no correlation between the motion analysis derived and biplane fluoroscopy derived variables; and, visual inspection of the data in the tables demonstrates no clear tendency to suggest perhaps a small effect size.

The analysis of a single trial per subject is a limitation as this removes intra-subject performance variability, which is an important aspect to motion assessment and possibly to injury potential. As biplane methods continue to improve and radiation exposure to the subject per trial decreases, future studies implementing multiple trial methodologies may prove valuable in this regard.

The field of view for our biplane system is approximately 24 cm^3 . It may be criticized that this low capture volume limited the type of landing to a vertical drop landing and not a side-cutting or stop-jumping motion, which may create knee joint angles, moments and external loads that more directly mimic ‘in game’ injury scenarios. Lastly, although kinematic comparisons between biplane fluoroscopy and traditional motion analysis techniques were made, the differences in the knee angles observed between the two systems in the present study are noted with caution as they may be prone to inherent differences in the anatomical coordinate systems utilized for each data collection system. The inverse dynamics process of the motion analysis system utilized an embedded coordinate system defined by a mechanical axis (i.e., the thigh segment is based on the long axis of the femur extending from the estimated knee joint center to the estimated hip joint center). Alternatively, the biplane fluoroscopy system utilized a distal femur cylinder fitting technique derived from the geometry of the 3D tibio-femoral bone model. Thus, the axes of the bone model derived

coordinate system may not be directly inline or perpendicular to the mechanical axis of the motion analysis system's anatomical coordinate system; and, unless calibrated with a common 3D calibration object, neither is in the same global coordinate system. These differences may have introduced bias in making the direct comparisons noted herein. However, previous work has shown that when global coordinate systems are spatially aligned only small (< 3.0 mm) differences in knee joint center estimations are found during walking compared to the 2-marker and point cluster technique even when different anatomical coordinate system methodologies are employed.⁵⁸

This study employed two very different motion analysis systems in an attempt to understand the relationships that may exist in knee function as measured by each. In general, the motion analysis system produced larger ($\sim 5\text{--}7^\circ$) joint angular values in all three planes at the knee during the landing. However, the biplane fluoroscopy system is limited by its small field of view and the use of radiation; and, both systems have inherent errors in anatomical coordinate assignments. Even so, published techniques such as those utilized by Taylor et al.⁵⁰ show that both of these motion capture modalities can be used in combination to complement each other and overcome some of each system's weaknesses.

The mechanisms leading to non-contact ACL injury are complex and difficult to directly observe; and, as reviewed by Hashemi et al.⁵⁹, numerous theories exist and have been studied with no clear *causal* mechanism identified. However, the "valgus knee collapse mechanism" has emerged as a leading performance factor predictive of ACL injury. This study was aimed at supporting this theory by demonstrating a direct link between the high risk performance factors and knee translations *in vivo*. We found no such associations. Within the scope and limitations of this study, the data suggests that while the performance metrics (high knee valgus angle and abductor moment) may be adequate and plausible *predictors* of the non-contact injury, they may ultimately not be the *isolated or causal* factors of the injury if one considers knee translation to be a key aspect of the injury mechanism. Although we found no relationships between either high knee valgus angle or high knee abduction moment with knee translations in the present study, we find it principle to state that our findings do not undermine the use of ACL injury prevention programs and current screening methods. The lack of relationships suggests that perhaps high knee valgus is a secondary indicator of risk to the knee, because no direct effect on knee translations was found in this study. We believe these programs should continue in the full effort toward prevention alongside efforts to discover new theories and paradigms⁵⁹ to support a deeper understanding of this injury.

In conclusion, although compelling evidence has identified the high knee valgus angle and the high knee abductor moment as predictors for ACL injury, subjects who possessed these performance parameters did not yield higher peak knee translations during the drop landing motion; suggesting valgus angles and abductor moment profiles observed visually or as measured via traditional motion analysis systems, may not necessarily equate to higher valgus angles and/or higher translations at the joint level.

