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# Gender Differences in Lower Extremity Landing Mechanics Caused by Neuromuscular Fatigue

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**Background:** Neuromuscular fatigue has been suggested as an extrinsic factor in the mechanism of noncontact anterior cruciate ligament injury in both genders.

**Purpose:** To determine and describe the lower extremity kinematic and kinetic differences caused by neuromuscular fatigue during drop landings and compare changes between age- and skill-matched male and female athletes.

**Methods:** Inverse dynamic solutions estimated lower extremity flexion-extension and varus-valgus kinematics and kinetics for 14 female and 16 male athletes performing a single-legged 50-cm drop landing. Subjects performed landings prefatigue and postfatigue with fatigue induced via a parallel squat exercise (60% of 1 repetition maximum) until failure. A mixed-model, repeated-measures analysis of variance (fatigue * gender) was performed on select kinematic and kinetic variables.

**Results:** Neuromuscular fatigue caused men and women to land with more hip flexion (main effect fatigue, $P = .012$; main effect gender, $P = .001$). Men exhibited greater peak knee flexion angles postfatigue; women did not alter knee flexion (fatigue * gender, $P = .028$). Men exhibited larger peak knee varus angles irrespective of fatigue (main effect gender, $P = .039$; main effect fatigue, $P = .127$; fatigue * gender, $P = .153$); women demonstrated larger peak valgus angles overall (main effects gender, $P = .009$). There were no changes with fatigue (main effect fatigue, $P = .127$) or a different response due to fatigue with gender (fatigue * gender, $P = .091$). Women exhibited greater knee anterior shear force postfatigue (fatigue * gender, $P = .010$). Men and women exhibited lower knee extension moments (main effect fatigue, $P = .000$; main effect gender, $P = .927$; fatigue * gender, $P = .309$) and abduction moments (main effect fatigue, $P = .014$; main effect gender, $P = .670$; fatigue * gender, $P = .191$).

**Conclusion:** Neuromuscular fatigue caused significant alterations in women that may be indicative of the noncontact anterior cruciate ligament injury mechanisms.

**Clinical Relevance:** Current noncontact anterior cruciate ligament prevention programs should incorporate a fatigue component to help minimize the deleterious effects of neuromuscular fatigue on landing mechanics.

**Keywords:** anterior cruciate ligament (ACL); neuromuscular fatigue; knee injury; gender; biomechanics

A large proportion of ACL ruptures occur as a result of noncontact injuries.[57] Arendt et al[1,2] reported an increased risk of noncontact ACL injuries in female athletes over their male counterparts. Several intrinsic and extrinsic factors have been debated and linked to the noncontact ACL injury disparity between genders.[2,58] Biomechanical performance differences between men and women during cutting and landing have emerged as significant risk factors that can be reliably measured in a controlled laboratory environment.[‖] The findings of these studies have contributed to the development of performance-based ACL injury prevention training programs used to identify athletes who may be predisposed to ACL injury based on their execution of a landing technique.[18,19,24,25] The subsequent implementation of these training programs has shown some success in reducing noncontact ACL injury in women in competitive arenas.[18,24,25] Despite these efforts, however, the overall occurrence of the noncontact ACL injury remains one of the more common knee injuries in athletics, and both men and women are still rupturing their ACLs. This suggests that other factors, not incorporated into the current ACL injury prevention protocols, such as neuromuscular fatigue, may contribute to the occurrence of the noncontact ACL injury in both genders.

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Neuromuscular fatigue is a multifaceted phenomenon that can occur anywhere along the neuromuscular pathway. This may involve both central fatigue (defined as processes above the neuromuscular junction[17]) and peripheral fatigue (defined as mechanisms involving muscle and contractile elements[17]). Rahnama et al[50] showed that more "potential injury situations" occur in the final 15 minutes of a soccer game than at any point earlier during play. Gabbett[14-16] reported significantly more injuries during the second half of rugby matches as well. The injury rate also increases during the latter portion of the season. However, these studies did not differentiate between contact and noncontact injuries. Hawkins et al[22,23] have shown the greatest distribution of noncontact injuries occurs during the last 15 minutes of the first half and the final 30 minutes of the second half of regulation soccer matches. These studies also report that noncontact knee injuries account for up to 17% of these events. Several studies have also specifically investigated neuromuscular fatigue effects on human performance as it relates to the noncontact ACL injury mechanism.[8,38,46,51] The general consensus of these studies suggests neuromuscular fatigue can cause alterations in lower extremity landing and cutting mechanics similar to the characteristics previously proposed to increase the risk of ACL injury, particularly in female athletes. These alterations include varus-valgus knee angles, knee shear forces, proprioception, and neural activation.[8,46,51]

Despite strong epidemiological evidence, relatively few studies have implemented a neuromuscular fatigue test condition to determine gender performance differences in the mechanical factors that seem to predict noncontact ACL injury. The paucity of biomechanical performance data collected under the influence of fatigued physiological states potentially obscures the ability to link human performance to ACL injury under realistic, competitive situations. We propose that by understanding the effects of neuromuscular fatigue on human biomechanical and performance factors, we may improve the prognostic abilities of current ACL injury prevention programs aimed at reducing ACL injury in both men and women.

The purpose of this current research investigation is to determine and describe the kinematic and kinetic differences caused by neuromuscular fatigue on the ankle, knee, and hip during drop landings and to compare these changes between age- and skill-matched male and female recreational athletes. The null hypothesis stated is that the fatigue state and gender of subjects, as independent variables, will not affect landing kinetics or kinematics, as the dependent variables.

## METHODS

### Subjects

Fourteen female (mean age, 23.0 ± 0.9 years; mean height, 169.3 ± 7.6 cm; mean weight, 63.9 ± 9.4 kg) and 16 male (mean age, 23.8 ± 0.4 years; mean height, 177.7 ± 7.1 cm; mean weight, 79.4 ± 9.0 kg) recreational athletes from the La Crosse campus and community were recruited to participate in this study. Participants had no history of serious lower extremity injury and were recreationally participating in 1 or more sports, such as tennis, basketball, volleyball, and soccer, at least twice weekly. All subjects signed a written consent form approved by the university that is in accordance with the National Institutes of Health mandated guidelines. With a statistical significance set at a 2-sided level of .05 (probability, type I error), a power of .80 (ie, probability, type II error = .20), and the correlation coefficient between 2 points of time (repeated measurements based on fatigue state) set at .50 or more with effect size of .90, a minimum of 14 participants was required for each group.

