Alterations to Landing Technique and Patellar Tendon Loading in Response to Fatigue

SUZI EDWARDS[1], JULIE R. STEELE[2], CRAIG R. PURDAM[3], JILL L. COOK[4], and DEIRDRE E. MCGHEE[2]

[1]*School of Human Movement Studies, Charles Sturt University, New South Wales, AUSTRALIA,* [2]*Biomechanics Research Laboratory, University of Wollongong, New South Wales, AUSTRALIA;* [3]*Australian Institute of Sport, Canberra, Australian Capital Territory, AUSTRALIA; and* [4]*Department of Physiotherapy, Monash University, Clayton, Victoria, AUSTRALIA*

ABSTRACT

EDWARDS, S., J. R. STEELE, C. R. PURDAM, J. L. COOK, and D. E. MCGHEE. Alterations to Landing Technique and Patellar Tendon Loading in Response to Fatigue. *Med. Sci. Sports Exerc.*, Vol. 46, No. 2, pp. 330–340, 2014. **Purpose**: Fatigue may contribute to knee joint injuries, such as patellar tendinopathy, by increasing joint loading and altering lower limb landing technique, which in turn may increase tissue loading. This study aimed to investigate the effect of lower limb muscle fatigue on the landing technique and patellar tendon loads generated during the horizontal and vertical landing phases of a stop–jump task. It was hypothesized that muscle fatigue would increase patellar tendon loading and alter the landing technique displayed during the horizontal, but not the vertical, landing phase of the stop–jump task. **Methods**: Sixteen men, recruited from team sports involving repetitive landing, performed repeated trials of a stop–jump task. During each trial, the participants' ground reaction forces and electromyographic activity of seven lower limb muscles were recorded, three-dimensional kinematics measured, and peak patellar tendon force (F_{PT}) calculated. **Results**: When fatigued, participants generated a significantly lower F_{PT} and F_{PT} loading rate, despite a higher vertical ground reaction force (F_V) and F_V loading rate, during the horizontal landing phase of the stop–jump task. During the vertical landing phase, participants displayed only minor changes to the kinetics and kinematics of their landing in response to fatigue, although fatigue caused substantial alterations to their lower limb muscle activation patterns during landing. **Conclusions**: During the horizontal landing phase of the stop–jump task, participants decreased their patellar tendon load when they were fatigued by altering their lower limb landing technique, including a reduced net knee joint extension moment associated with less knee and hip flexion. This decrease in patellar tendon loading when fatigued may be an inherent protective strategy to potentially decrease loading of the tendon during repetitive landing. **Key Words:** PATELLAR TENDINOPATHY, STOP–JUMP TASK, INJURY MECHANICS, BIOMECHANICS

Fatigue is thought to be a major risk factor in the development of knee joint injuries (9,25,36,41) as a higher incidence of knee injuries occur toward the end of both halves (25) or in the later part of competitive team games (25,41). Altered lower limb landing strategies have been observed in healthy athletes as a consequence of being fatigued (9,16,36) as well as in athletes suffering from (7,8,24,44) or at risk of developing the overuse knee injury patellar tendinopathy (17,34). It is remains unknown, however, whether fatigue-induced alterations to an athlete's lower limb landing technique contribute to increased patellar tendon loading. This is of considerable importance because repetitive loading of the patellar tendon is thought to be a major extrinsic risk factor in developing patellar tendinopathy in sports such as soccer, basketball, and volleyball, which involve repetitive landings (13,32,45) and incur a high prevalence of patellar tendinopathy (33).

Although there has been extensive research investigating the effects of muscle fatigue on the technique used by athletes to land, the results of these studies are disparate. Between-study differences in the results are mainly due to differences in experimental design, including differences in the experimental movement task (30,46) and/or the protocol used to induce participant fatigue (20,46). Interestingly, research investigating fatigue effects (3,16,36,38) and/or possible mechanisms of patellar tendinopathy (7,16) have predominantly been restricted to examining vertical landing tasks, such as drop landings, as the experimental task. Experimental movement tasks, however, should replicate gamelike situations (5,16,23,37), and drop landings, in which the landing phase is isolated from the whole jump-landing task, have limited application as they are rarely performed in gamelike situations (16). A more common movement task in gamelike situations within many team sports is a jump-landing task (9,10,47), such as the stop–jump task, which uses the stretch–shortening cycle (SSC) muscle action. Furthermore, compared with the vertical landing phase, the horizontal landing phase of a stop–jump task places the highest load on the patellar tendon, which may potentially

Address for correspondence: Suzi Edwards, Ph.D., M.Sc., B.AppSc., School of Human Movement Studies, Charles Sturt University, Panorama Ave., Bathurst, NSW 2795, Australia; E-mail: suzedwards@csu.edu.au.
Submitted for publication February 2013.
Accepted for publication July 2013.

