



Effects of fatigue on lower limb, pelvis and trunk kinematics and muscle activation: Gender differences



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ABSTRACT

Background: Muscle fatigue is associated with biomechanical changes that may lead to anterior cruciate ligament (ACL) injuries. Alterations in trunk and pelvis kinematics may also be involved in ACL injury. Although some studies have compared the effects of muscle fatigue on lower limb kinematics between men and women, little is known about its effects on pelvis and trunk kinematics. The aim of the study was to compare the effects of fatigue on lower limb, pelvis and trunk kinematics and muscle activation between men and women during landing. **Methods:** The participants included forty healthy subjects. We performed kinematic analysis of the trunk, pelvis, hip and knee and muscle activation analysis of the gluteal muscles, vastus lateralis and biceps femoris, during a single-leg landing before and after fatigue. **Results:** Men had greater trunk flexion than women after fatigue. After fatigue, a decrease in peak knee flexion and an increase in Gmax and BF activation were observed. **Conclusion:** The increase in the trunk flexion can decrease the anterior tibiofemoral shear force resulted from the lower knee flexion angle, thereby decreasing the stress on the ACL.

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1. Introduction

One of the most common injuries during sports activity is anterior cruciate ligament (ACL) disruption (Yu and Garrett, 2007). ACL injury tends to occur without contact in activities that involve cutting, pivoting, decelerating, or landing from a jump (Ireland, 1999; Hewett et al., 2005; Yu and Garrett, 2007). In addition, women are four to six times more prone to ACL injuries than men, when participating in the same sporting activity (Boden et al., 2000).

There are several intrinsic and extrinsic factors linked to the noncontact ACL injury disparity between genders (Arendt and Dick, 1995). Fatigue is an extrinsic factor affecting the neurological and musculoskeletal systems (Chappell et al., 2005). Most athletic injuries occur in the later stages of activities and competition, indicating that fatigue may play a crucial role in the incidence of injury (Hawkins and Fuller, 1999; Hawkins et al., 2001; Price et al., 2004). Muscle fatigue can lead to a reduced ability of the muscles to generate strength (Lattier et al., 2004) and altered neuromuscular control (McLean et al., 2007). In this context, fatigue causes abnormal

and potentially hazardous movement strategies, increasing the risk of a noncontact ACL injury during landing (Santamaria and Webster, 2010).

Several studies have evaluated the effects of fatigue on landing biomechanics, demonstrating the influence of fatigue on lower limb kinematics and differences between genders (McLean et al., 2007; Kernozek et al., 2008; Gehring et al., 2009; Brazen et al., 2010; Liederbach et al., 2014). The most commonly studied variables are knee and hip kinematics in the sagittal and frontal planes. However, studies involving the effects of lower limb muscle fatigue on landing kinematics in subjects of different genders did not include assessments of proximal segments, such as the pelvis or trunk.

Decreases of lower limb muscle activation due to fatigue can result in changes in pelvis and trunk position. The influence of fatigue on trunk position is an important aspect because it can change the loads on the knee joint and the stress on the ACL (Kulas et al., 2012). In the sagittal plane, the trunk extension is a common used strategy to decrease the demand on fatigued/weak hip extensor muscles (Powers, 2010). However, a smaller trunk flexion during landing increases the quadriceps muscle activation (Blackburn and Padua, 2009) and, consequently, the anterior tibiofemoral shear force and the stress on the ACL (especially with the knee close to full extension) (Kulas et al., 2012).

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the pelvis and trunk angles were calculated relative to the global coordinate system (global horizontal axis and global vertical axis). The knee joint center was determined as the midpoint between the medial and lateral epicondyles of the femur. The hip joint center was estimated as one-quarter of the distance from the ipsilateral to the contralateral greater trochanter (Weinhandl and O'Connor, 2010). Kinematic data were filtered using a fourth-order zero-lag Butterworth 12-Hz low-pass filter.

A custom program in Matlab (Mathworks, USA) was used to analyze the kinematic variables of interest. The first landing was analyzed. The joint angles at the first initial contact and the peak angles during the landing phase (from the initial contact to maximal knee flexion) were considered for analysis. The initial contact was defined as the instant in which the vertical velocity of the marker fixed on the second toe was zero. The velocity was calculated from the first derivative of the toe marker (Brindle et al., 2003). The kinematic angles of interest at initial contact and peak angles during landing included the following: trunk flexion, trunk ipsilateral lean, pelvis contralateral drop, hip flexion, hip adduction, knee flexion and knee abduction. By convention, the positive kinematic values represented flexion, abduction, ipsilateral trunk lean and pelvic drop angles.