### Testing Protocol

All subjects were instructed to refrain from any lower extremity weight training and/or exercise 48 to 72 hours before participation in this study. On arrival to the performance laboratory and before testing, all subjects performed a supervised stretching regimen and 10-minute warm-up on a motorized treadmill.

Subjects were instructed to perform a prequalifying parallel squat using a Smith machine with a self-selected weight that allowed the completion of 5 to 8 repetitions. Based on the results of this lift, a 1 repetition maximum (1 RM) was calculated using the weight and total number of repetitions performed. This method allowed the identification of maximum force production without the onset of neuromuscular fatigue.[36] Once the 1 RM was determined, 60% of that load was calculated and used as the testing weight during the fatigue protocol; 60% is considered a low load and was selected to minimize the risk of injury.[42] Subjects then performed as many repetitions as possible with the predetermined weight. The descent and ascent phase velocities of each lift were controlled using a prerecorded cadence of 3.2 seconds, with 1.7 seconds and 1.5 seconds corresponding to descent and ascent phases, respectively.[42] A "stop" was placed under the buttocks, as described by Isear et al,[30] to keep the thighs parallel to the ground at the lowest point in the descent phase, controlling for squat depth. At minimum, 4 sets of this fatigue cycle were performed by each subject, with 90 seconds of rest between each set. It has been reported that at least 4 sets of stressful resistance exercise training are necessary to show significant neuromuscular fatigue during latter sets.[34] The subjects were assumed to be fatigued when they had completed 4 or more sets and could no longer lift the weight.

Both before and immediately after performing the fatigue protocol, the subjects performed 6 drop landing trials by dropping from a stable hanging bar onto a force platform and landing on the dominant leg. Subjects were instructed to land as normally and as comfortably as possible without falling, losing balance, stepping off the plate, or touching the ground with either hand. Single-legged landings were performed in favor of a double-legged landing for 2 reasons: (1) Asymmetries in kinematics and kinetics often occur between legs in 2-legged landings,[47,52,53] and (2) most ACL injuries that occur during landing are single-legged landings.[48,56] The height of the bar for each subject was determined by having the subject hang from the bar with his or her body completely extended and feet flat in relation to the floor. The height of the bar related to a 50-cm distance measured from the bottom of the subject's feet to the floor.

## Data Collection

The 3D kinematics of each trial were captured by securing 18 retro-reflective, spherical markers (diameter, 25 mm) to the test limb of each subject in a standard Helen Hayes configuration (substituting regular markers for the wand markers) at anatomical landmarks reported previously.[31,32] With this marker configuration, the x, y, and z coordinate distances of the hip joint center from the anterior superior iliac spine were determined as a function of leg length and greater trochanter location.[7] The knee center was assumed to be in line with the plane defined by the thigh marker, the hip joint center, and the midpoint between the femoral condyles. The ankle joint center was assumed to fall in the plane defined by the estimated knee center, the tibial tuberosity, and the midpoint between the 2 malleoli markers.

All kinematic data were collected at 240 Hz using 6 Eagle cameras positioned at 60° intervals around the performance area. The cameras and subsequent performance area were calibrated, yielding mean residual errors of 1.1 to 1.53 mm over a volume of 1.5 × 1.1 × 1.5 m. The marker coordinate data were analyzed using Orthotrak (Motion Analysis Corporation, Santa Rosa, Calif) and custom Matlab programs (Mathworks Inc, Natick, Mass). Based on a frequency content analysis of the digitized coordinate data, marker trajectories were filtered at 10 Hz using a fourth-order Butterworth filter that retained 95% of the original signal content. Joint angular positions, velocities, and accelerations were calculated from the filtered 3D marker coordinate data using an Euler angle calculation with the assumption that flexion-extension was the first rotation, followed by abduction and internal-external rotation, respectively. Using the standing neutral trial as a reference, 0° at the hip, knee, and ankle corresponded to an erect, standing posture with the trunk, thigh, and lower leg in a straight line and the foot segment at a right angle to the leg when viewed from the sagittal plane. By this convention, the frontal plane varus and valgus kinematics were assigned positive and negative values, respectively.

The joint moments referred to in this article are internal joint moments, or moments applied from all the structures within and crossing the joint. Values for each joint moment were calculated by combining the kinematic and force plate data with anthropometric data[12] in an inverse dynamics solution.[31,32] By convention, hip and knee extensor and ankle plantarflexor moments were assigned positive values; varus and valgus knee joint moments were assigned positive and negative values, respectively. Thus, an external knee valgus moment would tend to produce a valgus knee rotation that would be resisted by an internal knee varus moment. All force values and all joint moment parameters were scaled to percentage body mass (%body mass) and newton meter/kilogram of body mass, respectively. The time series data sets were interpolated to 100 points during the impact phase (defined as the period from initial force platform contact to maximal knee flexion) for graphical purposes only.

## Data Analysis

A mixed-model, repeated-measures analysis of variance (fatigue * gender) was performed on select sagittal and frontal plane, hip, knee, and ankle kinematic and kinetic variables

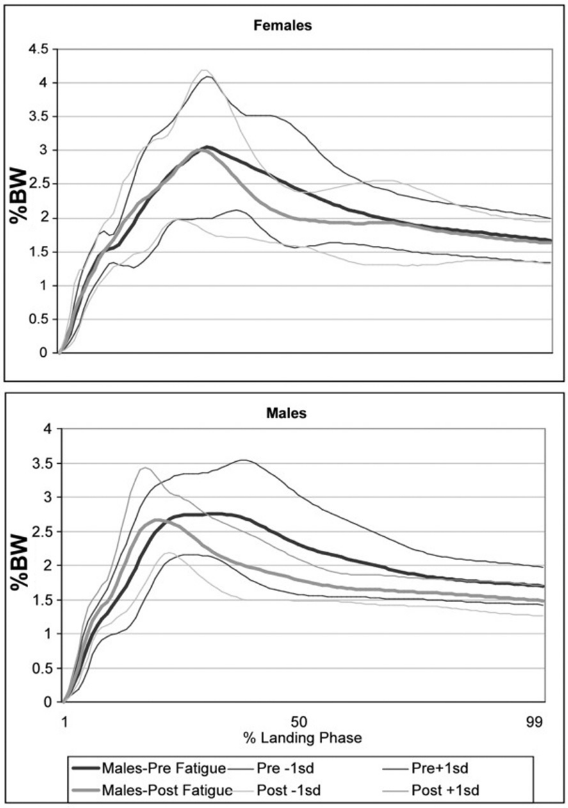


**Figure 1**. Male and female mean (±1 SD) ensemble vertical ground-reaction force profiles for the landing phase. BW, body weight.

using SPSS for Windows (version 12.0, SPSS Science Inc, Chicago, Ill) at an omnibus α level of .05. Dependent variables in this investigation were maxima and minima from kinematic and kinetic variables from initial contact until maximum knee flexion. From this, post hoc analyses using the Bonferroni technique were used to further examine the effects of neuromuscular fatigue on the ground-reaction forces, joint kinematics, and joint kinetics between genders.