0195-9131/14/4602-0330/0
MEDICINE & SCIENCE IN SPORTS & EXERCISE®

DOI: 10.1249/MSS.0b013e3182a42e8e

lead to patellar tendinopathy (18). Therefore, research investigating factors affecting the lower limb landing technique of athletes, with implications for developing patellar tendinopathy, should preferably use a holistic jump-landing task that incorporates a horizontal landing phase to ensure the most appropriate experimental framework to better identify possible risk factors associated with development of patellar tendinopathy in repetitive landing sports (18).

As the effects of fatigue are task dependent (19), differences in the protocols used to induce fatigue can also contribute to disparate results relating to fatigue effects and landing technique and make between-study comparisons inappropriate. The validity of previous fatigue protocols has been limited (36) by factors such as the range of lower limb motion typically displayed during dynamic landing tasks not being replicated in the experimental design and the rest intervals tested being inconsistent with the stop–start nature of gamelike conditions. Furthermore, a fatigue protocol requiring repeated use of the SSC offers an ideal model to investigate lower limb fatigue (39) as the SSC muscle action is characteristic of running and dynamic jumping and landing movements (29,39).

This study aimed to investigate the effect of lower limb muscular fatigue on the landing technique and patellar tendon loads generated by asymptomatic athletes with normal patellar tendons during the horizontal and vertical landing phases of the stop–jump task. The experimental movement and the fatigue protocol used in the present study overcame the limitations in experimental design of previous research. That is, the experimental movement was a holistic jump-landing task that was common to sports with a high incidence of patellar tendinopathy (3,10). It involved both a horizontal and a vertical landing phase and used the SSC muscle action, and the stop–jump movement is a sporting skill commonly performed in basketball, soccer, and volleyball (10). The fatigue protocol also involved a repetitive submaximal SSC exercise with the landing phase incorporating a lower limb range of motion typically displayed during a stop–jump task and rest intervals consistent with the stop–start nature of gamelike conditions. It was hypothesized that muscular fatigue would alter the participants' landing technique and, in turn, increase their patellar tendon loading during the horizontal landing phase but not the vertical landing phase of the stop–jump task. The results of this study will increase our understanding of how fatigue-induced alterations to an athlete's landing technique contribute to patellar tendon loading, with implications for the development of overuse injuries such as patellar tendinopathy.

METHODS

Participants. Sixteen skilled male basketball (n = 7) and soccer (n = 9) athletes (mean ± SD; age = 22.4 ± 2.9 yr, height = 182.1 ± 8.7 cm, mass = 75.7 ± 10.1 kg) volunteered as study participants. With the highest prevalence of patellar tendinopathy (33), volleyball players would be the most suitable group to study. Volleyball, however, has a low participation level in Australia compared with sports such as soccer and basketball (1). As both soccer and basketball also have a high prevalence of patellar tendinopathy (33), and these sports have high participation levels within Australia (1), players from these sports were deemed suitable participants to study. All the participants were right leg dominant, based on their preferred kicking leg (22), and reported no history of traumatic lower limb injuries. Each participant's patellar tendon morphology was documented as "normal," as noted by the absence of any ultrasound abnormalities (12) detected by an experienced musculoskeletal radiologist using a 13-MHz linear array ultrasound transducer (Siemens Antares, Siemens AG, Germany). Ankle joint range of motion (4), height, body mass, and lower limb dimensions were also evaluated for each participant as descriptive characteristics of the participant cohort and for later input into inverse dynamic modeling. Written informed consent was obtained from each participant before data collection, and all the study's methods were approved by the institutions Human Research Ethics Committee (HE06/205).

