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EXPERIMENTAL STUDY

# Fatigue alters lower extremity kinematics during a single-leg stop-jump task

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**Abstract** To examine the kinematic characteristics of the hip and knee during a single-leg stop-jump task before and after exercise-to-fatigue, and to determine if the fatigue response is gender-dependent. Lower extremity kinematic measurements were taken of male and female subjects while they performed a sports functional task before and after fatigue developed from exhaustive running. Thirty healthy, physically active subjects (15 males and 15 females) Knee and hip joint kinematics were calculated utilizing three-dimensional video analysis. Each subject performed five single-leg stop-jumps before and after an exercise-to-fatigue bout. All subjects underwent a fatigue protocol using the modified Astrand protocol. Fatigue was verified using the Rating of Perceived Exertion along with the subject's heart rate. All data were analyzed using two factor (test × gender) repeated measures ANOVA ($P < 0.05$). Both males and females demonstrated significantly less maximal knee valgus ($P = 0.038$) and decreased knee flexion at initial contact ($P = 0.009$) post-fatigue. No significant differences were identified in hip joint angles between sessions or between sexes. The results show that fatigue developed from exhaustive running alters lower extremity kinematics during a single-leg stop-jump task. The more neutral position in the frontal plane might be an effort to protect the knee. The decrease in knee flexion at initial contact may be an attempt to increase knee stability following fatigue. Our results did not reveal any gender differences in this specific task.

**Keywords** Anterior cruciate ligament · Fatigue · Injury · Kinematics · Neuromuscular control · Stop-jump

## Introduction

The majority of anterior cruciate ligament (ACL) injuries occur during sports participation and are a result of a noncontact mechanism of injury [1, 2]. These injuries typically occur during a sharp deceleration task such as landing from a jump, a quick stop, or a cutting maneuver [1]. A more detailed examination of the underlying mechanism to ACL injury is crucial in an attempt to try to reduce the amount of injuries. Researchers have investigated the potential risk factors for noncontact ACL injuries and they can be categorized into two different areas: non-modifiable (anatomical and hormonal) and modifiable (neuromuscular and biomechanical) [3]. The majority of research into noncontact ACL injury risk factors has focused on the neuromuscular and biomechanical risk factors because of their potential for modification. Despite such efforts to prevent noncontact ACL injuries by modifying landing strategies, the incidence remains high [4]. Key factors associated with ACL injuries still need to be explored. Another potential modifiable factor is neuromuscular fatigue. However, the role of fatigue on neuromuscular control has not been sufficiently studied.

Fatigue is an extrinsic factor affecting the musculoskeletal and neurological systems [5]. It seems to create an environment that increases the risk of noncontact ACL

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injuries by altering lower extremity landing strategies. It has been reported that most athletic injuries occur in the later stages of activities and competition [6–12] indicating that vigorous exercise and fatigue may influence dynamic knee stability. Fatigue has been reported to result in decreased motor control performance [13, 14], increased knee joint laxity [14–16], decreased balance skill [13] and decreased proprioception [16–20]. These effects indicate a decreased capacity for controlling body movement after fatigue and may indicate fatigue as a contributor to non-contact ACL injuries [6, 17, 21]. Fatigue may have a cumulative, unfavorable effect, especially in latter stages of competition, on neuromuscular control and may potentially result in hazardous movement strategies [22]. Fatigue or resistance to fatigue could potentially be a modifiable risk factor that could improve the lower extremity alignment throughout movement.

Previous studies looking at fatigue and the effect on kinematic and kinetic characteristics found that subjects land with significantly increased peak proximal tibial anterior shear forces, increased valgus moments and decreased knee flexion angles during landings of stop-jump tasks [5]. Others investigated the influence of isolated quadriceps and hamstrings fatigue on landing motion [21]. They found that fatigue of the hamstrings resulted in decreased peak impact knee flexion moments, increased internal tibial rotation at peak knee flexion and decreased peak ankle dorsiflexion. Whereas quadriceps fatigue did result in increased peak ankle dorsiflexion moments, decreased peak knee extension moments, delayed peak knee flexion and delayed subtalar peak inversion moments [21]. More recently, others found that fatigue resulted in increased initial and peak knee abduction and internal rotation motions and peak knee internal rotation, adduction, and abduction moments, with the latter being more pronounced in females [22].

Considering the current knowledge regarding the effects of fatigue on the noncontact ACL injury mechanism, we decided to examine the kinematic characteristics of the hip and knee joints during a single-leg stop-jump task before and after exercise-to-fatigue, and to examine if fatigue affects each gender equally.