# RESULTS

## Landing Style and Joint Kinematics

All subjects were visually observed to perform forefoot-to-rearfoot landings, and the group mean landing phase times were not different between groups (male group, 0.55 ± 0.19 seconds; female group, 0.52 ± 0.18 seconds; $P > .46$, power = .21). There was no main effect of fatigue (0.53 ± 0.11 seconds prefatigue; 0.51 ± 0.12 seconds postfatigue; $P = .39$, power = .51) or a fatigue * gender interaction ($P = .04$, power = .23).

## Ground-Reaction Force

Main effects for gender showed women exhibited an 8% larger peak vertical ground-reaction force (VGRF) normalized to body weight compared with men across both prefatigue and postfatigue conditions (main effects gender, $P = .002$) (Figure 1). Although the general trend was for a

TABLE 1
Mean, SDs, and *P* Values for Hip Kinematics and Kinetics Based on State of Fatigue and Gender

| Variable | Mean | ±1 SD | Test Within: Fatigue | Test Within: Fatigue * Gender | Test Between: Gender |
|---|---|---|---|---|---|
| Hip abductor, deg | | | | | |
| Maximum | | | .271 | .673 | .183 |
| Males | | | | | |
| Prefatigue | 1.22 | 6.95 | | | |
| Postfatigue | 3.36 | 5.60 | | | |
| Females | | | | | |
| Prefatigue | 6.44 | 4.92 | | | |
| Postfatigue | 3.75 | 6.52 | | | |
| Minimum | | | .196 | .389 | .365 |
| Males | | | | | |
| Prefatigue | –13.28 | 6.82 | | | |
| Postfatigue | –14.31 | 6.73 | | | |
| Females | | | | | |
| Prefatigue | –11.86 | 3.73 | | | |
| Postfatigue | –12.07 | 4.59 | | | |
| Hip flexion, deg | | | | | |
| Maximum | | | .012 | .633 | .001 |
| Males | | | | | |
| Prefatigue | 26.66 | 14.04 | | | |
| Postfatigue | 31.74 | 12.47 | | | |
| Females | | | | | |
| Prefatigue | 40.72 | 9.57 | | | |
| Postfatigue | 48.02 | 14.44 | | | |
| Minimum | | | .772 | .995 | .000 |
| Males | | | | | |
| Prefatigue | 0.62 | 7.81 | | | |
| Postfatigue | 0.28 | 9.09 | | | |
| Females | | | | | |
| Prefatigue | 12.28 | 5.83 | | | |
| Postfatigue | 11.95 | 7.79 | | | |
| Hip compression, deg | | | .000 | .057 | .905 |
| Males | | | | | |
| Prefatigue | 2.04 | 0.18 | | | |
| Postfatigue | 1.76 | 0.14 | | | |
| Females | | | | | |
| Prefatigue | 2.03 | 0.17 | | | |
| Postfatigue | 1.78 | 0.13 | | | |
| Hip shear, percentage body weight | | | .147 | .618 | .770 |
| Males | | | | | |
| Prefatigue | 0.07 | 0.13 | | | |
| Postfatigue | 0.05 | 0.03 | | | |
| Females | | | | | |
| Prefatigue | 0.10 | 0.16 | | | |
| Postfatigue | 0.04 | 0.02 | | | |
| Hip moments, N·m/kg – body weight | | | .019 | .367 | .453 |
| Males | | | | | |
| Prefatigue | 2.08 | 0.78 | | | |
| Postfatigue | 1.50 | 0.65 | | | |
| Females | | | | | |
| Prefatigue | 2.70 | 2.64 | | | |
| Postfatigue | 2.08 | 0.78 | | | |

reduction in the peak VGRF postfatigue, this trend was not significant (main effects fatigue, $P = .08$), and both genders responded similarly (fatigue * gender, $P = .276$).

## Joint Kinematics

Group means and SDs for the hip, knee, and ankle joint kinematics, prefatigue and postfatigue, are presented in Tables 1 through 3. Time series plots of VGRF and select hip, knee, and ankle kinematic and kinetic landing profiles, prefatigue and postfatigue, are presented in Figures 1 through 5.

In the sagittal plane, women landed with the hip in increased flexion and achieved maximum hip flexion angles that were 14° greater than the angles seen in men (Figure 2). Neuromuscular fatigue caused both men and women to increase hip flexion during landing compared

TABLE 2
Mean, SDs, and *P* Values for Knee Kinematics and Kinetics Based on State of Fatigue and Gender[a]