All EMG data were processed using Matlab (Mathworks, USA). Raw EMG signals were band-pass filtered at 20–400 Hz, full-wave rectified and smoothed by a symmetrical moving root mean square (RMS) filter (20 ms time constant). The RMS activity (mean average amplitude) was calculated during the landing phase (from the initial contact to maximal knee flexion). The peak RMS activity during the landing phase represents 100% activity and the average RMS data during landing were expressed as a percentage of the peak RMS during landing (Zebis et al., 2011, 2008). The EMG normalization was performed for each landing separately.

For the kinematic and EMG variables, the average of three trials was used for the statistical analysis.

3. Statistical analysis

All statistical analyses were performed using SPSS statistical software (version 17.0; SPSS, Inc., IL). All data were expressed as the mean and standard deviation. The Student's *t*-tests for independent samples were used to verify the differences in the demographic characteristics of the groups. The kinematic and EMG data were considered dependent variables. The effects of fatigue on the dependent variables were evaluated by a two-way (gender X fatigue) ANOVA with a mixed-model design, with the fatigue as a repeated measure. When significant differences were found, a pairwise comparison was performed using the Bonferroni-adjusted *t*-tests. All analyses were conducted with an alpha level of 0.05.

4. Results

No significant difference was observed between participants in terms of the amount of physical activity practice per week (men 4.3 ± 1.1 h per week and women 3.8 ± 0.9 h per week; $P > 0.05$). Men completed an average of 10.4 ± 2.8 successful series of the fatigue protocol and reported a mean Borg rating of 8.2 ± 1.2 . Women completed an average of 9.1 ± 3.0 successful series of the fatigue protocol and reported a mean Borg rating of 8.2 ± 0.8 ; there were no significant differences between the groups for these variables ($P > 0.05$).

The kinematic results are reported in Tables 1 and 2, and Fig. 1. The RMS activity results are reported in Table 3. Gender X fatigue interactions were observed for the peak trunk flexion during landing ($P = 0.024$), knee in frontal plane at initial contact ($P = 0.047$) and BF average amplitude of activation during landing

($P = 0.037$). Men had increased peak trunk flexion during landing after fatigue compared to before fatigue ($P = 0.001$; mean difference (MD) = 7.7° ; 95% confidence interval for difference (95%CI) = 4.5 – 11.0). At the initial contact after fatigue women presented greater knee abduction angles compared to men after fatigue ($P = 0.013$; MD = 2.8° ; 95%CI = 0.6 – 4.9). Men had decreased knee abduction at initial contact after fatigue compared to before fatigue ($P = 0.035$; MD = 0.8° ; 95%CI = 0.06 – 1.6). Furthermore, women presented greater BF average amplitude of activation after fatigue compared to before fatigue ($P = 0.004$; MD = 7.0% ; 95%CI = 2.4 – 11.6). After fatigue women shown greater BF activation compared to men after fatigue ($P = 0.043$; MD = 5.9% ; 95%CI = 0.2 – 11.6).

Moreover, some main effects were observed. After fatigue, the following peak angles were observed during landing: increased peak trunk flexion ($P < 0.001$; MD = 5.1° ; 95%CI = 2.7 – 7.4); increased peak contralateral pelvic drop ($P < 0.001$; MD = 1.1° ; 95%CI = 0.5 – 1.6) and decreased peak knee flexion ($P < 0.001$; MD = 4.1° ; 95%CI = 2.6 – 5.6). Also, after fatigue an increased contralateral pelvic drop was observed at the initial contact ($P < 0.001$; MD = 0.9° ; 95%CI = 0.4 – 1.4) and the average amplitude of activation during landing for BF ($P = 0.037$; MD = 3.5% ; 95%CI = 0.2 – 6.7) and GMax ($P = 0.013$; MD = 4.4% ; 95%CI = 1.0 – 7.6) were increased. Furthermore, women presented significantly greater knee abduction at the initial contact ($P = 0.039$; MD = 2.2° ; 95%CI = 0.1 – 4.3) and greater peak knee abduction during landing ($P = 0.005$; MD = 3.3° ; 95%CI = 1.1 – 5.5) than men.

5. Discussion

The aim of this study was to evaluate the effects of muscle fatigue in trunk, pelvis, hip and knee kinematics, and lower limb muscle activation between men and women. The hypothesis that fatigue would alter landing biomechanics in men and women, but with different alterations between genders was confirmed.