Our main purpose of this study was to examine the kinematic characteristics of the hip and knee joint during a single-leg stop-jump task before and after exercise-to-fatigue. The secondary purpose was to examine if the joint kinematics changes following exercise-to-fatigue are different between males and females. We hypothesized that fatigue developed from exhaustive running would result in: (1) increased internal hip rotation angle, (2) increased hip adduction angle, (3) increased knee valgus angle and (4) decreased knee flexion angle during a single-leg stop-jump task in both genders. We also hypothesized that females would exhibit increased alterations post-fatigue in knee and hip joint kinematics compared to males: (1) greater hip internal rotation changes compared to males, (2) greater hip adduction changes compared to male, (3) greater knee valgus changes compared to males and (4) greater knee flexion angle changes compared to males.

## Materials and methods

### Subjects

Thirty (15 males and 15 females) healthy and physically active subjects between the ages of 18–30 years old were recruited for this study. The term “physically active” was defined as individuals performing aerobic exercise for a minimum of 30 min a day, three times a week. All subjects were recreational athletes. Subject demographics are presented in Table 1. Written informed consent according to the University’s Institutional Review Board was obtained from the subjects prior to participation in the study.

### Data collection

Subjects wore their own clothes and shoes for the testing session. A Polar Tempo Heart Rate Monitor™ (U.S. model A3, Polar Electro Inc., Woodbury, NY) was attached to the subject according to the manufacturer’s instructions prior to data collection. Anthropometric data including height, weight, and lower extremity linear (anterior superior iliac spine width, thigh length, lower leg length, malleolus height, knee diameter, ankle diameter, foot breadth, and foot length) and circumferential (mid-thigh and mid-calf) measurements were obtained. Reflective markers, with a diameter of 0.025 m, were positioned on anatomical landmarks of each lower extremity as follows: posterior heel, lateral malleolus, second metatarsal head, lateral femoral condyle, anterior superior iliac spine and sacrum. Four additional markers were attached to wands (distance of 0.09 m from the skin) and placed at the lateral side of the mid-thigh and mid-calf [23]. A three-dimensional optical capture system (Vicon Motion Systems, Inc., Centennial,

**Table 1** Subject demographics

| | Age (years) Mean ± SD | Height (cm) Mean ± SD | Mass (kg) Mean ± SD |
|---|---|---|---|
| Males | 22.7 ± 1.6 | 180.3 ± 7.7 | 80.0 ± 10.5 |
| Females | 22.1 ± 1.7 | 169.1 ± 6.6 | 60.7 ± 7.4 |

CO) was used to collect coordinate data of the lower extremities during the single-leg stop-jump task. Six high speed (120 Hz) optical cameras (Pulnix Industrial Product Division, Sunnyvale, CA) encircled two force plates. A static calibration trial was conducted prior to data collection (pre- and post-fatigue). Based on the data collected from the camera recordings, the anatomical hip and knee joint angles were calculated to examine clinically applicable angles and were based on studies by Chao [24] and Grood and Suntay [25]. All of the kinematic calculations were performed in the Kinecalc module of the Vicon Motus software package (Vicon Motion Systems, Inc., Centennial, CO) and were based on the work of Vaughan et al. [23].

Kistler force plates (Kistler Instrument Corporation, Amherst, NY) embedded within adaptable flooring and instrumented with the Motus System were used for identification of four dependent variables (hip internal–external rotation, hip abduction–adduction, knee valgus–varus and knee flexion–extension) at initial contact. Initial contact was defined as the subject touching the force plate with 5% of his or her body weight.

The stop-jump task consisted of a single-leg standing jump onto a force plate followed immediately by a maximal effort vertical jump. This type of task was chosen as this is a high demanding task that requires a sudden deceleration that replicates a task frequently performed in basketball, volleyball and team handball. Noncontact ACL injuries frequently occur during this kind of task [1, 26]. Also, it is suggested that even during double-legged movements the weight during noncontact ACL injuries is for 65–100% on one leg (with 8 out of 12 being 100%) [26]. Other research has indicated that the hip plays an important role in the kinetic chain of the lower extremity [27, 28] and might therefore play an important role in the gender differences in ACL injury rate. We therefore decided to include the knee and hip in our motion analyses.