| Variable | Mean | ±SD | Test Within: Fatigue | Test Within: Fatigue * Gender | Test Between: Gender |
|---|---|---|---|---|---|
| Knee flexion, deg | | | | | |
| Maximum | | | .025 | .028 | .105 |
| Males | | | | | |
| Prefatigue | 67.24 | 11.79 | | | |
| Postfatigue | 73.81 | 10.85 | | | |
| Females | | | | | |
| Prefatigue | 64.19 | 10.48 | | | |
| Postfatigue | 64.27 | 10.48 | | | |
| Minimum | | | .989 | .891 | .472 |
| Males | | | | | |
| Prefatigue | 9.08 | 4.29 | | | |
| Postfatigue | 8.90 | 4.96 | | | |
| Females | | | | | |
| Prefatigue | 7.78 | 4.50 | | | |
| Postfatigue | 7.86 | 4.51 | | | |
| Knee JRF compression maximum, %BW | | | .000 | .209 | .074 |
| Males | | | | | |
| Prefatigue | 2.08 | 0.21 | | | |
| Postfatigue | 1.80 | 0.19 | | | |
| Females | | | | | |
| Prefatigue | 2.14 | 0.28 | | | |
| Postfatigue | 2.00 | 0.25 | | | |
| Knee JRF shear maximum, %BW | | | .000 | .010 | .282 |
| Males | | | | | |
| Prefatigue | 1.00 | 0.13 | | | |
| Postfatigue | 0.62 | 0.11 | | | |
| Females | | | | | |
| Prefatigue | 0.95 | 0.20 | | | |
| Postfatigue | 0.76 | 0.15 | | | |
| Knee extension moment maximum, N·m/kg of BW | | | .000 | .309 | .927 |
| Males | | | | | |
| Prefatigue | 1.39 | 0.33 | | | |
| Postfatigue | 1.13 | 0.31 | | | |
| Females | | | | | |
| Prefatigue | 1.44 | 0.64 | | | |
| Postfatigue | 1.05 | 0.37 | | | |
| Knee adduction (varus) maximum | | | .071 | .153 | .039 |
| Males | | | | | |
| Prefatigue | 8.51 | 4.50 | | | |
| Postfatigue | 8.00 | 5.35 | | | |
| Females | | | | | |
| Prefatigue | 3.86 | 5.15 | | | |
| Postfatigue | 2.55 | 6.36 | | | |
| Knee abduction (valgus) maximum | | | .127 | .095 | .009 |
| Males | | | | | |
| Prefatigue | –0.97 | 4.11 | | | |
| Postfatigue | –1.13 | 4.45 | | | |
| Females | | | | | |
| Prefatigue | –3.86 | 4.37 | | | |
| Postfatigue | –3.91 | 4.63 | | | |
| Knee abduction (varus) moment maximum | | | .014 | .191 | .670 |
| Males | | | | | |
| Prefatigue | 1.55 | 0.53 | | | |
| Postfatigue | 1.30 | 0.40 | | | |
| Females | | | | | |
| Prefatigue | 1.66 | 0.46 | | | |
| Postfatigue | 1.13 | 0.64 | | | |

[a]BW, body weight; JRF, joint reaction force.

TABLE 3
Mean, SDs, and *P* Values for Ankle Kinematics and Kinetics Based on State of Fatigue and Gender[a]

| Variable | Mean | ±SD | Test Within: Fatigue | Test Within: Fatigue * Gender | Test Between: Gender |
|---|---|---|---|---|---|
| VGRF maximum, %BW | | | .087 | .276 | .002 |
| Males | | | | | |
| Prefatigue | 3.52 | 0.41 | | | |
| Postfatigue | 3.05 | 0.48 | | | |
| Females | | | | | |
| Prefatigue | 3.84 | 0.65 | | | |
| Postfatigue | 3.74 | 0.81 | | | |
| Ankle PF maximum, deg | | | .452 | .190 | .452 |
| Males | | | | | |
| Prefatigue | –24.55 | 9.28 | | | |
| Postfatigue | –24.16 | 7.40 | | | |
| Females | | | | | |
| Prefatigue | –25.49 | 5.78 | | | |
| Postfatigue | –26.49 | 5.67 | | | |
| Ankle DF maximum, deg | | | .007 | .495 | .919 |
| Males | | | | | |
| Prefatigue | 24.25 | 8.03 | | | |
| Postfatigue | 25.73 | 8.58 | | | |
| Females | | | | | |
| Prefatigue | 23.55 | 4.65 | | | |
| Postfatigue | 25.96 | 5.08 | | | |

[a]BW, body weight; DF, dorsiflexion; PF, plantar flexion; VGRF, vertical ground-reaction force.