Experimental protocol. No less than 1 wk before data collection, participants performed a familiarization session of the full experimental protocol, which included a 5- to 10-min warm-up, a familiarization of the stop–jump task and the fatigue protocol, and completion of a full trial of the fatigue protocol. The participants then returned on a second occasion for data to be collected during the full experimental protocol, which involved the warm-up, followed by standardized trials of the stop–jump task both before and after the fatigue protocol (Fig. 1).

Experimental task. The stop–jump task consisted of a horizontal landing phase, immediately followed by a vertical landing phase (15). The horizontal landing phase required the participants to accelerate forward for four steps toward two force platforms, to stop and perform a simultaneous two-foot landing with their right and left feet contacting separate force platforms. The vertical landing phase immediately followed and required the participants to jump vertically upward to strike a ball suspended from the ceiling with both hands and to then perform a simultaneous two-foot landing a second time, with each foot again contacting separate force platforms. During the stop–jump task familiarization, the effort at which the participants performed the task was standardized by using a set starting position away from the force platform. The participants' average approach speed was measured immediately before the preparation phase of the first landing, using infrared timing lights (OnSpot, University of Wollongong). Jump height effort was standardized among the participants by positioning the ball at the maximum height in which each participant could touch the ball with both hands after performing the stop–jump task during task familiarization.

Fatigue protocol. The protocol used to induce lower limb muscular fatigue in the present study required the participants to perform three maximal SSC exercises (described in the next paragraph), followed by repeatedly performing

APPLIED SCIENCES

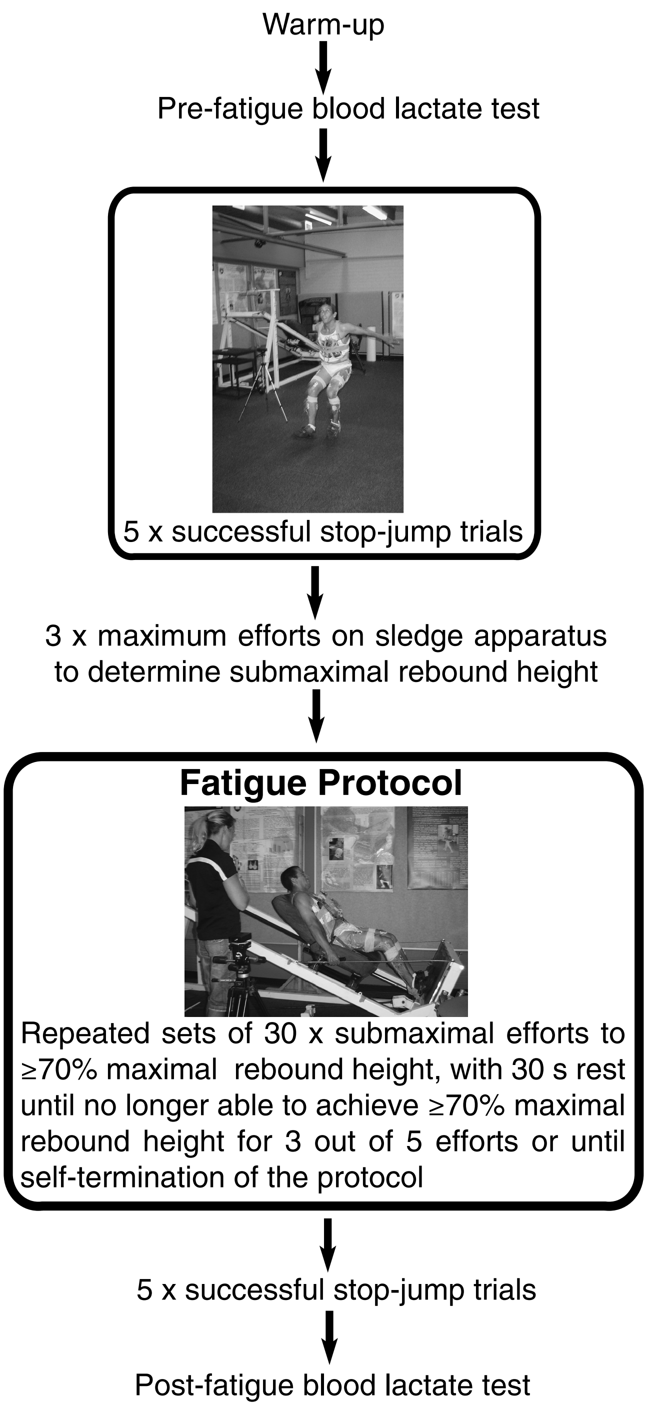

FIGURE 1—Flow chart illustrating the experimental protocol.

