Effect of Fatigue on Knee Kinetics and Kinematics in Stop-Jump Tasks

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Background: Altered motor control strategies in landing and jumping maneuvers are a potential mechanism of noncontact anterior cruciate ligament injury. There are biomechanical differences between male and female athletes in the landing phase of stop-jump tasks. Fatigue is a risk factor in musculoskeletal injuries.

Hypothesis: Lower extremity muscle fatigue alters the knee kinetics and kinematics during the landing phase of 3 stop-jump tasks and increases an athlete's risk of anterior cruciate ligament injury.

Study Design: Controlled laboratory study.

Methods: Three-dimensional videography and force plate data were collected for 20 recreational athletes (10 male and 10 female athletes) performing 3 stop-jump tasks before and after completing a fatigue exercise. Knee joint angles and resultant forces and moments were calculated.

Results: Both male and female subjects had significantly increased peak proximal tibial anterior shear forces ($P = .01$), increased valgus moments ($P = .03$), and decreased knee flexion angles ($P = .03$) during landings of all 3 stop-jump tasks when fatigued. Fatigue did not significantly affect the peak knee extension moment for male or female athletes.

Conclusion: Fatigued recreational athletes demonstrate altered motor control strategies, which may increase anterior tibial shear force, strain on the anterior cruciate ligament, and risk of injury for both female and male subjects.

Clinic Relevance: Fatigued athletes may have an increased risk of noncontact anterior cruciate ligament injury.

Keywords: anterior cruciate ligament (ACL); sports; fatigue; injury; biomechanics

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Anterior cruciate ligament disruptions are common injuries in sports requiring cutting, pivoting, sudden stops, or landing from a jump, and the majority of the ACL injuries are noncontact in nature.[6,10,35] Multiple studies have shown an increased relative risk for noncontact ACL injuries for female athletes compared to male athletes.[1,6,15] Several intrinsic and extrinsic factors have been suggested as contributors to the increased risk in female athletes.[1,3] Altered motor control strategies in high-risk maneuvers are believed to be a potential injury mechanism. Biomechanical analysis demonstrated differences in the "landing phase of stop-jump tasks" with respect to gender that could increase an athlete's risk of injury.[8]

Fatigue is an extrinsic factor affecting the musculoskeletal and neurologic systems. Fatigue is associated with decreased knee proprioception and increased joint laxity compared to baseline values.[31,32] Muscle fibers have a decreased capacity to absorb energy when fatigued, and altered neuromuscular function with fatigue has been shown to increase anterior tibial translation.[23,36] Studies have investigated the effects of fatigue on ground reaction forces, lower extremity kinematics, and muscle activation in running, rapid stop tasks, and cross-cutting.[27,28] In running and rapid stop tasks, late onset of quadriceps and hamstring muscle activation and early occurrence of maximal knee flexion occurred with fatigue. These biomechanical changes are believed to decrease shock absorption and knee stabilization during landing. During cross-cutting tasks, quadriceps fatigue resulted in increased ankle dorsiflexion moments, decreased peak posterior breaking forces, decreased peak extension moments, and delayed peak knee flexion angles. Hamstring fatigue resulted in

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No potential conflict of interest declared.

The American Journal of Sports Medicine, Vol. 33, No. 7
DOI: 10.1177/0363546504273047

decreased peak impact knee flexion moment, increased internal tibial rotation, and decreased peak ankle dorsiflexion.[27]

It is still not clear, however, how fatigue contributes to noncontact ACL injuries. Previous studies have suggested that injuries are associated with poor conditioning,[19] and improved conditioning has been shown to reduce injury.[3,14] However, the association between fatigue and lower extremity injury, including ACL ruptures, remains anecdotal, and the relationship between fatigue and ACL loading in high-risk athletic tasks has not been explored. Recent studies suggest an association with ACL injury and stop-jump tasks,[6] and biomechanical analysis of these tasks showed increased peak proximal tibial anterior shear forces for female subjects compared to male subjects in the "landing phase."[8] The increased anterior shear forces indicate an increased tendency for greater anterior translation of the tibia and possibly increased ACL loading.