Subjects were asked to land on the force plate following the maximal effort vertical jump. Subjects started the single-leg jump at a distance equal to 40% of their height from the edge of the force plate on their preferred leg. The preferred leg was defined as the leg that the subject was most comfortable completing the task with. Subjects were instructed in the proper technique (proper start position, single-leg landing on the force plate, maximum effort vertical jump) and provided practice trials to familiarize themselves with the task. Five acceptable trials were collected for pre- and post-fatigue sessions on the preferred leg. An acceptable trial was defined as landing on the force plate with the foot completely on the force plate without losing balance (i.e. touching nonjumping leg or ground prior to jump), prior to the maximal effort vertical jump. When the subject failed to jump single-legged and/or lost his or her balance, the trial was considered as invalid, and the subject performed another trial. Post-exercise procedures were completed immediately after the fatigue protocol and were identical to the pre-test measures.

All subjects underwent a fatigue protocol using the modified Astrand protocol [29, 30] in conjunction with Borg's 15-category rating of perceived exertion (RPE) [31], in addition their heart rate was monitored to establish the threshold of exercise intensity. The equation of the estimated maximal heart rate ($HR_{max}$) was determined by subtracting subject's age from 220 (Estimated $HR_{max}$ = 220 – age) [32]. The heart rate difference between the end of the fatigue protocol and the initiation of the post-fatigue test, and the time between the end of the fatigue protocol and the last post-fatigue trial were also averaged. The fatigue developed in this protocol is defined as fatigue developed from exhaustive running, as the subjects gave their maximal effort and ran until exhaustion.

After proper explanation, each subject was instructed to start walking for 3 min as a warm-up on a Wellness Treadmill WTM410 (model no. 945-240, Biodex Medical System Inc., Shirley, NY), at a speed of 2 mph (0% grade). As soon as the 3-min warm-up session was completed, the speed was increased to 5–8 mph for 3 min (0% grade) according to the subjects' comfortable running pace. After 3 min at 0% grade running, the grade was increased 2.5% every 2 min throughout the session until the termination of the fatigue protocol (i.e., subject could not run anymore at maximum effort). During the test the subject was verbally encouraged to continue running until exhausted. Mean running speed was 6.3 ± 0.62 mph in males and 5.8 ± 0.43 mph in females. The running speed set at the beginning remained constant throughout the fatigue protocol. Immediately following the termination of the fatigue protocol all subjects performed the same single-leg stop-jump task as before running. The time between the end of the fatigue protocol and the initiation of the post-fatigue test, and the time between the end of the fatigue protocol and the last trial were averaged (Table 3).

## Data reduction

The kinematic variables of knee and hip joints were calculated using Motus Software video analysis system (Version 7.2.3; Vicon Motion Systems, Inc., Centennial, CO) based on Vaughan et al. [23]. Raw three-dimensional coordinate data from single-leg stop-jump were filtered using a fourth order zero lag Butterworth digital filter with an optimal cut-off frequency [33]. All kinematic data was normalized to the static calibration trial that was collected prior to each set of jumps (pre- and post-fatigue). Hip and knee joint kinematics (hip internal–external rotation, hip adduction–abduction, knee valgus–varus, and knee flexion–extension) were evaluated at initial contact and at the maximum value of each variable respectively. For both test sessions, the first three successful trials were digitized and averaged for each subject from the initial contact onto the force plate to the second landing. Vertical jump height was measured using the sacral marker (change in percent body height with respect to the static trial) and was expressed as percent body height.

### Data analysis

The knee and hip joint kinematic measurements (hip internal–external rotation, hip abduction–adduction, knee valgus–varus, and knee flexion–extension) were used for data analysis. All data were analyzed using two factor (test × gender) repeated measures ANOVA on the knee and hip variables. The independent variables in each analysis were gender and fatigue condition. The alpha level for all statistical analysis was set at $P < 0.05$.

## Results

There were no gender differences for subject demographics (age, $P = 0.85$; height, $P = 0.59$; mass, $P = 0.09$), mean running speed ($P = 0.36$), maximal heart rate ($P = 0.15$), heart rate at the last averaged trial ($P = 0.74$), or percent maximal heart rate ($P = 0.43$). Overall, the jump height as a measure of performance did decrease from 17.71 ± 4.53 pre-fatigue to 16.70 ± 4.01 post-fatigue (% body height). For males, the jump height pre-fatigue was 20.16 ± 4.52 compared to 18.08 ± 4.19 post-fatigue. For females however, the jump height did not change after the fatigue protocol: 15.26 ± 3.06 pre-fatigue compared to 15.32 ± 3.42 post-fatigue (Table 2). Information about the fatigue developed from high-intensity running is listed in Table 3. The average time to fatigue was 14:00 ± 2:34 min in males, and 11:43 ± 1:50 min in females ($P = 0.18$). The mean elapsed time between the termination of fatigue and the beginning of the first post-fatigue jump was 1:13 ± 0:18 min in males and 1:10 ± 0:15 min in females ($P = 0.61$). Male and female subjects completed the post-fatigue jump trials in 1:52 ± 0:23 and 1:49 ± 0:17 min ($P = 0.20$) respectively after termination of fatigue protocol. The heart rate was also recorded after each post-fatigue single-leg stop-jump.