Acknowledgments

This study was funded in part by the Steadman Philippon Research Institute and a grant from the National Institutes of Health (AR39683 to PI: Savio L-Y. Woo). The authors wish to thank Andrea North, Elizabeth Hageman and Nicole Pinwell for help with data collection and the reduction process. Medis Specials is acknowledged for providing the MBRSA software.

This research was performed at the Biomechanics Research Department of the Steadman Philippon Research Institute

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Figure 1. FIGURE 1A (top): Regression plot and equation of peak knee valgus angle versus anterior tibial translation; 1B (bottom): Regression plot and equation of peak knee valgus angle versus lateral tibial translation.

Figure 2.

FIGURE 2A (top): Regression plot and equation of peak knee abductor moment versus anterior tibial translation; 2B (bottom): regression plot and equation of peak knee abductor moment versus lateral tibial translation.

Table 1

Stratification of subjects based on Peak Knee Valgus Angle and comparison of peak knee translations across groups.

Group	Subject ID	Peak Valgus Angle (deg)	MAX ATT (mm)	MAX LTT (mm)	MAX MTT (mm)
Low	10	-2.90	4.18	1.82	-3.28
	1	-1.90	3.93	1.28	-1.13
	3	-0.50	6.35	-0.35	-3.04
	6	0.20	3.87	1.92	-1.37
	15	2.53	4.33	0.32	-1.91
Intermediate	13	3.68	9.22	0.08	-3.60
	8	4.20	3.05	0.39	-2.10
	4	4.80	6.34	-0.69	-4.14
	5	5.00	3.77	-0.01	-1.55
	14	7.95	3.01	1.05	-2.25
High	11	9.23	4.87	0.86	-3.99
	7	10.10	3.31	-1.95	-3.38
	12	10.51	4.61	1.16	-1.25
	9	19.60	2.97	-0.75	-1.54
	2	20.80	5.19	0.15	-1.38
Group (N = 15)					
Mean ($\pm$ 1SD)		6.2 (7.0)	4.6 (1.7)	0.3 (1.0)	-2.4 (1.1)
Low Knee Valgus Group (n = 5)					
Mean ($\pm$ 1SD)		-0.5 (2.0)	4.5 (1.0)	1.0 (0.9)	-2.15 (1.0)
Intermediate Knee Valgus Group (n = 5)					
Mean ($\pm$ 1SD)		4.3 (2.8)	5.0 (2.7)	0.2 (0.6)	-2.73 (1.1)
High Knee Valgus Group (n = 5)					
Mean ($\pm$ 1SD)		14.1 (5.6)	4.2 (.9)	0.1 (1.1)	-2.31 (1.3)
ANOVA		p < .0001			
Post Hoc Comparisons (Bonferroni)		p = .80; β = .076 p = .25, β = .079 p = .72; β = .111			

Group	Subject ID	Peak Valgus Angle (deg)	MAX ATT (mm)	MAX LTT (mm)	MAX MTT (mm)
Low vs High Knee Valgus Groups		p = .0006	na	na	na
Low vs Intermediate Knee Valgus Groups		p < .0001	na	na	na
Intermediate vs High Knee Valgus Groups		p = .013	na	na	na

Peak Valgus Angle (deg) = peak knee valgus angle measured by motion capture system

Max ATT = peak anterior tibial translation measured by bi-plane fluoroscopy

Max LTT and MTT = peak (+)lateral and (-)medial translation measured by biplane fluoroscopy

Bonferroni adjusted alpha for p-value is p = .016

na = no post hoc required

Table 2

Stratification of subjects based on Peak Knee Abduction Moment and comparison of peak knee translations across groups.