The main innovation of this study was the evaluation of the effects of fatigue on the trunk and pelvis angles. The change in the trunk position can alter the loads on the knee joint (Powers, 2010) and a deficit in trunk neuromuscular control is a predictor of knee injury risk in women (Zazulak et al., 2007). In the present results, after fatigue, men showed greater trunk flexion when compared to women. After the fatigue protocol, an increase in the peak trunk flexion was also observed by Liederbach et al. (2014) during a single-leg landing in healthy individuals. Trunk flexion moves the resultant vector of the ground reaction force forward, increasing the external hip flexion moment and in order to control this increase, the hip extensor muscles must generate a greater amount of force, activating more muscle fibers (Powers, 2010). This could explain the increase in the GMax and BF activation observed during landing after fatigue. According to the study of Kulas et al. (2012), increased trunk flexion minimized ACL deformation during squatting even in lower angles of knee flexion. In addition, Blackburn and Padua (2008) report that females typically display a more erect posture during landing compared to men, potentially contributing to the higher female ACL injury rate.

In this study, fatigue increased the contralateral pelvic drop in initial contact and during landing. It should be noted that although there was an increase in the contralateral pelvic drop, there was no change in the GMed activation. Note that the fatigue protocol emphasized the quadriceps and hip extensor muscles. Therefore, it is possible that fatigue reduced GMax's ability to generate strength. The GMax muscle acts in hip extension and lateral rotation; also, its upper fibers act in hip abduction (Lyons et al., 1983). Therefore, that muscle can contribute to the maintenance of pelvic stability in the frontal plane, assisting in the GMed action. Then, GMax fatigue may be involved in the increased pelvic con-

Table 1
Mean ± standard deviation (degrees) for the kinematic data at initial contact.

	Men		Women		Group		Fatigue	
	Fatigue		Fatigue		Men	Women	Pre	Post
	Pre	Post	Pre	Post				
Knee sagittal plane	12.4 ± 4.3	13.0 ± 5.7	11.1 ± 4.9	9.9 ± 3.9	12.7 ± 5.0	10.5 ± 4.5	11.8 ± 4.6	11.5 ± 5.1
Knee frontal plane	0.1 ± 3.9	-0.7 ± 3.8 ^a	1.8 ± 2.8	2.0 ± 2.8 ^b	-0.3 ± 3.8	1.9 ± 2.8 ^c	0.9 ± 3.4	0.7 ± 3.6
Hip sagittal plane	22.1 ± 5.3	22.1 ± 7.7	23.3 ± 4.0	24.0 ± 7.9	22.1 ± 6.5	23.7 ± 6.2	22.7 ± 4.7	23.0 ± 7.8
Hip frontal plane	5.3 ± 5.1	5.4 ± 3.4	5.4 ± 3.3	4.9 ± 3.8	5.4 ± 4.2	5.1 ± 3.5	5.4 ± 4.2	5.1 ± 3.6
Pelvis frontal plane	-4.9 ± 3.1	-4.2 ± 2.2	-4.8 ± 2.18	-3.7 ± 2.3	-4.5 ± 2.6	-4.2 ± 2.3	-4.8 ± 2.6	-3.9 ± 2.2 ^d
Trunk sagittal plane	3.1 ± 6.8	4.3 ± 8.3	3.5 ± 7.9	5.0 ± 7.8 ^b	3.7 ± 7.5	4.3 ± 7.8	3.3 ± 7.3	4.7 ± 7.9
Trunk frontal plane	0.6 ± 2.5	0.3 ± 3.7	1.3 ± 3.7	0.6 ± 3.5	0.5 ± 3.1	0.9 ± 3.6	0.9 ± 3.1	0.5 ± 3.6

Flexion (+); extension (-); abduction (+); adduction (-); drop (+); elevation (-); ipsilateral lean (+); contralateral lean (-).

^a Significant difference compared to men prefatigue ($P < 0.05$).

^b Significant difference compared to men postfatigue ($P < 0.05$).

^c Significant difference compared to men ($P < 0.05$).

^d Significant difference compared to prefatigue ($P < 0.05$).

Table 2
Mean ± standard deviation (degrees) for the kinematic peak angles during landing.