The pre-fatigue session and post-fatigue session mean test scores for the dependent measures of hip internal–external rotation angles, hip adduction–abduction angles, knee valgus–varus angles, and knee flexion–extension angles for both genders are presented in Table 4. There were no significant differences for any of the hip kinematic variables the between pre-fatigue session and post-fatigue session or between genders. Post-fatigue, both males and females showed significantly less maximal knee valgus angles ($P = 0.038$) and knee flexion angles ($P = 0.009$) at initial contact. No gender difference existed in the other knee valgus–varus variables or in the knee flexion–extension variables between males and females either in the pre- and post-fatigue sessions.

**Table 2** Maximal height of jumping (% body height)

| | Mean ± SD |
|---|---|
| Pre-fatigue | |
| Males | 20.16** ± 4.52 |
| Females | 15.26** ± 3.06 |
| Total | 17.71* ± 4.53 |
| Post-fatigue | |
| Males | 18.08** ± 4.19 |
| Females | 15.32** ± 3.42 |
| Total | 16.70 ± 4.01 |

* Statistical significance at $P < 0.05$ (pre–post)

** Statistical significant gender difference at $P < 0.05$

**Table 3** Fatigue information

| | Males Mean ± SD | Females Mean ± SD | $P$ value |
|---|---|---|---|
| Time to fatigue (min) | 14:00 ± 2:34 | 11:43 ± 1:50 | 0.182 |
| Time 1[a] (min) | 1:13 ± 0:18 | 1:10 ± 0:15 | 0.614 |
| Time 2[b] (min) | 1:52 ± 0:23 | 1:49 ± 0:17 | 0.203 |
| Max HR (bpm) | 193 ± 9.5 | 188 ± 6.2 | 0.153 |
| HR at the end (bpm) | 156 ± 13.6 | 148 ± 10.9 | 0.742 |
| % Max HR | 80.8 ± 5.9 | 78.5 ± 5.2 | 0.431 |

[a] Time 1 is the time between termination of fatigue and the start of the post-fatigue jump trials

[b] Time 2 is the time between termination of fatigue and the completion of the post-fatigue jump trials

## Discussion

We were interested in the effect of fatigue developed from exhaustive running on lower extremity kinematics during landing due to reports of athletic injuries occurring in different sports in the later stages of activities and competition [6, 7]. In addition, since research studies have reported that female athletes are sustaining noncontact ACL injuries more frequently than their male counterparts [1, 34–36], we were interested in whether fatigue developed from exhaustive running would change lower