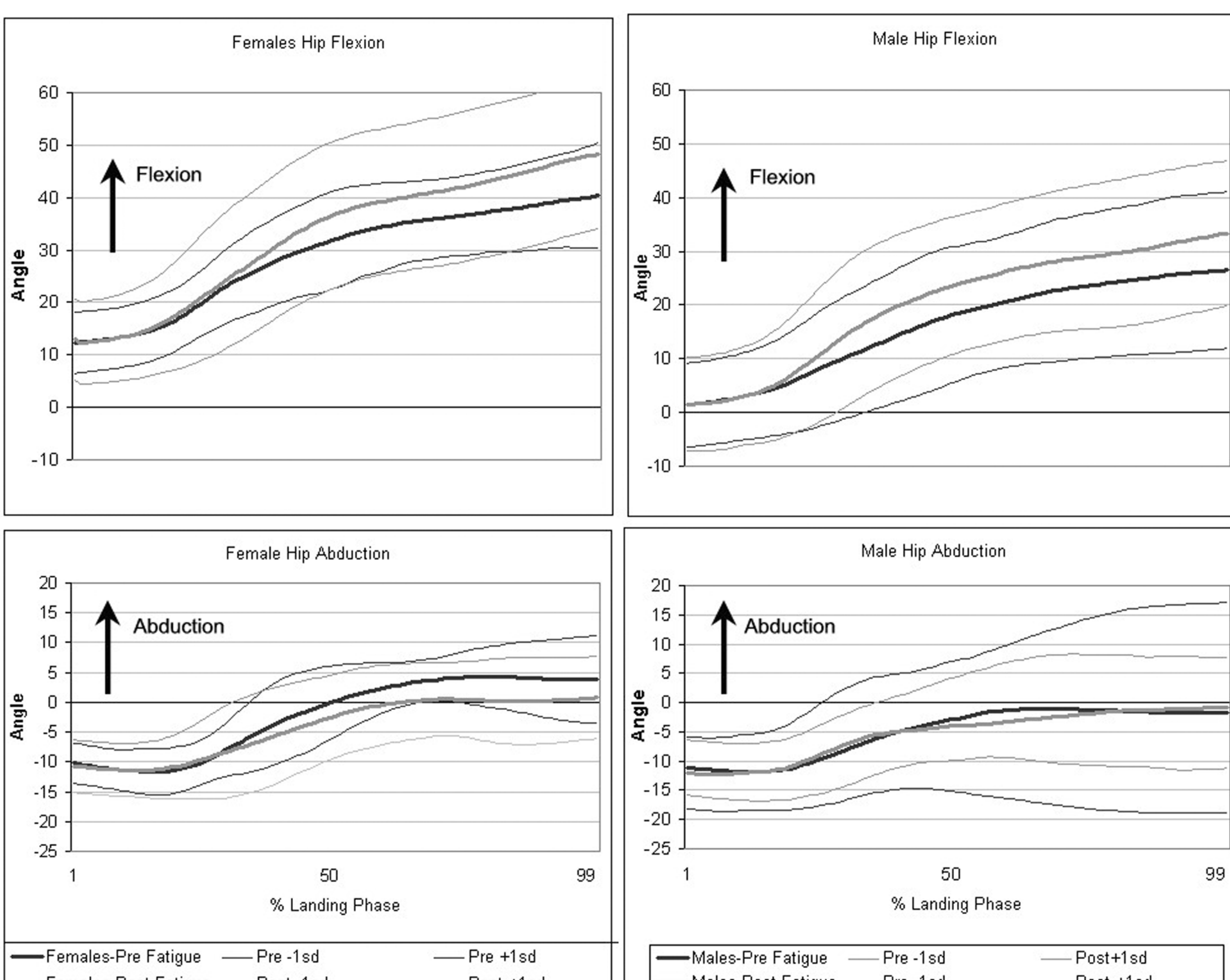


**Figure 2.** Male and female mean (±1 SD) ensemble hip flexion angle and hip abduction angle profiles over the landing phase.

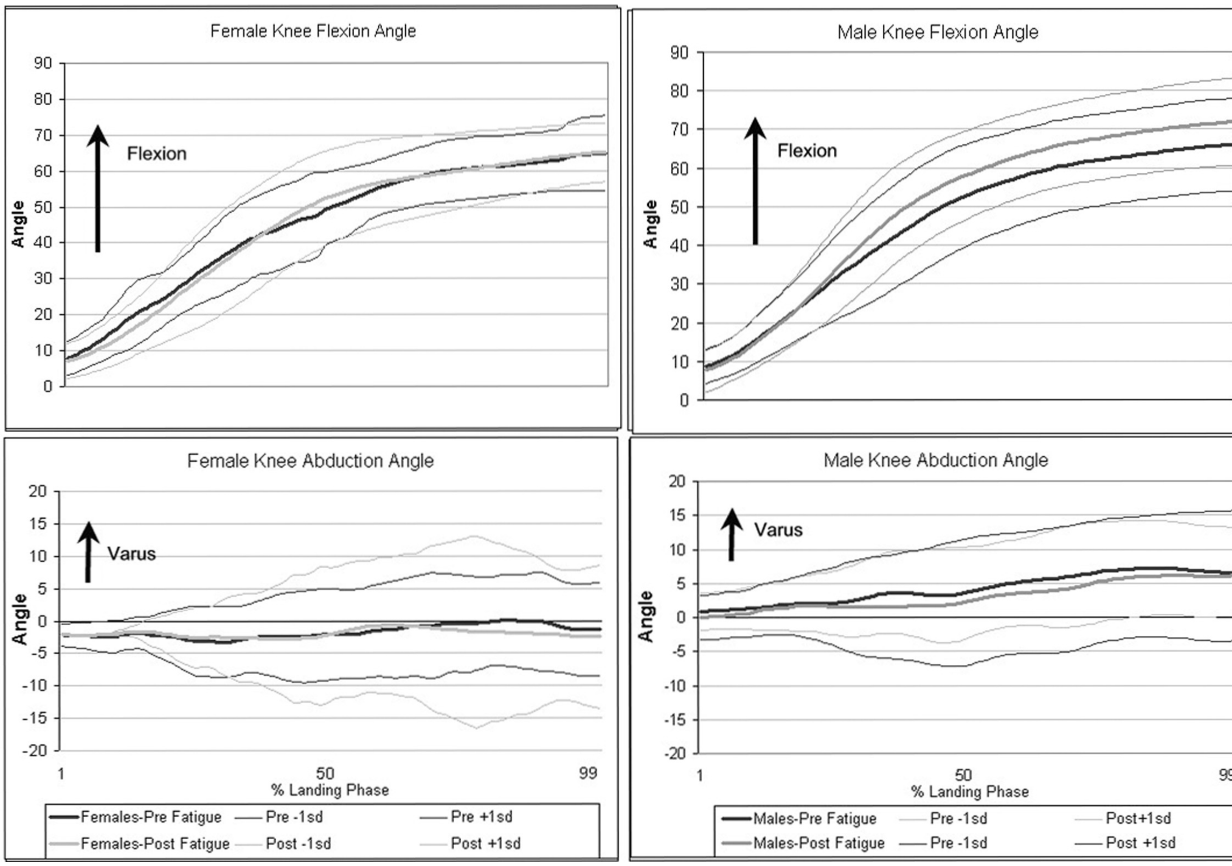


**Figure 3.** Male and female mean (±1 SD) ensemble knee flexion and knee valgus angle profiles over the landing phase.

with prefatigue trials (main effect fatigue, $P = .012$). However, women exhibited greater maximum hip flexion angles, by roughly 34% more than men, regardless of fatigue status (main effect gender, $P = .001$). There was no gender * fatigue interaction ($P = .66$). In the frontal plane, women tended to land with their hips in a more abducted position than did men during the prefatigue trials. Postfatigue, however, hip abduction angles decreased by 40%. Men showed the opposite trend and increased their hip abduction angles postfatigue. However, none of the frontal plane hip kinematic patterns (hip adductor/abductor minima or maxima) were significantly different (main effect fatigue, $P = .0271$; main effect gender, $P = .183$; fatigue * gender, $P = .673$).

At the knee in the sagittal plane, women tended to land with less knee flexion than did men and achieved less maximum knee flexion angles during the prefatigue trials (Figure 3). Neuromuscular fatigue caused a different performance response between men and women with regard to maximum knee flexion angle (fatigue * gender, $P = .028$). Men tended to increase their maximal knee flexion angles by about 14% (67.2° prefatigue vs 73.8° postfatigue), whereas women did not alter their knee flexion angles compared with the prefatigue condition (64.1° prefatigue vs 64.2° postfatigue). Men exhibited larger peak knee varus angles with and without fatigue (main effect gender, $P = .039$; main effect fatigue, $P = .071$; fatigue * gender, $P = .153$). Women demonstrated larger peak valgus angles overall, approximately 3.4° versus 1.0°, regardless of fatigue state compared to men (main effect gender, $P = .009$; main effect fatigue, $P = .127$). There was no gender * fatigue interaction (gender * fatigue, $P = .095$), indicating a similar fatigue response.