	Men		Women		Group		Fatigue	
	Fatigue		Fatigue		Men	Women	Pre	Post
	Pre	Post	Pre	Post				
Knee sagittal plane	60.3 ± 7.8	56.3 ± 8.6	58.8 ± 8.7	54.6 ± 9.2	58.3 ± 8.4	56.7 ± 9.1	59.5 ± 8.2	55.4 ± 8.8 ^c
Knee frontal plane	2.5 ± 4.3	1.9 ± 3.4	5.4 ± 3.63	5.6 ± 3.5	2.2 ± 3.9	5.5 ± 3.5 ^b	4.0 ± 4.2	3.7 ± 3.9
Hip sagittal plane	55.4 ± 10.7	56.4 ± 11.8	53.6 ± 13.3	51.7 ± 13.1	55.9 ± 11.2	52.6 ± 13.1	54.5 ± 12.0	54.0 ± 12.6
Hip frontal plane	-4.5 ± 6.4	-4.1 ± 3.8	-6.6 ± 6.1	-5.9 ± 4.2	-4.3 ± 5.2	-6.3 ± 5.2	-5.6 ± 6.3	-5.0 ± 4.1
Pelvis frontal plane	-1.6 ± 3.6	-0.6 ± 3.1	-0.1 ± 3.1	1.1 ± 2.5	-1.1 ± 3.4	0.5 ± 2.9	-0.9 ± 3.4	0.2 ± 2.9 ^c
Trunk sagittal plane	25.5 ± 12.3	33.2 ± 12.5 ^a	24.5 ± 15.1	26.9 ± 14.2	29.4 ± 12.9	25.7 ± 14.5	25.0 ± 13.6	30.1 ± 13.6 ^c
Trunk frontal plane	9.5 ± 5.1	9.5 ± 5.6	11.0 ± 3.0	9.34 ± 2.9	9.5 ± 5.3	5.1 ± 5.4	5.2 ± 6.2	9.4 ± 4.4

Flexion (+); extension (-); abduction (+); adduction (-); drop (+); elevation (-); ipsilateral lean (+); contralateral lean (-).

^a Significant difference compared to men prefatigue ($P < 0.05$).

^b Significant difference compared to men ($P < 0.05$).

^c Significant difference compared to prefatigue ($P < 0.05$).

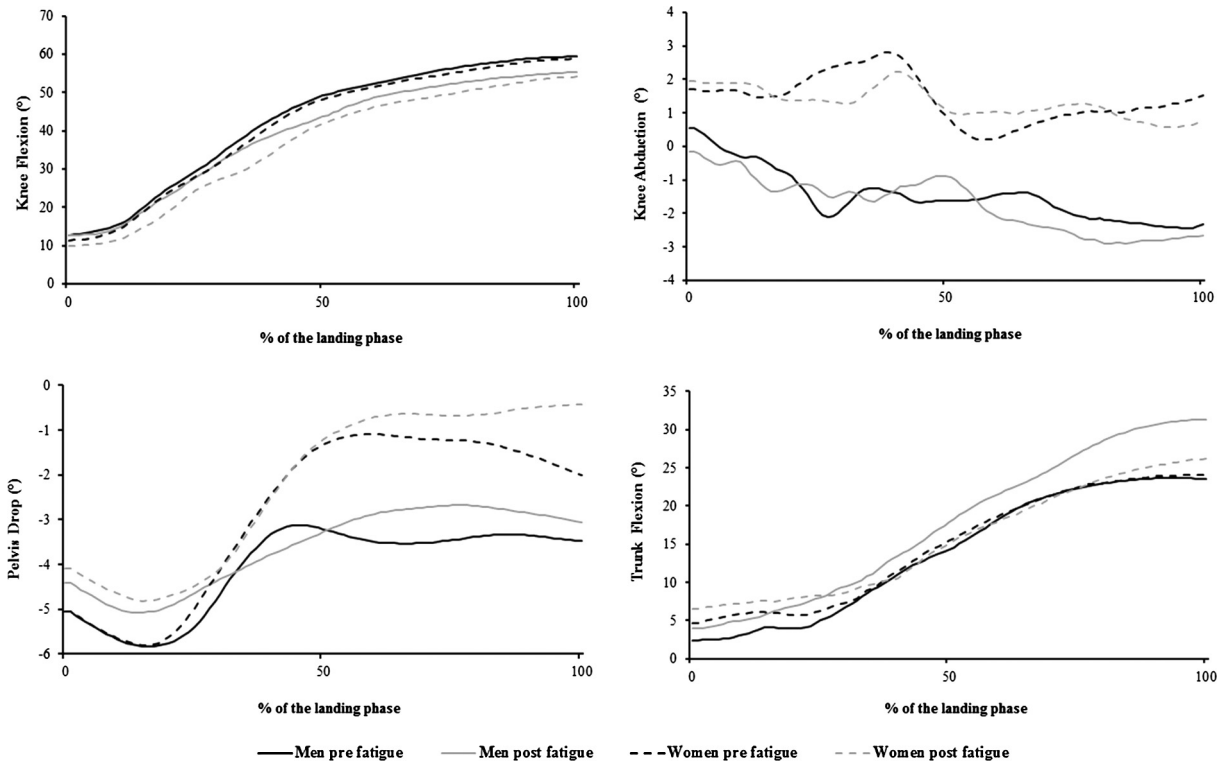


Fig. 1. Average time normalized curves for joint angles during the landing phase before and after fatigue for knee flexion, knee abduction, pelvis drop and trunk flexion.