The purpose of this study was to determine the effects of lower extremity fatigue on the knee kinetics and kinematics of recreational athletes in 3 stop-jump tasks (Figure 1). We hypothesized that lower extremity muscle fatigue alters baseline knee kinetics and kinematics during the landing phase of stop-jump tasks. We hypothesized that fatigue would result in greater proximal tibial anterior shear forces, knee extension moments, and knee valgus moments and smaller knee flexion angles for both male and female subjects during the landings of the stop-jump tasks. In addition, we hypothesized that female subjects would have greater proximal tibial anterior shear forces, knee extension moments, and knee valgus moments and smaller knee flexion angles compared to male subjects during the landings of the stop-jump tasks under nonfatigued and fatigued conditions.

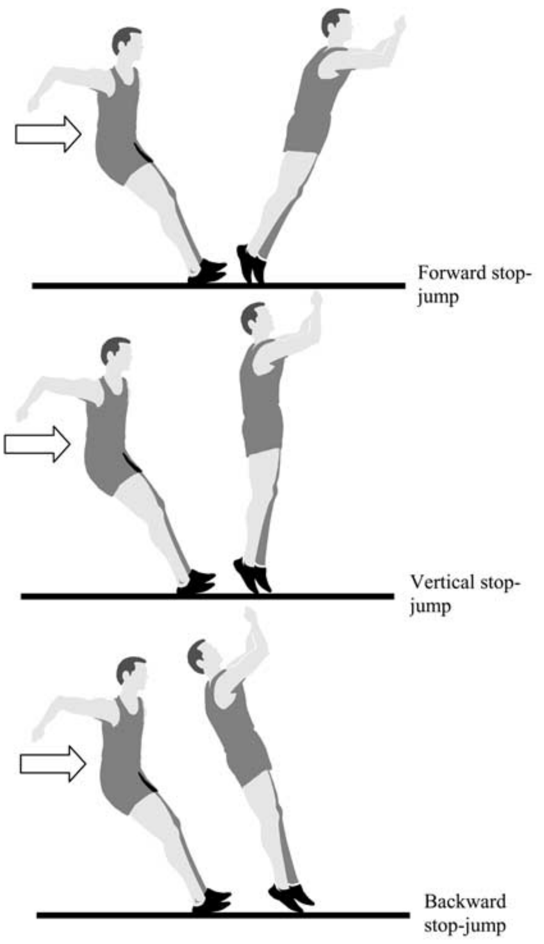

Figure 1. Three stop-jump tasks.

METHODS

Subjects

Twenty healthy recreational athletes (10 men and 10 women) were recruited as the subjects. A recreational athlete was defined as a person who competes in a sport such as basketball, soccer, or volleyball less than or equal to 3 times per week but does not follow a professionally designed training regimen. The mean age, body mass, and height were 23.7 ± 0.8 years, 71.4 ± 9.5 kg, and 1.79 ± 0.06 m, respectively, for male subjects and 21.7 ± 2.1 years, 55.6 ± 5.8 kg, and 1.66 ± 0.04 m, respectively, for female subjects.

Experimental Procedure

A 5-minute self-directed warm-up was allowed for each subject. The stop-jump tasks were described and demonstrated to the subject. Each of the 3 stop-jump tasks consisted of a 3-step approach run followed by a 1-footed takeoff, a 2-footed landing with each foot on a separate force plate, and a 2-footed takeoff for maximum height. The distance of the approach run was not restricted because of the intersubject variation in stride length, but the restriction of a 3-step approach run was strictly enforced. Each subject was allowed to practice each task until he or she felt comfortable performing the task. No instructions regarding jumping techniques were given to the subject to avoid a coaching effect on the subject's natural performances of these tasks.

The subjects wore a spandex outfit and personal socks and shoes. Eleven reflective markers were attached to the body surface at the right and left anterior-superior iliac spines, lateral thighs, lateral femoral epicondyles, lateral shanks, and lateral malleoli, as well as the spinous process of the fourth lumbar vertebrae. The lateral thigh markers were placed in line with the lateral condyles and the greater trochanters. The lateral shank markers were placed in line with the lateral malleoli and lateral condyles at the point of greatest curvature of the shank.

Each subject performed 5 trials for each stop-jump task in a prefatigue exercise test and immediately after completion of the fatigue protocol for the postfatigue exercise test. The order of the 3 tasks was randomized for each subject.