At the ankle, men and women performed similarly (Table 3). Both landed in a plantarflexed position (main effect gender, $P = .919$; main effect fatigue, $P = .452$; fatigue * gender, $P = .190$), and fatigue caused increases in the maximum ankle dorsiflexion in both groups (main effects fatigue, $P = .007$; fatigue * gender, $P = .495$).

## Joint Reaction Forces

At the hip, neuromuscular fatigue caused each gender to land with approximately 12.5% less hip compression (men, 13.7%; women, 12.3%; main effects fatigue, $P < .000$; main effects gender, $P = .905$; fatigue * gender, $P = .057$) and 48% less anterior hip shear force (men, 38.3%; women, 58.9%; main effects fatigue, $P = .147$; main effects gender, $P = .770$; fatigue * gender, $P = .618$) compared with prefatigue values.

At the knee, neuromuscular fatigue caused each gender to land with approximately 9% (men, 13.4% reduction; women, 6.5% reduction) lower peak compression force (main effects fatigue, $P = .000$) (Figure 4). There were no differences between genders (main effects gender, $P = .074$), and each gender responded similarly with a reduction in knee compression force due to the fatigue protocol (fatigue * gender, $P = .209$). Neuromuscular fatigue caused all subjects to adopt a landing

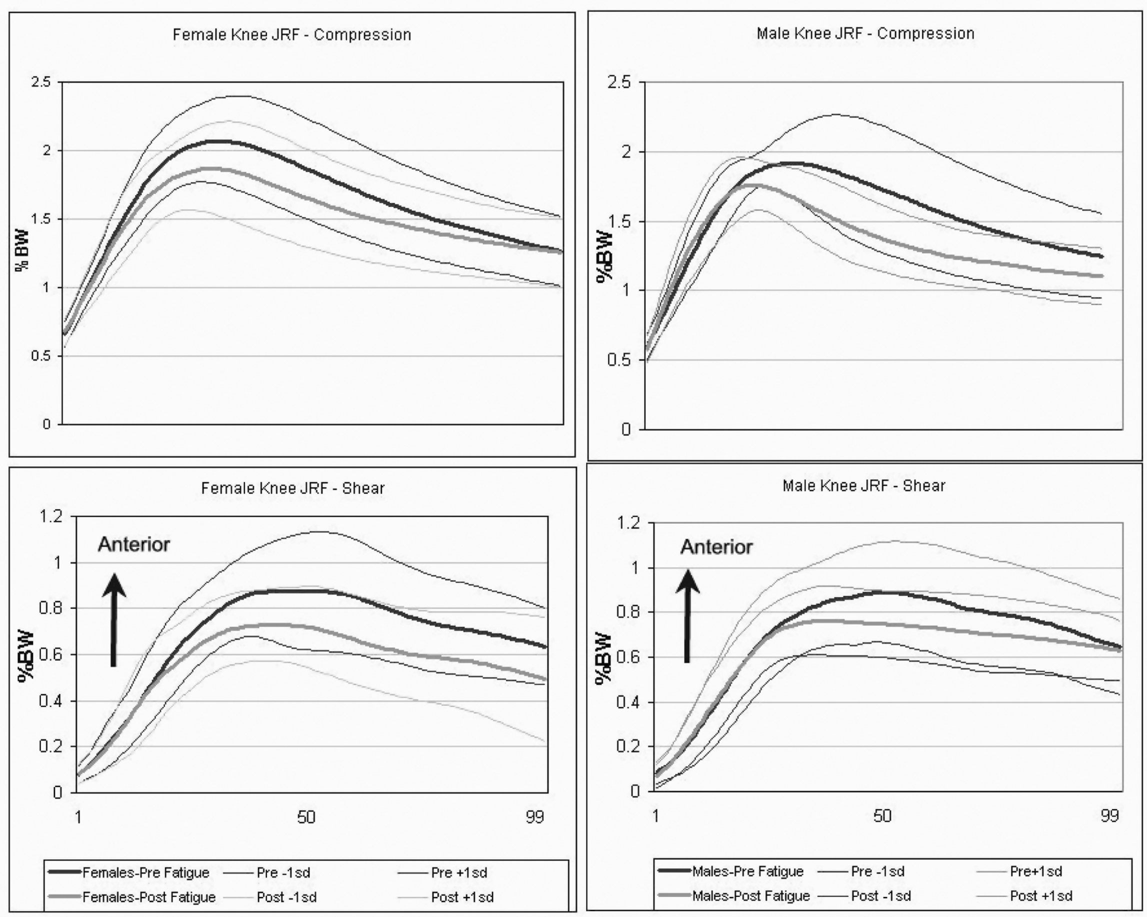


**Figure 4**. Male and female mean (±1 SD) ensemble knee joint compression (positive values) and knee joint shear (posterior shear forces = positive values) reactive force profiles over the landing phase. BW, body weight; JRF, joint reaction force.

style that effectively reduced the peak magnitude of the anterior knee shear force by a mean of 29%. Neuromuscular fatigue, however, affected the knee shear force pattern within each gender differently (fatigue * gender, $P = .010$). Men were able to reduce the magnitude of this force by 38% compared with the female group, who were only able to reduce this force by 20%.

## Joint Moments

At the hip, neuromuscular fatigue caused each gender to land with approximately 25% (men, 27.8%; women, 22.9%) less hip extensor moment (main effects fatigue, $P = .019$). There were no differences between genders (main effects gender, $P = .453$) as each gender responded similarly to the fatigue protocol with a reduction in this moment (fatigue * gender, $P = .367$) (Figure 5).

Neuromuscular fatigue caused each gender to land with approximately 22% less knee extensor moment (main effects fatigue, $P = .000$). There were no differences between genders (main effects gender, $P = .927$), and each gender responded similarly to the fatigue protocol with a reduction in this moment (fatigue * gender, $P = .309$).

In the frontal plane, both genders demonstrated a mean 25% reduction in the peak knee abduction moment postfatigue (main effects fatigue, $P = .014$). Overall, there was no difference between genders (main effects gender, $P = .670$), and both genders showed a similar rate of varus moment reduction (fatigue * gender, $P = .091$).