Table 3

EMG average amplitude of EMG during landing.

	Men		Women		Group		Fatigue	
	Pre fatigue	Post fatigue	Pre fatigue	Post fatigue	Men	Women	Pre	Post
VL	62.2 ± 6.7	64.1 ± 9.2	63.6 ± 6.6	64.2 ± 8.7	63.2 ± 8.0	63.9 ± 7.7	62.9 ± 6.6	64.2 ± 8.8
BF	57.7 ± 6.2	57.6 ± 9.7	56.5 ± 9.1	63.5 ± 8.1 ^{a,b}	57.7 ± 8.0	60.0 ± 9.2	57.1 ± 7.7	60.6 ± 9.3 ^c
GMed	57.6 ± 9.6	57.2 ± 9.0	56.0 ± 11.8	60.5 ± 11.6	57.4 ± 9.2	58.3 ± 11.8	56.8 ± 10.6	58.9 ± 10.4
GMax	51.4 ± 8.9	56.7 ± 10.1	52.7 ± 9.7	56.1 ± 9.2	54.1 ± 9.7	54.4 ± 9.5	52.0 ± 9.2	56.4 ± 9.5 ^c

Data are given as the mean ± standard deviation, expressed as a percentage of the peak EMG during landing.

^a Significant difference compared to women prefatigue ($P < 0.05$).

^b Significant difference compared to men postfatigue ($P < 0.05$).

^c Significant difference compared to prefatigue ($P < 0.05$).

tralateral drop. We must consider that although a significant increase in the contralateral pelvic drop was observed, the increase was approximately one degree, and the clinical implications of this small increase are still unclear.

The present results showed greater knee abduction in women compared to men, agreeing with the results reported in the literature (Kernozek et al., 2005; Jacobs et al., 2007). These results indicated that in a single-leg landing women are in greater injury risk compared to men, since Hewett et al. (2005) demonstrated that the knee valgus angle (abduction) is a predictor of the ACL injury risk in women. Fatigue has failed to increase knee abduction angle, perhaps because the fatigue protocol has focused on the muscles that control movements in the sagittal plane and not in the frontal plane (as noted the GMed had no activation change).

The knee sagittal plane position is also related to ACL injury. In the present study, no differences were observed between the genders, but the peak knee flexion during landing decreased after fatigue. Chappell et al. (2005) reported that the decrease in knee flexion angle during landing was associated with increased anterior tibiofemoral shear force. Therefore, it is possible that the greater trunk flexion and consequent increased activation of the hamstring muscles could minimize the increase in the tibiofemoral shear force resulting from the lower knee flexion. In this way, trunk flexion could be used, as a strategy to reduce stress on the ACL in the presence of fatigue.

The authors acknowledge that this study has some limitations. The evaluation of the strength of the hip and knee muscles might improve our understanding of the alterations resulting from fatigue. In addition, the fatigue protocol primarily consisted of bilateral activities in the sagittal plane. Therefore, while one of the objectives of the study was to evaluate the effects of fatigue in the GMed activation, it is possible that the protocol used did not generate fatigue in this muscle. Therefore, future studies should apply a fatigue protocol that emphasizes the action of GMed and assess whether the kinematic and EMG changes differ from those observed in the present study.

6. Conclusions

The increase in the peak trunk flexion may help to decrease the anterior tibiofemoral shear force resulted from lower knee flexion angle, thereby decreasing the stress on the ACL. Men presented greater trunk flexion after fatigue compared to women, suggesting that males were able to better adapt to the fatigue condition.

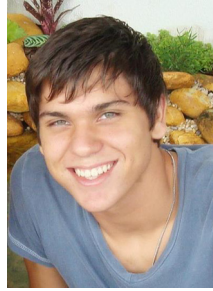
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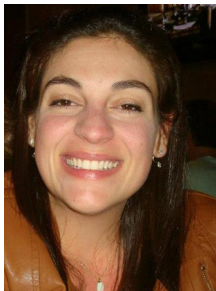
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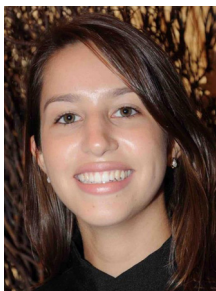
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