sets of 30 submaximal SSC exercises, interspersed with a 30-s rest, on a custom-built sledge apparatus. Fatigue was deemed when the participant could no longer achieve 70% of their maximal rebound height for three out of five submaximal SSC exercises or when participants self-terminated the sets because they felt that they could no longer continue. On the basis of blood lactate changes previously observed during a soccer (31) or basketball game (35), an increase in postfatigue blood lactate levels to at least 6 mmol·L^{-1}, analyzed using an Accusport blood lactate analyzer (Boehringer, Mannheim, Germany), was also measured at the end of the experimental protocol to confirm fatigue.

The SSC exercises were performed on a custom-built 23-kg sledge apparatus seat, which glided along a track inclined 23.6° from the horizontal (Fig. 1). To ensure validity of the fatigue protocol, each participant was seated so their hip flexion angle (40°) replicated hip joint angles typically displayed during the landing phase of a stop–jump task (18). For each maximal SSC exercise, the participants performed two submaximal SSC efforts, immediately followed by a maximal SSC effort. Each SSC effort required the participants to begin with both their feet contacting a metal base plate that was secured to the base frame of the sledge apparatus. From this starting position, the participants pushed off the metal base plate and accelerated upward along the gliding track of the sledge. For the maximal and submaximal SSC efforts, maximal rebound height or submaximal rebound height of at least 70% of their maximal rebound height was required (29,30), respectively, before the participant glided down the track to stop in a two-foot landing, with each foot again contacting the metal base plate. Maximum knee flexion (75°) during the landing phase of the SSC efforts replicated angles typically displayed during the landing phase of a stop–jump task (15,18). To ensure participants met the 75° maximum knee flexion criterion and reached 70% of their maximal SSC exercise rebound height during the entire SSC efforts (fatigue protocol), they were given real-time verbal feedback from the investigator based on two marks, which were set on the frame of the sledge apparatus and indicated that the participant had achieved each criterion.

Experimental procedures. During each stop–jump trial, the ground reaction forces generated during each landing were recorded (1000 Hz) using two multichannel force platforms (Type 9281B and Type 9253B; Kistler, Winterthur, Switzerland) embedded in the floor, with each platform connected to a multichannel charge amplifier (Type 9865A and Type, 9865B; Kistler). The participant's three-dimensional lower limb motion was recorded (100 Hz) using an Optotrak® 3020 motion analysis system (Northern Digital, Waterloo, Canada). Infrared light-emitting diodes were placed on each participant's dominant lower limb and pelvis, on the shoe at the first and fifth metatarsal head and mid-anterior foot, lateral malleolus, medial malleolus, lateral leg, anterior distal leg, anterior proximal leg, lateral femoral epicondyle, medial femoral epicondyle, lateral femur, anterior distal femur, anterior proximal femur, greater trochanter, anterior superior iliac spine, and iliac crest. To avoid losing view of the infrared light-emitting diodes, the participants wore minimal clothing (a T-shirt and shorts), and their own socks and athletic shoes were worn during testing.

Electromyographic activity was recorded bilaterally for vastus lateralis (VL), rectus femoris (RF), vastus medialis (VM), biceps femoris (BF), semitendinosus (ST), tibialis anterior (TA), and medial gastrocnemius (MG) using two TeleMyo systems (Noraxon, Scottsdale, AZ). Following standard preparation (2), bipolar silver–silver-chloride surface electrodes (Ambu® Blue Sensor M, electrode size = 40.8 × 32 mm, detection area = 13.2 mm^2; Noblesville, IN) were placed longitudinally on each muscle belly (interelectrode distance of 20 mm). A common reference electrode was located on the tibial tuberosity of each lower limb. The electromyographic signals for each lower limb were sampled (1000 Hz; bandwidth, 16–500 Hz) and relayed from two TeleMyo 900 battery powered transmitters (Noraxon), firmly fixed around the participant's waist, to two TeleMyo 900 receivers. Kinetic, kinematic, and electromyographic data were time synchronized and collected using First Principles software (Version 1.00.2; Northern Digital).