At the ankle, fatigue had no effect on the peak plantarflexion moment (main effect fatigue, $P = .452$; main effect gender, $P = .452$; fatigue * gender, $P = .190$). Fatigue did cause both groups to perform with a greater peak dorsiflexor moment (main effect fatigue, $P = .007$; main effect gender, $P = .919$; fatigue * gender, $P = .0495$).

## DISCUSSION

In 1984, Bigland-Ritchie[5] defined neuromuscular fatigue as "any reduction in the maximum force generating capacity, regardless of the force required in any given situation." Because muscles act as joint stabilizers in motions such as cutting[37] and landing,[49] deficiency in this utility due to central and/or peripheral neuromuscular fatigue may inhibit the body's ability to protect itself during dynamic movements. Numerous studies have shown that neuromuscular fatigue can reduce the force-generating capacity of muscle, as well as affect motor control and proprioception[55] and muscle reaction times.[20] Poor muscular conditioning can increase injury rates[29] and alter athletic performance during landing and stop-jump tasks.[8,38] The results of the current investigation support the previous studies and

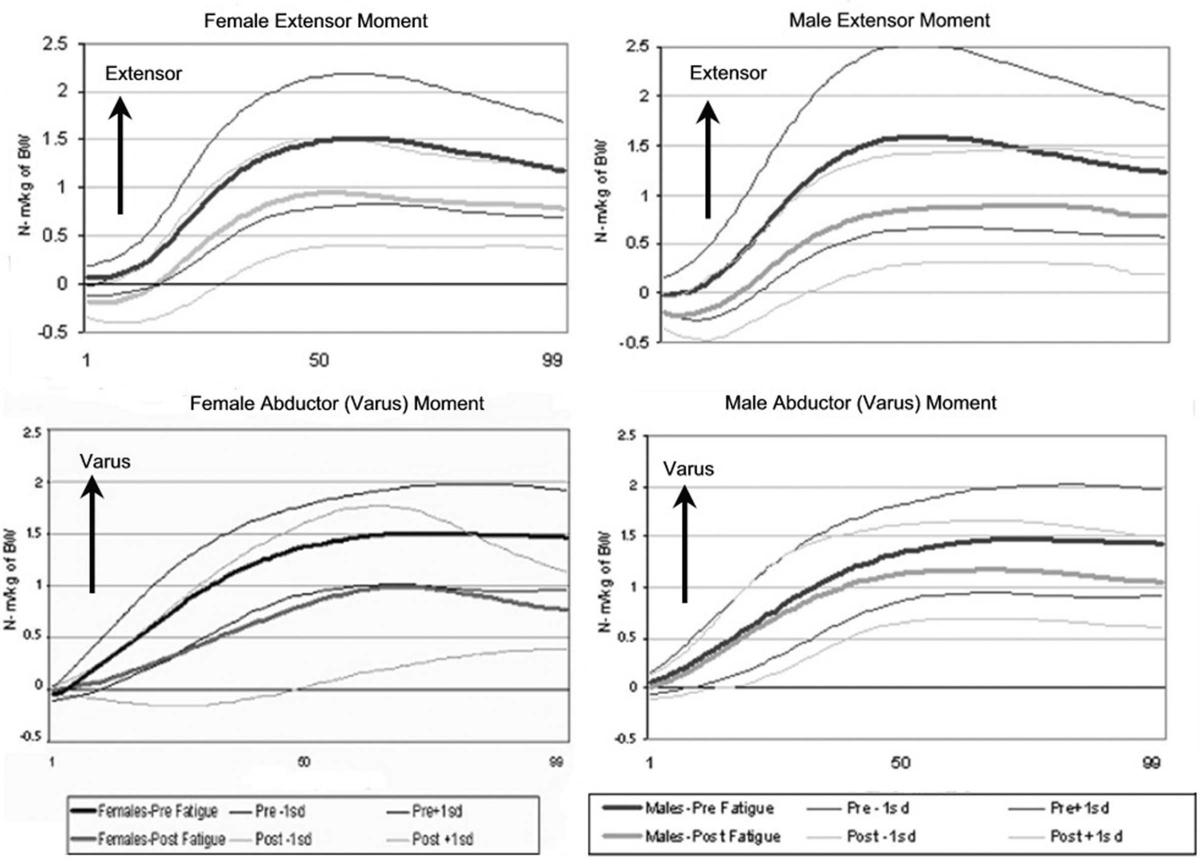


**Figure 5.** Male and female mean (±1 SD) ensemble knee extensor (positive values) knee abductor (positive values) moment profiles over the landing phase. BW, body weight.

suggest that landing characteristics for both male and female subjects are significantly altered owing to neuromuscular fatigue. The postfatigue landing characteristics of both men and women in the present study generally appear similar to landing profiles resembling the proposed noncontact ACL injury mechanisms[26] in which increased knee valgus loading contributes to ACL strain, as previous in vivo,[26] cadaveric,[40,41] and computer modeling[3,4] studies have demonstrated. In the present study, men exhibited greater knee varus, whereas women exhibited greater knee valgus. Neither of these measures appeared to be influenced by fatigue. Moreover, women showed a greater postfatigue effect in knee anterior shear joint reaction force and less knee flexion than did their male counterparts. In a previous investigation,[33] we showed that women did not generate similar internal knee varus moments to those of men at the time of peak knee valgus position. Combining these findings with the present investigation could suggest women may exhibit greater performance changes that increase the risk of noncontact ACL injury when neuromuscular fatigue is considered. Because current prevention programs do not employ fatigued states in their designs, the present study suggests that typical performances during prevention training sessions may change with fatigued states. Thus, identifying fatigue and the influences of fatigue on individual performances may improve the prognostic abilities of these prevention programs over the successful prevention trends currently being reported.[27]

In the sagittal plane, the present study noted men achieved greater knee flexion angles than did women during fatigued states. This is in disagreement with Decker et al,[10] who reported women (during nonfatigued states) exhibited greater knee flexion angles and suggested that this could serve to reduce ACL loads. The importance of sagittal plane knee motions lies in their relationship to anterior and posterior knee shear forces. A great deal of controversy exists over whether aberrant sagittal plane kinetics can be considered causative to ACL injury.[11,43,49,57,59] Chappell et al[8] noted increased anterior tibial shear force during stop jumps in association with increased knee flexion angles. In contrast, the present study noted that both men and women accommodated to the fatigued state in such a manner that reduced both maximum shear and compressive knee joint reaction forces. This is particularly evident for the anterior knee shear force, in which men were more effective in reducing the maximum anterior knee shear force produced than were women by achieving positions of greater knee flexion during fatigued states. Thus, although frontal plane knee kinematics suggest an increased risk of ACL injury, the reduction of anterior knee shear force suggests otherwise.