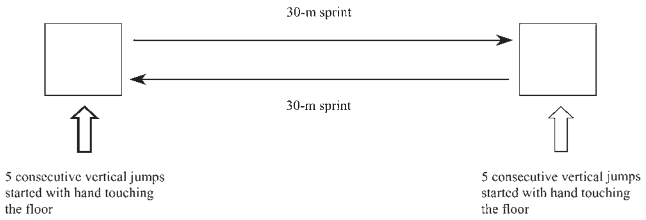

Figure 2. Setup of the fatigue protocol.

Fatigue Protocol

A customized fatigue exercise was used to create the fatigue state for the subjects. This fatigue exercise consisted of unlimited repetitions of 5 consecutive vertical jumps followed by a 30-m sprint (Figure 2). The subject was instructed to start each vertical jump from a squat position and jump to reach a target height of 115% of the subject's vertical reach (Figure 3). The subject was also instructed to accelerate and decelerate as quickly as possible during the 30-m sprint. The fatigue exercise was continued until the subject reached a state of volitional exhaustion. To maintain the fatigued state, the subject was instructed to perform 5 consecutive vertical jumps as done previously after every 5 stop-jump trials during the postfatigue exercise test.

Data Collection

Four video cameras were used at a frame rate of 180 frames/s with a setup for the direct linear transformation procedure (Peak Performance, Englewood, Colo). Two Bertec 4060A force plates (Bertec Corporation, Worthington, Ohio) collected ground reaction forces at a sampling rate of 540 samples/s. The video cameras and force plates were time synchronized using a Peak Motus data acquisition system (Peak Performance). Video cameras were calibrated using a 25-point calibration frame.

Data Reduction

The recorded video images were digitized using the Peak Motus video analysis system (Peak Performance), and 3-dimensional (3D) coordinates of the 13 reflection markers were estimated from the digitized 2-dimensional trajectories of the markers using an MSDLT computer program package (MotionSoft, Chapel Hill, NC). The 3D coordinates of the hip, knee, and ankle joint centers were estimated from the 3D coordinates of the 13 reflection markers.[20] Raw 3D coordinates were filtered through a low-pass Butterworth digital filter with an estimated optimum cutoff frequency.[37]

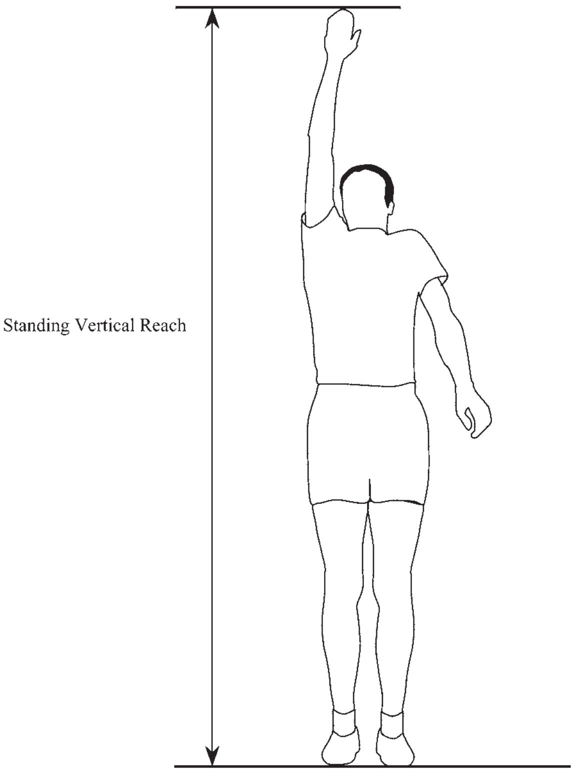

Figure 3. Standing vertical reach.