Group	Subject ID	Peak Abd. Moment (Nm)	MAX ATT (mm)	MAX LTT (mm)	MAX MTT (mm)
Low	15	2.53	4.33	0.32	-1.91
	1	4.02	3.93	1.28	-1.13
	2	5.40	5.19	0.15	-1.38
	13	6.84	9.22	0.08	-3.60
	7	7.69	3.31	-1.95	-3.38
Intermediate	5	14.11	3.77	-0.01	-1.55
	14	15.32	3.01	1.05	-2.25
	11	16.49	4.87	0.86	-3.99
	12	17.40	4.61	1.16	-1.25
	10	17.43	4.18	1.82	-3.28
High	3	17.59	6.35	-0.35	-3.04
	6	21.68	3.87	1.92	-1.37
	4	32.02	6.34	-0.69	-4.14
	8	33.41	3.05	0.39	-2.10
	9	37.16	2.97	-0.75	-1.54
Group (N = 15)					
Mean ($\pm$ 1SD)		16.6 (10.8)	4.6 (1.7)	0.3 (1.0)	-2.4 (1.1)
Low Knee Abd. Moment Group (n = 5)					
Mean ($\pm$ 1SD)		5.3 (2.1)	5.2 (2.4)	-0.02 (1.2)	-2.28 (1.14)
Intermediate Knee Abd. Moment Group (n = 5)					
Mean ($\pm$ 1SD)		16.2 (1.4)	4.1 (0.7)	1.0 (0.7)	-2.46 (1.16)
High Knee Abd. Moment Group (n = 5)					
Mean ($\pm$ 1SD)		28.4 (8.3)	4.5 (1.7)	0.1 (1.1)	-2.44 (1.15)
ANOVA		p < .0001	p = .614; β = .128	p = .265; β = .062	p = .964; β = .054
Post Hoc Comparisons (Bonferroni)					

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Group	Subject ID	Peak Abd. Moment (Nm)	MAX ATT (mm)	MAX LTT (mm)	MAX MTT (mm)
Low vs High Knee Abd. Moment Groups		p < .0001	na	na	na
Low vs Intermediate Knee Abd. Moment Groups		p < .0001	na	na	na
Intermediate vs High Knee Abd. Moment Groups		p = .012	na	na	na

Peak Abd. Moment = peak knee abductor moment measured by traditional motion analysis system

Max ATT = peak anterior tibial translation measured by bi-plane fluoroscopy

Max LTT and MTT = peak (+)lateral and (-)medial translation measured by bi-plane fluoroscopy

Bonferroni adjusted alpha for p-value is p = .016

na = no post hoc required

Table 3

Stratification of subjects based on Peak External Tibial Rotation Angle and comparison of peak knee translations across groups.

Group	Subject ID	Max Ext. Tib. Rotation (deg)	MAX ATT (mm)	MAX LTT (mm)	MAX MTT (mm)
Low	10	2.90	4.18	1.82	-3.28
	1	1.80	3.93	1.28	-1.13
	3	0.45	6.35	-0.35	-3.04
	15	0.40	4.33	0.32	-1.91
	13	0.09	9.22	0.08	-3.60
Intermediate	6	-0.20	3.87	1.92	-1.37
	12	-1.80	4.61	1.16	-1.25
	8	-4.20	3.05	0.39	-2.10
	4	-4.80	6.34	-0.69	-4.14
	5	-4.90	3.77	-0.01	-1.55
High	11	-5.48	4.87	0.86	-3.99
	7	-10.40	3.31	-1.95	-3.38
	14	-13.29	3.01	1.05	-2.25
	9	-19.65	2.97	-0.75	-1.54
	2	-20.00	5.19	0.15	-1.38
Group (N = 15)					
Mean ($\pm$ 1SD)		-5.2 (7.3)	4.6 (1.7)	0.3 (1.0)	-2.4 (1.1)
Low Ext. Tib. Rotation Group (n = 5)					
Mean ($\pm$ 1SD)		1.1 (1.3)	4.5 (1.0)	0.6 (0.9)	-2.6 (1.0)
Intermediate Ext. Tib. Rotation Group (n = 5)					
Mean ($\pm$ 1SD)		-3.2 (2.1)	5.0 (2.7)	0.6 (1.0)	-2.1 (1.2)
High Ext. Tib. Rotation Group (n = 5)					
Mean ($\pm$ 1SD)		-13.8 (6.2)	4.2 (1.0)	-0.1 (1.2)	-2.5 (1.1)
ANOVA		p < .0001	p = .773; β = .220	p = .492; β = .158	p = .755; β = .074
Post Hoc Comparisons (Bonferroni)					