**Table 4** Hip and knee kinematics pre- and post-fatigue

| | Group | Pre-fatigue | Post-fatigue |
|---|---|---|---|
| Hip kinematics | | | |
| Hip internal–external rotation at initial contact (deg) | Males | –2.6 ± 8.5 | –4.4 ± 8.2 |
| | Females | –3.2 ± 5.2 | –3.6 ± 3.4 |
| | Total | –2.9 ± 7.0 | –4.0 ± 6.2 |
| Maximum hip internal–external rotation (deg) | Males | 3.4 ± 6.0 | 1.1 ± 5.5 |
| | Females | 3.2 ± 5.3 | 2.3 ± 3.8 |
| | Total | 3.3 ± 5.5 | 1.7 ± 4.7 |
| Hip adduction–abduction at initial contact (deg) | Males | 7.1 ± 4.7 | 6.2 ± 6.7 |
| | Females | 4.7 ± 4.3 | 5.9 ± 4.3 |
| | Total | 5.9 ± 4.6 | 6.0 ± 5.5 |
| Maximum hip adduction–abduction (deg) | Males | 12.2 ± 4.7 | 11.9 ± 5.2 |
| | Females | 9.7 ± 4.3 | 10.7 ± 4.3 |
| | Total | 11.0 ± 4.6 | 11.3 ± 4.7 |
| Knee kinematics | | | |
| Knee valgus–varus angles initial contact (deg) | Males | –0.4 ± 3.0 | –0.9 ± 2.2 |
| | Females | 3.1 ± 1.9 | 3.0 ± 1.9 |
| | Total | 1.36 ± 3.0 | 1.0 ± 2.8 |
| Maximal knee valgus–varus (deg) | Males | 1.5 ± 3.6 | 0.8 ± 3.5 |
| | Females | 5.5 ± 2.5 | 4.6 ± 2.3 |
| | Total* | 3.5 ± 3.7 | 2.7 ± 3.5 |
| Knee flexion angles at initial contact (deg) | Males | 12.2 ± 5.1 | 10.6 ± 4.9 |
| | Females | 14.9 ± 5.3 | 12.3 ± 6.6 |
| | Total* | 13.6 ± 5.3 | 11.5 ± 5.8 |
| Maximal knee flexion (deg) | Males | 51.2 ± 6.8 | 51.3 ± 7.4 |
| | Females | 49.0 ± 7.3 | 48.5 ± 8.0 |
| | Total | 50.1 ± 7.0 | 49.9 ± 7.7 |

Hip internal rotation (+), hip external rotation (–), hip abduction (+), hip adduction (–), knee valgus (+), knee varus (–)

* Statistically significant at $P < 0.05$

extremity kinematics and if the effect of fatigue would be more pronounced in females than in males. The first purpose of this research study was therefore to investigate the influence of fatigue on hip and knee joint kinematics in healthy, active males and females. Secondly, we wanted to determine if fatigue affects both genders differently in terms of joint kinematics, in order to hopefully provide more insight in why females are more prone to noncontact ACL injuries.

Females did not demonstrate a significant difference in jump height pre- versus post-fatigue, however males did show a decrease in jump height after the fatigue protocol. Although the decrease in jump height in males was statistically significant, we feel that this has minimal clinical implications as the difference was only ~2% of their body height.

In the current study, it was hypothesized that fatigue would result in altered movement patterns during a single-leg stop-jump task in both genders, with more change post-fatigue in females. Our first hypothesis was partially fulfilled in that there was a significant decrease in the maximal knee valgus angle and knee flexion angle at initial contact between pre- and post-fatigue in both males and females. There were no significant pre- and post-fatigue differences in other variables analyzed.

The result of this study did not support our hypothesis that fatigue developed from exhaustive running would result in an increased knee valgus angle. However, both males and females showed more neutral knee position (less valgus angle) in the post-fatigue condition. The maximal knee valgus angle was decreased in both groups post-fatigue (females 4.6 ± 2.3°, males 0.8 ± 3.5°, $P = 0.038$). Even though both females and males were still in knee valgus, this result suggests that both groups tended to bring the knee in more neutral position in the frontal plane in post-fatigue condition. This might be a protective effort as studies have suggested that an increase in knee varus or valgus moment tends to increase the stress on the ACL [37–40].

Previous cadaveric studies have shown that a decreased knee flexion angle can increase ACL strain [40–44]. The results of this study did support our hypothesis that the subjects would land with less knee flexion. The decrease in knee flexion at initial contact was significant in both genders *(P* = 0.009). Both females and males landed with less knee flexion post-fatigue (12.3 ± 6.6° for females and 10.6 ± 4.9° for males). This type of landing may increase ACL strain and support evidence suggesting that more injuries occur at the later stages of a competition. Although the decrease was small, this may be an attempt to increase knee stability following fatigue, because decreased knee flexion prevents valgus and landing with the knee less flexed does not require as much eccentric muscular strength. With less knee flexion, subjects could rely more on the bony architecture and noncontractile tissues surrounding the knee, like the tightened collateral ligaments. Muscle fibers have a decreased capacity to absorb energy when fatigued [45], but although this might be a position requiring less "energy", the subject may be at greater risk for injury due to the increased strain in the ACL. The decreased knee flexion angle combined with previous research demonstrating increased anterior tibial translation with fatigue [14] may increase the risk of ACL injury during the latter stages of competition.

Also, although less knee flexion does not require as much eccentric strength of the quadriceps, the mechanism seen in this study in the decrease in knee flexion and valgus angles may not be present anymore in an extreme level of fatigue towards the end of a game, as this position probably required more muscle activity from the hip abductors, especially the gluteus medius. It can be expected that more detrimental loading patterns, including more knee valgus, will be seen during a higher level of exhaustion.