Jump height was defined as the maximum mean vertical coordinates of the right and left hip joint centers relative to the floor. Knee joint angles were determined using an MSKIME computer program package (MotionSoft) in which knee joint angles were customarily defined as 3 Euler angles[7,13] with flexion-extension, valgus-varus, and internal-external rotation as the first, second, and third rotations, respectively. Ground reaction force, free moment, and the location of the center of pressure were determined from force plate data records using an MSF-PLT computer program package (MotionSoft). Joint resultant forces and moments of the knee on the tibia were estimated using an inverse dynamic procedure[13] instrumented in an MSKINE computer program package (MotionSoft). The modified Clauser segment inertia parameters[17] were used to determine segment masses, moments of inertia, and locations of the segment center of mass. Segment Euler parameters were determined and used to estimate the segment angular velocities and accelerations.[16]

The estimated joint resultant forces and moments were normalized to the subject's body weight (BW) and the product of the subject's body weight and height (BW × BH), respectively. The duration from the landing to the takeoff was defined as the stance phase. A full stance phase was divided into 100 time intervals, and each time interval was normalized as 1% of the full stance phase. The landing phase was approximately 20% of the entire stance phase characterized with a maximum vertical ground reaction force because of the impact between the subject's feet and the force plates (Figure 4). The knee flexion-extension moment, knee valgus-varus moment, and knee flexion angle at the peak proximal tibial anterior shear force (Figure 5) were also identified.

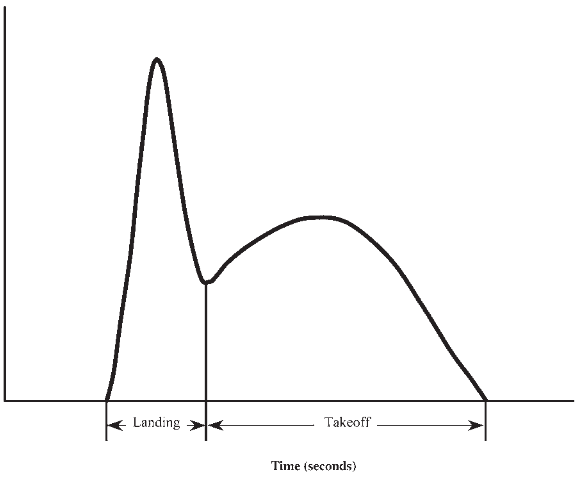

Figure 4. Vertical ground reaction force during a stop-jump task (N). Landing and takeoff phases during a stop-jump task identified from vertical ground reaction force.

Data Analysis

The jumping height, peak proximal tibial anterior shear force, knee flexion-extension moment, knee valgus-varus moment, and knee flexion angle were dependent variables, whereas fatigue state, gender, and task were independent variables in the data analysis. Fatigue state and task were treated as repeated measures, whereas gender was treated as an independent factor. A 3-factor analysis of variance with a mixed design was conducted for each dependent variable. A .05 type I error rate was chosen to indicate statistical significance. Post hoc paired *t* tests were performed to compare each dependent variable between tasks. The Bonferroni procedure was used to adjust the actual critical *P* values in post hoc *t* tests.

RESULTS

Jumping Height

The jumping height, as indicated by the mean maximum height of the right and left hips, was significantly decreased for both female and male subjects in the postfatigue exercise test in comparison to that in the prefatigue exercise test (*P* = .01). The fatigue effect on the jumping height was consistent across stop-jump tasks and genders (Figure 6). The mean decrease in the jumping height after fatigue was 0.05 m.

Peak Proximal Tibial Anterior Shear Force

The peak proximal tibial anterior shear force during the landing phase of the 3 stop-jump tasks in the postfatigue exercise test was significantly increased in comparison to that in the prefatigue exercise test (*P* = .01). This fatigue effect on the peak proximal tibial anterior shear force during landing was consistent across tasks and genders (Figure 7). The mean peak proximal tibial anterior shear force for all subjects in 3 stop-jump tasks increased from 0.24 times the body weight for the 3 stop-jump tasks in the prefatigue exercise test to 0.29 times the body weight in the postfatigue exercise test. This result represents a mean 21% increase in the peak proximal tibial anterior shear force.