Data reduction. Analysis of the kinematic and kinetic data was performed using Visual 3D software (Version 3; C-Motion, Germantown, MD). The raw ground reaction force data were initially filtered using a fourth-order zero-phase-shift Butterworth digital low-pass filter (f_c = 50 Hz) before calculating the ground reaction force variables. The raw kinematic coordinates, ground reaction forces, free moments, and center of pressure data were then filtered using a fourth-order zero-phase-shift Butterworth digital low-pass filter (f_c = 18 Hz) before calculating individual joint kinematics, net internal knee extension–flexion joint moments, and patellar tendon forces (6). The patellar tendon forces were calculated by dividing the net internal knee extension–flexion joint moment by the patellar tendon moment arm (40). Using a polynomial regression equation, the patellar tendon moment arm was estimated as a function of knee joint angle using the method of Herzog and Read (26).

The raw electromyographic signals were filtered using a fourth-order zero-phase shift Butterworth (high-pass f_c = 15 Hz) to eliminate any movement artifact. To quantify temporal characteristics of the muscle bursts, the filtered electromyographic data were full-wave rectified, filtered with a 20-Hz fourth-order zero-phase shift Butterworth low-pass filter, and then full-wave rectified to create linear envelopes using customized software (LabVIEW 8; National Instruments, Austin, TX). The time of the onset of each muscle burst activity was determined using a threshold crossing of 8% of the maximum amplitude of the linear envelope for each muscle (15). Each individual muscle's filtered signal was visually inspected to confirm the validity of the calculated results for the temporal characteristics of the muscle bursts to minimize the probability of error.

Data analysis. The two landing phases within the stop–jump task were firstly identified from the vertical ground reaction force trace: (i) the "horizontal" landing phase and (ii) the "vertical" landing phase (15). The primary outcome variable analyzed during these two landing phases was the peak patellar tendon force (F_{PT}). The F_{PT} calculated for each participant's right (dominant) lower limb was selected for analysis as asymmetry between lower limbs in F_{PT} has been observed during the horizontal landing phase of a stop–jump task (15). Secondary variables analyzed during the two landing phases included the peak vertical ground reaction force (F_V) and peak anterior–posterior ground reaction force (F_{AP}); ankle, knee, and hip joint kinematics; the net peak knee joint moment; and the time of the onset and peak muscle activity of each of the seven lower limb muscles relative to the time of the F_{PT} in each landing phase. The loading rate of the F_V (LR F_V, body weight per second) was calculated by dividing the F_V by the time interval between initial foot–ground contact (IC) and the time of the F_V. The loading rate of the F_{PT} (LR F_{PT}, body weight per second) was calculated by dividing the F_{PT} by the time interval between IC and the time of the F_{PT}. The temporal events (IC, and at the time of the peak F_V and F_{AP}) were defined using the 18-Hz filtered kinetic data, with initial contact defined when the ground reaction force exceeded 30 N. The jump height (cm) attained by each participant during the stop–jump task was also calculated as the difference in the maximum vertical displacement of the greater trochanter marker minus the vertical displacement of the same marker measured while each participant stood motionless.

Statistical analysis. Mean and SD values were calculated for each kinetic, kinematic, and muscle activity variable during the horizontal and vertical landing phases in a non-fatigued (NF) and fatigued (F) condition. The data were then analyzed using a series of paired-samples *t*-tests to determine whether there were any substantial differences in the primary and secondary outcome variables between the two fatigued conditions. Magnitude-based inferences on the differences in the mean changes between the two fatigue conditions (NF and F) and precision of estimation were then used to avoid the shortcomings of research based on null-hypothesis significance testing (28). To show the magnitude of the effects, effect sizes were calculated and magnitudes were assessed using the following criteria: 0.19 = trivial, 0.20–0.49 = small, 0.50–0.79 = moderate, and >0.80 = large (11). Clear moderate or large effect sizes were defined as substantial. The 95% confidence limits, which define the range representing the uncertainty in the true value of the (unknown) population mean, were used to indicate the precision of estimates. Although the chance of inflation of a type I error increases when multiple statistical tests are conducted, no adjustment to the alpha level was deemed necessary (28), as such adjustments may increase the likelihood of type II errors (42). All statistical procedures were conducted using the Statistical Package for the Social Sciences (Version 15; SPSS Inc., Chicago, IL) and a customized Excel spreadsheet (27).