Neuromuscular fatigue caused an increase in the maximal ankle dorsiflexion angles observed. Landing in this manner may be 1 way individuals attempt to reduce knee loads under fatigued states. Increased dorsiflexion affords a greater capacity for energy absorption[10,54] and can minimize energy transfer to the knee.[54] Whether this kinematic adaptation can directly reduce ACL loads remains to be determined.

Previous reports have used varying types of neuromuscular fatigue protocols, including a knee flexion-extension isokinetic fatigue protocol,[51] a fatiguing landing protocol[38] consisting of a sequence of single-leg landing and squatting motions, and a neuromuscular fatigue exercise protocol[8] consisting of alternating vertical jumps and 30-m sprints (until verbal exhaustion). The 60% of 1 RM squat to exhaustion exercise was chosen in this fatigue protocol versus an open-chain, isokinetic, or aerobic exercise because squats are closed-chain activities with muscle activation patterns similar to those encountered during landings. Landings are characterized as short, energy-absorbing, eccentric activity to counteract the effects of gravity of the mass of a falling body. Squats incorporate the quadriceps and hamstrings, as well as hip extensors (ie, gluteals), hip adductors (ie, adductor longus), and plantarflexors (ie, gastrocnemius).[30,42] In comparison, isokinetic knee flexion/extension fatigue protocols only isolate 2 muscle groups: the quadriceps and the hamstrings. Squats also produce an anaerobic neuromuscular fatigue similar to that experienced after repeated landing and cutting. Aerobic fatigue or combinations of aerobic plus anaerobic fatigue protocols may disproportionately stress the circulatory and/or respiratory systems, rather than the muscles that will subsequently be used to absorb landing forces. However, it is important to note that across all studies using varying types of neuromuscular fatigue protocols,[8,38,46,51] the performance changes due to neuromuscular fatigue appear more similar than different. Thus, although arguments regarding the type of neuromuscular fatigue protocol used may be an important issue to consider in this type of research, it is possible that the type of neuromuscular fatigue protocol actually used may be irrelevant. Common performance characteristics like increased hip flexion,[8,51] reduced knee extensor moment,[8] increased knee valgus angle,[8] and larger VGRF[38,51] seem to emerge as a result of neuromuscular fatigue regardless of how fatigue (central, peripheral, or in combination) was achieved. Given this, we acknowledge differing fatigue protocols may produce different results. Because we do not currently understand "how much" fatigue or what combination of cardiovascular and neuromuscular fatigue may produce these seemingly common changes in lower extremity performance characteristics, we encourage future research in this area to focus on determining appropriate fatiguing protocols that are able to be controlled consistently among athletes of varying strength, stature, and skill level so that consistent methods may be established and thus readily applied to similar studies in the future.

Although the framework of this article assumes that the gender differences in knee valgus angle during landing are a performance-based occurrence, it is noted that these findings may also be owing to anatomical differences between genders. Women tend to possess a larger Q angle, or the relationship of femoral axis relative to the tibial axis measured at the knee in the frontal plane, than do men, and this is often associated with an increased knee valgus position. Although we did not measure subject-specific Q angles and cannot directly assess this association, no relationship between Q angle and ACL injury rates in female athletes has been found to date.[18,45] Nonetheless, noted gender differences in femoral anteversion, tibial torsion, and foot pronation[56,58] suggest future research is warranted in these areas and that female anthropometrics may be a confounding factor that cannot be discounted when attempting to understand gender-specific, in vivo, performance-based research.

Although in vivo human performance studies using the methods presented in the present article are considered traditional and acceptable techniques in estimating joint kinetics, it is acknowledged that (1) a skin-based marker system and the motion artifact associated with such a data collection scheme, the selected 10-Hz kinematic filter, and the absence of filtering the ground-reaction forces as proposed by Bisseling and Hof[6] may produce data that may not definitively reflect underlying bone translations and rotations; (2) these errors are further propagated by the estimation of joint forces and moments derived from the inverse dynamic approach; (3) the joint forces and moment profiles reported herein do not represent actual in vivo ACL loads; (4) this study investigated normal landing techniques within healthy male and female participants, and thus, injurious performances cannot be truly assessed; and (5) although we have made arguments that support and implicate the coupling of the peak knee varus-valgus knee moments and angles as potential determinants of ACL load, we are unaware of published data that have determined how or even when (during the land) an ACL rupture occurs. Thus, sophisticated models[43,44,49] are needed to deterministically define ACL load and the individual mechanical components that may contribute to its loading pattern.

The women in this study were recreational athletes. Other authors[2,21] have noted that skill level is a confounding factor that predisposes female athletes to ACL injury. In the present study, we have assumed that skill level, defined in terms of relevant exposure to landings during recreational sports and previous training and history of participation in sports in which landing is common, is not a cause of the observed differences presented. Although we cannot completely discount skill level as a factor that caused the apparent performance differences here, both groups in this study were selected on a voluntary basis and screened for recreational participation in sports that required repeated jumping and landing (eg, high school, varsity-level volleyball and basketball participation).

## CONCLUSION

Within the scope and limitations of the present study, the following conclusions are drawn: (1) Prefatigue, the majority of the differences in kinematic and kinetic variables between male and female recreational athletes during landing were observed in the frontal plane and not in the sagittal plane; (2) postfatigue, women were not able to reduce the magnitude of the anterior knee shear force as effectively as were men, and men effectively limited this force by a greater increase in knee flexion angle; and (3) this study details frontal and sagittal plane differences between men and women during the drop landings prefatigue and postfatigue, and because no one was injured during this investigation, we are limited in making inferences regarding joint injuries of any kind. Thus, it is quite plausible that women and men land differently owing to neuromuscular fatigue and that this performance difference is not a cause for injury in either group.

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