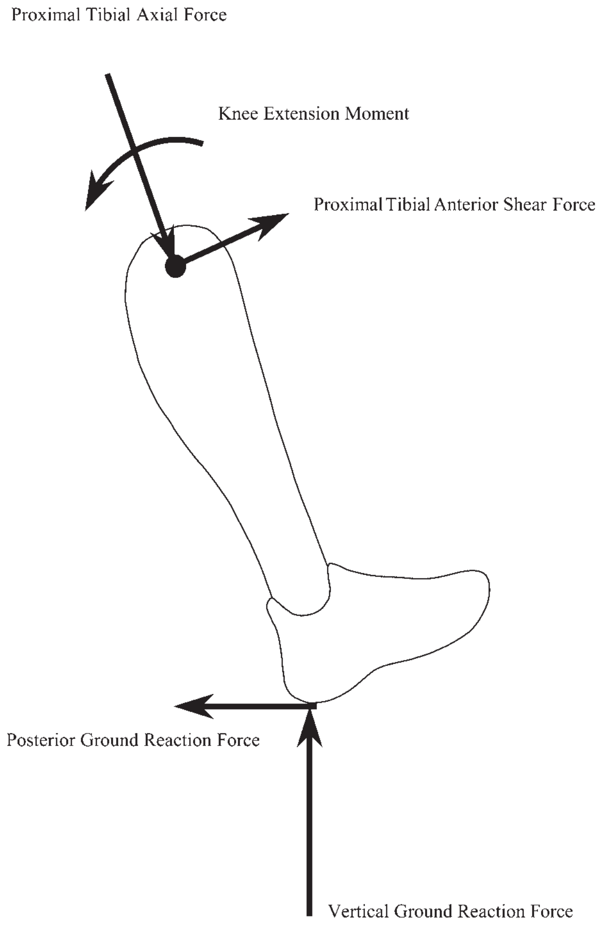

Figure 5. Proximal tibial anterior shear force and other forces and moments on the lower leg in the sagittal plane.

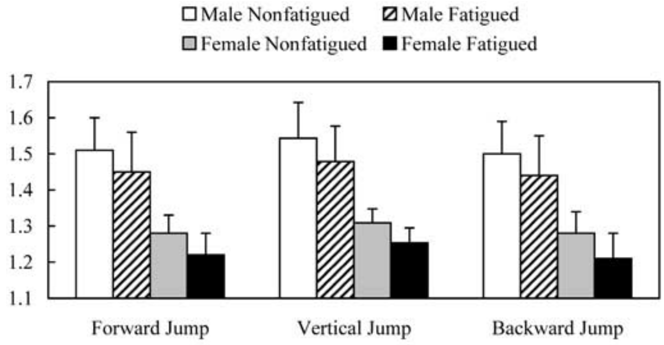

Figure 6. Maximum flight heights of the hip joints (m) used as a measure of jumping heights in 3 stop-jump tasks before and after fatigue exercises.

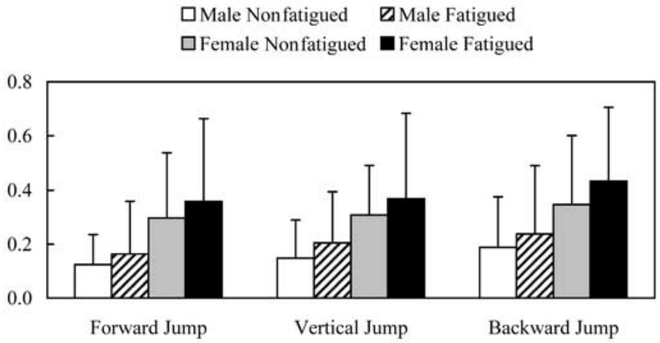

Figure 7. A comparison of the peak proximal tibial anterior shear force (body weight) between fatigue states, genders, and tasks.

Task also significantly affected the peak proximal tibial anterior shear force ($P = .01$). This task effect on the peak proximal tibial anterior shear force was consistent across fatigue states and genders (Figure 7). The mean peak proximal tibial anterior shear forces of all subjects in both prefatigue and postfatigue exercise tests were 0.23 times the body weight in the stop-jump forward task, 0.26 times the body weight in the stop-jump vertical task, and 0.30 times the body weight in the stop-jump backward task.

Female subjects had significantly greater peak proximal tibial anterior shear force than did male subjects ($P = .001$). This gender effect on the peak proximal tibial anterior shear force was consistent across fatigue states and tasks (Figure 7). The mean proximal tibial anterior shear force was 0.35 times the body weight for female subjects in 2 tasks and 2 fatigue states and 0.18 times the body weight for male subjects. This result represents a mean 94% increase in the peak proximal tibial anterior shear force for female subjects in comparison to male subjects.