RESULTS

Fatigue Variables

During the fatigue protocol, participants performed on average 14.5 ± 9.6 submaximal SSC sets and displayed a significant increase in postfatigue blood lactate level above the 6-mmol·L^{-1} fatigue criterion (NF = 3.5 ± 0.7 mmol·L^{-1}, F = 8.0 ± 1.9 mmol·L^{-1}, d = 6.76, P < 0.001, 95% confidence interval [CI] = 2.7).

APPLIED SCIENCES

Stop–Jump Task Variables

Despite attempts to standardize both the approach speed and vertical jump height achieved during the stop–jump task, fatigue led to a significant decrease in the participants' mean approach speed (NF = 4.5 ± 0.4 $m{\cdot}s^{-1}$, F = 4.1 ± 0.4 $m{\cdot}s^{-1}$, d = 0.93, P < 0.001, 95% CI = 0.2). Despite being significant, the decrease in the mean vertical jump height attained (NF = 48 ± 6 cm, F = 45 ± 6 cm, d = 0.47, P = 0.02, 95% CI = 10) during the stop–jump task was deemed only trivial due to the small effect size and high 95% CI.

Patellar tendon loading. During the horizontal landing phase of the stop–jump task, participants generated a significantly lower F_{PT}, LR F_{PT}, and peak knee joint moment when fatigued compared with when nonfatigued (Table 1). In contrast, during the vertical landing phase, there were no significant differences between fatigue conditions in patellar tendon loading.

Ground reaction forces. When fatigued during the horizontal landing phase, the participants displayed a significantly higher F_V and LR F_V compared with when not fatigued. In addition, although participants used significantly shorter time from IC-F_V when fatigued compared with when nonfatigued, the magnitude of the absolute difference was functionally irrelevant due to the small effect size and high 95% CI. During the vertical landing phase of the stop–jump task, the only significant difference was a significant increase in F_{AP} during the fatigued condition compared with the nonfatigued condition (Table 1).

Horizontal landing phase joint kinematic data. During the horizontal landing phase of the stop–jump task, the participants maintained similar ankle joint positions from IC to the time of the F_{PT} during both the fatigued and the nonfatigued conditions (Table 2). In fact, the only significant difference in ankle kinematics that the participants displayed when fatigued was a slower ankle inversion velocity at the time of the F_V compared with when nonfatigued. However, participants landed with significantly less knee flexion throughout the horizontal landing phase (IC, F_V, and F_{PT}) when fatigued compared with when nonfatigued, although at IC the magnitude of the absolute difference was deemed trivial due to the small effect size. In contrast, the range of knee joint motion from the time of IC to the time of F_{PT} (NF = 50° ± 12°, F = 50° ± 10°, d = 0.03, P = 0.805, 95% CI = 6.4) was similar between fatigue conditions. When fatigued, the participants displayed significantly less hip flexion throughout the horizontal landing phase (IC, F_V, and F_{PT}), slower hip flexion velocity at IC, and faster hip adduction velocity at IC when they were fatigued.

Vertical landing phase joint kinematic data. During the vertical landing phase of the stop–jump task, the participants displayed a significantly less plantar flexed position of the ankle and a higher plantarflexion velocity at IC when fatigued compared with when nonfatigued (Table 3). Furthermore, the participants underwent a significantly smaller ankle plantarflexion-dorsiflexion range of motion from the time of IC to the time of the F_{PT} (NF = 55° ± 7°, F = 49° ± 8°, d = 0.80, large, P = 0.005, 95% CI = 6) when fatigued. A significantly slower knee flexion velocity at IC, a slower knee adduction velocity at IC, and a faster knee abduction velocity at both the times of the F_V were also all evident during the fatigued condition compared with the nonfatigued condition. However, the only between-fatigue condition difference in hip kinematics was hip abduction at IC, but the magnitude of the absolute difference was considered functionally irrelevant due to the small effect size.

Muscle activation patterns. During the horizontal landing phase, the participants used a relatively similar muscle recruitment strategy to stabilize their lower limbs during both the nonfatigued and fatigued conditions (Table 4). Nevertheless, when landing horizontally, participants displayed a later peak muscle burst activity for MG, BF, and VL (moderate) relative to the time of the F_{PT} during the fatigued condition compared with the nonfatigued condition. In contrast, during the vertical landing phase, the participants used a relatively different muscle activation pattern strategy to stabilize their lower limbs when fatigued compared with when nonfatigued (Table 5). That is, when fatigued, participants displayed a

TABLE 1. Effect of fatigue on the forces (normalized to body weight) generated during the horizontal and vertical landing phases of a stop–jump task.