Group	Subject ID	Max Ext. Tib. Rotation (deg)	MAX ATT (mm)	MAX LTT (mm)	MAX MTT (mm)
Low vs High Ext. Tib. Rotation Groups		$p < .0001$	na	na	na
Low vs Intermediate Ext. Tib. Rotation Groups		$p = .038$; $\beta = .892$	na	na	na
Intermediate vs High Ext. Tib. Rotation Groups		$p = .0003$	na	na	na

Max Ext Tib. Rotation (deg) = peak (–) external tibial rotation angle for each subject as measured by traditional motion analysis system

Max ATT = peak anterior tibial translation measured by bi-plane fluoroscopy

Max LTT and MTT = peak (+)lateral and (–)medial translation measured by bi-plane fluoroscopy

Bonferroni adjusted alpha for p-value is $p = .016$

na = no post hoc required

Table 4

Stratification of subjects based on Peak Internal Tibial Rotation Angle and comparison of peak knee translations across groups.

Group	Subject ID	Max Int Tib. Rotation (deg)	MAX ATT (mm)	MAX LTT (mm)	MAX MTT (mm)
Low	9	-3.10	2.97	-0.75	-1.54
	14	-2.30	3.01	1.05	-2.25
	11	-1.50	4.87	0.86	-3.99
	2	0.60	5.19	0.15	-1.38
	13	2.10	9.22	0.08	-3.60
Intermediate	12	2.30	4.61	1.16	-1.25
	15	4.55	4.33	0.32	-1.91
	8	7.10	3.05	0.39	-2.10
	4	7.80	6.34	-0.69	-4.14
	7	9.70	3.31	-1.95	-3.38
High	5	9.90	3.77	-0.01	-1.55
	6	12.20	3.87	1.92	-1.37
	10	13.76	4.18	1.82	-3.28
	3	16.30	6.35	-0.35	-3.04
	1	16.90	3.93	1.28	-1.13
Group (N = 15)					
Mean ($\pm$ 1SD)		6.2 (6.7)	4.6 (1.7)	0.3 (1.0)	-2.4 (1.1)
Low Int. Tib. Rotation Group (n = 5)					
Mean ($\pm$ 1SD)		-0.8 (2.1)	5.1 (2.5)	0.3 (0.7)	-2.6 (1.2)
Intermediate Int. Tib. Rotation Group (n = 5)					
Mean ($\pm$ 1SD)		6.3 (2.9)	4.3 (1.3)	-0.2 (1.2)	-2.6 (1.2)
High Int. Tib. Rotation Group (n = 5)					
Mean ($\pm$ 1SD)		13.8 (2.9)	4.4 (1.1)	0.9 (1.0)	-2.1 (1.0)
ANOVA		p < .0001	p = .781; β = .077	p = .236; β = .258	p = .723; β = .087
Post Hoc Comparisons (Bonferroni)					

Group	Subject ID	Max Int Tib. Rotation (deg)	MAX ATT (mm)	MAX LTT (mm)	MAX MTT (mm)
Low vs High Int. Tib. Rotation Groups		p < .001	na	na	na
Low vs Intermediate Int. Tib. Rotation Groups		p = .002	na	na	na
Intermediate vs High Int. Tib. Rotation Groups		p = .003	na	na	na

Max Int. Tib. Rotation (deg) = peak (+) internal tibial rotation angle for each subject as measured by traditional motion analysis system

Max ATT = peak anterior tibial translation measured by bi-plane fluoroscopy

Max LTT and MTT = peak (+)lateral and (-)medial translation measured by bi-plane fluoroscopy

Bonferroni adjusted alpha for p-value is p = .016

na = no post hoc required