Knee Extension Moment

Female subjects also had significantly greater knee extension moments at the peak proximal tibial anterior shear force than did male subjects ($P = .001$). This gender effect on the knee extension moment at the peak proximal tibial anterior shear force was consistent across fatigue states and tasks (Figure 8). The mean knee extension moment was 0.123 times the product of body height and weight for female subjects and 0.024 times the product of body height and weight for male subjects. Fatigue and task did not significantly affect the knee extension moment at the peak proximal tibial anterior shear force.

Knee Valgus-Varus Moment

During the landing phase of the stop-jump tasks, the knee valgus-varus moment at the peak proximal tibial anterior shear force in the postfatigue exercise test was significantly increased in comparison to that in the prefatigue exercise test ($P = .03$). This fatigue effect on the knee valgus-varus moment was consistent across stop-jump tasks and genders (Figure 9). The mean knee valgus moment at the peak proximal tibial anterior shear force of female subjects increased from 0.026 times the product of body height and weight in the prefatigue exercise test to 0.051 times the product of body height and weight in the postfatigue exercise test. This result represents a mean increase of 96% in the knee valgus moment for female subjects. The mean knee varus moment at the peak proximal tibial anterior shear force of male subjects decreased from 0.030 times the product of body height and weight in the prefatigue exercise test to 0.017 times the product of body height and weight in the postfatigue exercise test. This result represents a mean decrease of 43% in the knee varus moment for male subjects.

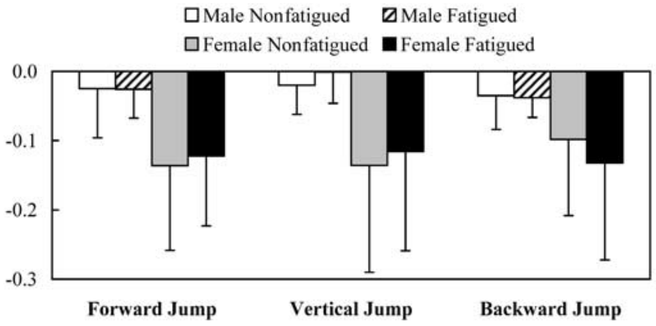

Figure 8. A comparison of the knee extension(–)-flexion(+) moment on the tibia at the peak proximal tibial anterior shear force (body height × body weight) between fatigue states, genders, and tasks.

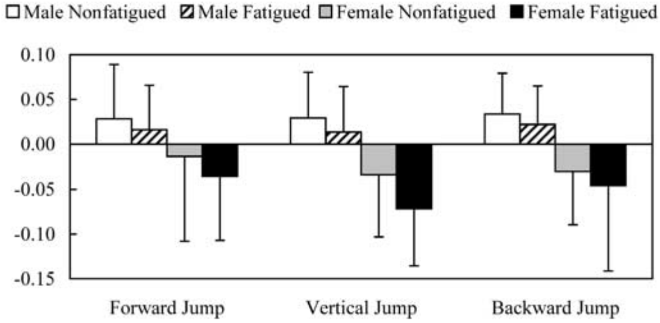

Figure 9. A comparison of the knee valgus(–)-varus(+) moment on the tibia at the peak proximal tibial anterior shear force (body height × body weight) between fatigue states, genders, and tasks.

There was a significant difference in the knee valgus-varus moment at the peak proximal tibial anterior shear force between female and male subjects ($P = .001$). Female subjects, on average, had a knee valgus moment at the peak proximal tibial anterior shear force, whereas male subjects, on average, had a knee varus moment (Figure 9). This gender difference in the knee valgus-varus moment was consistent across fatigue states and tasks.