	Horizontal Landing Phase					Vertical Landing Phase				
	Nonfatigued	Fatigued	d^a	CI	P	Nonfatigued	Fatigued	d	CI	P
F_V (body weight)	2.32 ± 0.61	2.63 ± 0.64	0.51^b	0.19	0.00^c	3.07 ± 0.78	2.94 ± 0.80	0.16^d	0.28	0.35^c
F_{AP} (body weight)	−1.07 ± 0.35	−1.13 ± 0.29	0.18^b	0.13	0.34^c	−0.46 ± 0.14	−0.58 ± 0.10	0.82^d	0.06	0.00^c
F_{PT} (body weight)	6.76 ± 1.61	5.53 ± 1.63	0.76^b	0.94	0.01^c	5.30 ± 1.14	4.72 ± 0.91	0.51^b	0.69	0.09^c
Knee moment ($N{\cdot}m{\cdot}kg^{-1}$)	3.18 ± 0.80	2.72 ± 0.76	0.57^b	0.38	0.02^c	2.41 ± 0.40	2.25 ± 0.40	0.40^d	0.23	0.15^c
LR F_V (body weight per second)	83 ± 39	106 ± 52	0.58^b	12	0.00^c	60 ± 29	62 ± 31	0.00^d	10	0.73^c
LR F_{PT} (body weight per second)	93 ± 22	78 ± 29	0.68^b	12	0.02^c	57 ± 16	51 ± 16	0.38^d	8	0.11^c
IC-F_V (ms)	34 ± 12	29 ± 7	0.39^b	4	0.01^c	60 ± 16	58 ± 15	0.12^d	6	0.49^c
IC-F_{AP} (ms)	40 ± 13	37 ± 12	0.26^b	4	0.10^c	14 ± 3	14 ± 2	0.15^d	1	0.48^c
IC-F_{PT} (ms)	76 ± 16	74 ± 10	0.17^b	6	0.30^c	92 ± 17	97 ± 19	0.29^d	7	0.13^c

Data are presented as mean ± SD. Peak vertical ground reaction force (F_V), peak anterior–posterior ground reaction force (F_{AP}), peak patellar tendon forces (F_{PT}), loading rate of the F_V (LR F_V), loading rate of the F_{PT} (LR F_{PT}), time interval between initial foot–ground contact (IC) to the time of the F_V (IC-F_V), time interval between IC to the time of the F_{AP} (IC-F_{AP}), time interval between IC to the time of the F_{PT} (IC-F_{PT}), and 95% CI defines the range representing the uncertainty in the true value of the (unknown) population mean.
[a]Effect size.
[b]Moderate between-fatigue condition difference in the effect size (value, 0.50–0.79).
[c]Significant between-fatigue condition difference, P < 0.05.
[d]Large between-fatigue condition difference in the effect size (value, ≥0.80).

TABLE 2. Effect of fatigue on the joint angles (°) and velocities (°·s^{-1}) displayed during the horizontal landing phase of a stop–jump task.