The second hypothesis was not supported by our results and likely resulted from the combination of the small sample size for gender comparisons (15 males and 15 females) and small effect size for gender differences. If males and females did adopt different landing strategies as a result of fatigue for this study, the low power of the gender results prohibits robust conclusions. Therefore future research in this area is needed to further explore if fatigue affects gender differently.

In this study, we did not see compensation in the hip joint for the loss in knee flexion. As hip joint motion did not change significantly and ankle kinematics were not assessed, we could say that our subjects landed with a stiffer landing and energy of the lower extremity was therefore not dissipated as efficiently in the fatigued condition. Knee joint flexion decreased and no compensatory response to absorb the mechanical energy of the impact was seen. This suggests that the subjects landed in a more dangerous position and the probability of ACL injury may be increased [46].

Several fatigue protocols have been discussed in the literature. An isokinetic dynamometer has been used to induce fatigue for the local muscles [14, 15, 47] and whole lower extremity [13]. Rodacki et al. induced a fatigue by repeating vertical jumping until subjects no longer achieved 70% of their maximal jump height [6]. Nyland et al. used a general lower extremity fatigue protocol using uphill treadmill walking [48]. They terminated the protocol when subjects could no longer continue [48]. In the current study fatigue was induced by having subjects run on a treadmill using the modified Astrand protocol [29, 30] in conjunction with RPE. The protocol utilized a standard RPE and heart rate to maximize accuracy and minimize errors of fatigue status [49]. We selected this fatigue protocol because running would create a similar condition as actual sports activities, and the inclination would induce even more fatigue localized in the lower extremity. Because of the nature of the sports in regard to repetitive movement and running, not only whole body aerobic fatigue, but also local muscle fatigue will occur during the athletic event. A focus of the current research was to induce a fatigue pattern that simulates the fatigue as seen in these real athletic events. The termination of fatigue was at the time when the subject was no longer able to continue the protocol. Although Chappell et al. [5] used a different fatigue protocol compared to ours (unlimited repetitions of five consecutive vertical jumps followed by a 30-m sprint), our results are similar to theirs in that fatigued recreational athletes demonstrated altered motor control strategies. It appears as though fatigue developed from exhaustive running alters the strategies to maintain joint stability and supports the hypothesis of more injuries occurring at the later stages of athletic competition.

The average time to reach fatigue status for this protocol was 14:00 ± 2:34 min in males and 11:43 ± 1:50 min in females. This average time may potentially simulate the fatigue experienced in the middle or later stages of competition when research reports have indicated that many injuries occur [6, 7]. Furthermore, we measured the time between termination of fatigue and the post-fatigue jump tasks. We tested the post-fatigue jumps immediately after the termination of fatigue, and the last jump task was collected within 2 min after the termination of the fatigue protocol (Table 3). HR was also recorded (Table 3). These measurements assure that the subjects were still in the state of fatigue during the post-fatigue stop-jump tasks.

## Limitations

Since no EMG measurements were taken we cannot exactly comment on the muscular contribution to the observed kinematic changes. The decrease in knee flexion at initial contact could however very likely be a result of quadriceps fatigue, making eccentric contraction difficult. As maximum knee valgus decreased, it would also be interesting to compare the firing patterns of hip musculature pre-and post-fatigue. Future research is therefore needed to investigate the influence of fatigue on muscle firing patterns, which is important to know in an attempt to integrate neuromuscular and biomechanical findings with the noncontact ACL injury mechanism. Also, ankle kinematics was not assessed and it is therefore difficult to thoroughly analyze the work distribution of the entire lower extremity. For future research it is recommended to include analysis of ankle motion [22]. Although the effect of fatigue on lower extremity kinematics was our primary objective, limitations to the current study also included a small sample size for gender comparisons. Furthermore, the generalization of the study results is limited to the specific fatigue protocol utilized in this study. The results from this study do not hold true for all forms of exercise to fatigue.

## Conclusion

Our results revealed that both males and females used a stiff landing strategy following fatigue by landing with less maximal knee valgus and less knee flexion at initial contact of the single-leg stop-jump, without changing the hip joint angles. The decrease in knee valgus angle might be a protective effort as an increase in knee varus or valgus moment tends to increase the stress on the ACL. The decrease in knee flexion at initial contact may be an attempt to increase knee stability while relying on the static structures of the knee more than the dynamic structures following fatigue. Our results did not reveal any gender differences in this specific task.

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