Knee Flexion Angle

The knee flexion angle at the peak proximal tibial anterior shear force in the postfatigue exercise test was signifi-

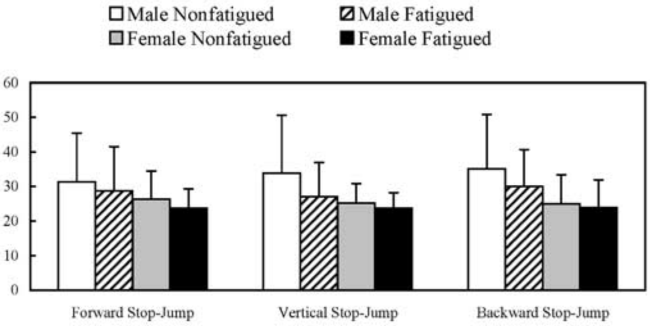

Figure 10. A comparison of the knee flexion angle (degrees) at the peak proximal tibial anterior shear force between fatigue states, genders, and tasks.

cantly decreased in comparison to that in the prefatigue exercise test (*P* = .03). This fatigue effect on the knee flexion angle was consistent across stop-jump tasks and genders (Figure 10). The knee flexion angle at the peak proximal tibial anterior shear force for all subjects decreased from a mean of 29.9° for the 3 stop-jump tasks in the prefatigue exercise test to a mean of 25.7° in the postfatigue exercise test. This result represents a mean decrease of 14% in the knee flexion angle after the fatigue exercise.

Female subjects had significantly smaller knee flexion angles at the peak proximal tibial anterior shear force than did male subjects (*P* = .001). This gender effect on the peak proximal tibial anterior shear force was consistent across fatigue states and tasks (Figure 10). The knee flexion angles at the peak proximal tibial anterior shear force were 26.3° in the prefatigue exercise test and 23.1° in the postfatigue exercise test for female subjects, compared to 33.4° in the prefatigue exercise test and 28.3° in the postfatigue exercise test for male subjects. Therefore, female subjects decreased their knee flexion angles in 3 stop-jump tasks by 12%, whereas male subjects decreased their knee flexion angles by 15% in the postfatigue exercise test.

DISCUSSION

The purpose of this study was to investigate the effects of lower extremity muscle fatigue on knee kinematics and kinetics in stop-jump tasks and attempt to provide biomechanical evidence that fatigue may be a risk factor for noncontact ACL injuries. The study used an independently designed fatigue protocol to induce lower extremity muscle fatigue and maintain the fatigued state for postfatigue exercise tests. This fatigue exercise was unique because it simulated the aerobic and anaerobic fatigue that basketball, soccer, and volleyball players experience during actual competitions. A multitude of fatigue protocols have been described in prior studies. Many generated localized fatigue of quadriceps and hamstring muscle groups through repetitions of isometric or eccentric contractions.[26,28] Other studies have focused on whole-body fatigue through simulated games with sprints, periods of rest, and prolonged low levels of activity.[9,17,22]

The fatigue protocol used for this study was designed to induce volitional exhaustion with general aerobic fatigue from sprints and localized lower extremity muscle fatigue with repetitive squat jumps. The decreased performance in the task of interest is commonly used as an indicator of fatigue because of the difficulty in directly measuring volitional exhaustion.[4,22,34] The significant decrease in jumping height for all subjects and tasks in the postfatigue test in this study suggested that the fatigue protocol in this study effectively created the desired lower extremity fatigue. In addition, the decreased jumping heights in the postfatigue exercise test suggested that the fatigue protocol maintained the state of lower extremity fatigue. The absence of an effect on jumping height with respect to gender in the postfatigue test suggested that male and female subjects were equally fatigued. The knee kinematics and kinetics in the nonfatigued state in this study are comparable to those in a previous study.[8]

The results of this study supported our hypotheses that lower extremity fatigue is associated with increased peak proximal tibial anterior shear force and decreased knee flexion angle. The results of this study also supported our hypothesis that fatigue is associated with an increased knee valgus moment for female subjects but not for male subjects. The results of this study, however, did not support our hypothesis that lower extremity fatigue increases knee extension moment.

Lower extremity muscle fatigue may increase an athlete's risk for noncontact ACL injury. The literature has repeatedly shown increased injury rates during the later portion of games in a variety of sports,[11,12,24,29,30] which indicates that fatigue is a risk factor for injuries. The results of this study showed that the peak anterior shear force on the proximal tibia was significantly increased for both male and female subjects in the fatigued state. Previous studies have suggested that proximal tibial anterior shear force may be an indication of increased strain in the ACL.[18,19,23] Therefore, the results of this study suggested that ACL strain may be increased when stop-jump tasks are performed with lower extremity fatigue. Repeated ACL loading with increased strain may further increase the risk for ACL injury. However, it should be noted that the postfatigue tests were performed in a state of volitional exhaustion, and the degree of the lower extremity fatigue created by the fatigue protocol in this study may be greater than the level of fatigue reached in athletic competition.

The results also suggested a gender effect, with lower extremity muscle fatigue increasing the risk of ACL injury more for women than for men. Previous studies repeatedly have shown that women have an increased risk for ACL injuries.[1,6,9] Recent research[8] and the present study demonstrated that female subjects, on average, have a significantly greater peak proximal tibial anterior shear force during the landing phase of stop-jump tasks in comparison to men. The present study further showed that female subjects have an even greater anterior shear force on the proximal tibia in the fatigued state compared to the nonfatigued state. Although male subjects also had increased anterior shear force on the proximal tibia in the fatigued state, this force magnitude was still lower than that of female subjects in the nonfatigued state. These results combined together, to a certain degree, support the theory that the altered movement patterns are risk factors for noncontact ACL injuries.

The cause of the increased peak anterior shear force on the proximal tibia with fatigue may be different for male and female subjects. Both male and female subjects showed smaller knee flexion angles in the postfatigue tests, but different changes in knee valgus-varus moments were observed for male and female subjects. The male subjects showed smaller knee varus moments, whereas the female subjects showed increased knee valgus moments. Previous studies have suggested that a decreased knee flexion angle tends to increase the stress on the ACL.[26,33] As the knee flexion angle decreases, the anterior shear force applied to the tibia by the quadriceps muscle increases as a result of the increased patellar tendon–tibia angle. The posterior shear force applied to the tibia by the hamstring muscle, however, decreases as a result of the decreased hamstring-tibia angle. Anterior tibial translation increases as a result of increased anterior shear force and decreased posterior shear force applied to the tibia. Previous studies also have suggested that an increased knee valgus or varus moment tends to increase the stress on the ACL.[2,5,21] Although both male and female subjects tended to have an increase in valgus moment with fatigue, the male subjects' changes had a protective effect with fatigue because the knee is brought into a more neutral position. These results suggested that the increased peak proximal tibial anterior shear force due to lower extremity muscle fatigue may be mainly because of a decreased knee flexion angle for male subjects but a combined effect of decreased knee flexion angle and increased knee valgus moment for female subjects.

Women and men may have different lower extremity motor control strategies in stop-jump tasks, and these differences may be responsible for the increased risk of noncontact ACL injuries in women. The results of this study showed that female subjects had a greater knee extension moment than did men when the peak proximal tibial anterior shear force occurred. The resultant knee moment in this study, especially the knee extension-flexion moment, was pure torque at the knee, mainly owing to muscle contractions. An increased knee extension moment indicates (1) increased quadriceps muscle contraction, (2) decreased hamstring muscle contraction, or (3) a combination of both conditions. Women have increased quadriceps activities and decreased hamstring activities in running, cutting, and jumping tasks in comparison to men.[25] A combination of increased quadriceps muscle contraction and decreased hamstring muscle contraction is likely to be an explanation for the increased knee extension moment for women in comparison to men. The difference in the quadriceps and hamstring contraction pattern is one of the major contributors to the difference in the peak proximal tibial anterior shear force between men and women.[6] The results of this study also showed that women, on average, had a knee valgus moment, whereas men had a varus moment, when the peak proximal tibial anterior shear force occurred. The cause of the difference in the knee valgus-varus moment between men and women and the subsequent mechanical advantages of these moments in terms of the performance of athletic tasks still remains unclear.

In conclusion, lower extremity muscle fatigue significantly increased the peak anterior shear force on the proximal tibia of recreational athletes, especially female athletes, performing stop-jump tasks. The increased anterior shear force indicates a possible increased strain on the ACL and thus an increased risk for ACL injury. The greater peak anterior shear force on the proximal tibia due to fatigue is associated with decreased knee flexion angle and increased valgus moment. Female and male recreational athletes have different lower extremity motor control strategies when performing the 3 stop-jump tasks. This difference in the lower extremity control strategies may significantly contribute to the increased risk for ACL injuries in women. Further studies are needed to establish the association between lower extremity motor control strategies and the risk for ACL injuries.

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