	At the Time of IC					At the Time of the Peak F_V					At the Time of the Peak F_{PT}				
Variables	Nonfatigued	Fatigued	d[a]	CI	P	Nonfatigued	Fatigued	d	CI	P	Nonfatigued	Fatigued	d	CI	P
Joint angles (°)															
Ankle dorsiflexion	65.5 ± 20.1	67.1 ± 18.3	0.08[b]	3.6	0.35[c]	59.3 ± 8.6	58.4 ± 7.8	0.10[b]	2.4	0.45[c]	60.3 ± 17.0	58.8 ± 16.9	0.09[b]	2.3	0.18[c]
Forefoot adduction	5.0 ± 7.2	4.7 ± 6.6	0.03[b]	2.3	0.84[c]	4.1 ± 6.3	5.7 ± 8.8	0.25[b]	2.2	0.14[c]	−1.3 ± 5.8	−2.9 ± 9.8	0.28[b]	5.7	0.56[c]
Ankle eversion	−16.1 ± 5.9	−13.9 ± 8.0	0.37[b]	2.5	0.09[c]	−11.7 ± 8.1	−12.1 ± 7.3	0.05[b]	2.4	0.74[c]	−17.0 ± 18.6	−14.0 ± 12.9	0.16[b]	3.9	0.12[c]
Knee flexion	30.4 ± 6.3	27.7 ± 4.4	0.44[b]	2.3	0.02[c]	46.7 ± 6.1	38.5 ± 3.7	1.35[d]	3.1	0.00[c]	62.5 ± 5.8	58.5 ± 5.8	0.68[b]	3.5	0.03[c]
Knee adduction	3.7 ± 4.0	2.2 ± 5.1	0.37[b]	2.5	0.21[c]	2.3 ± 7.7	0.5 ± 6.9	0.24[b]	3.2	0.24[c]	0.0 ± 7.8	−3.2 ± 9.6	0.40[b]	6.0	0.28[c]
Knee internal rotation	−10.4 ± 9.6	−9.6 ± 9.2	0.08[b]	4.0	0.68[c]	−7.1 ± 8.2	−7.4 ± 10.0	0.03[b]	4.5	0.90[c]	−3.0 ± 7.1	−1.3 ± 9.8	0.23[b]	5.2	0.51[c]
Hip flexion	40.5 ± 7.6	36.5 ± 5.2	0.52[b]	2.9	0.01[c]	43.5 ± 8.9	36.5 ± 8.7	0.78[b]	3.0	0.00[c]	46.8 ± 9.9	42.6 ± 10.1	0.68[b]	3.5	0.03[c]
Hip adduction	−8.1 ± 6.2	−7.1 ± 4.7	0.15[b]	3.7	0.57[c]	−7.5 ± 5.7	−5.9 ± 5.1	0.28[b]	2.4	0.16[c]	−4.1 ± 4.5	−2.2 ± 3.7	0.40[b]	6.0	0.28[c]
Hip internal rotation	1.4 ± 11.9	0.2 ± 11.5	0.10[b]	5.7	0.64[c]	3.5 ± 9.3	0.1 ± 12.2	0.37[b]	5.7	0.21[c]	4.7 ± 8.8	0.1 ± 14.0	0.23[b]	5.2	0.51[c]
Joint velocities (°·s^{-1})															
Ankle dorsiflexion	−87 ± 89	−170 ± 292	0.93[d]	134	0.21[c]	−27 ± 399	−114 ± 493	0.22[b]	120	0.14[c]	270 ± 76	286 ± 74	0.21[b]	36	0.37[c]
Forefoot adduction	−2 ± 76	22 ± 78	0.32[b]	37	0.18[c]	−149 ± 248	−125 ± 206	0.10[b]	64	0.41[c]	−81 ± 90	−68 ± 49	0.14[b]	44	0.55[c]
Ankle eversion	32 ± 48	58 ± 72	0.55[b]	38	0.16[c]	−136 ± 136	−54 ± 173	0.60[b]	66	0.02[c]	29 ± 115	2 ± 128	0.23[b]	63	0.38[c]
Knee flexion	159 ± 103	121 ± 77	0.37[b]	54	0.15[c]	564 ± 152	524 ± 194	0.27[b]	77	0.28[c]	500 ± 165	526 ± 108	0.16[b]	65	0.40[c]
Knee adduction	−62 ± 124	−42 ± 62	0.16[b]	55	0.45[c]	−34 ± 135	−59 ± 138	0.18[b]	58	0.37[c]	−35 ± 96	−48 ± 95	0.14[b]	54	0.61[c]
Knee internal rotation	35 ± 79	27 ± 74	0.10[b]	40	0.67[c]	86 ± 82	106 ± 150	0.24[b]	79	0.60[c]	−56 ± 90	−7 ± 146	0.53[b]	78	0.20[c]
Hip flexion	99 ± 136	9 ± 80	0.66[b]	87	0.05[c]	77 ± 134	91 ± 157	0.11[b]	78	0.69[c]	10 ± 114	87 ± 131	0.56[b]	70	0.08[c]
Hip adduction	−42 ± 118	22 ± 123	0.54[b]	64	0.05[c]	29 ± 93	59 ± 81	0.32[b]	70	0.38[c]	43 ± 69	−22 ± 142	0.92[b]	89	0.16[c]
Hip internal rotation	41 ± 81	−20 ± 147	0.74[b]	95	0.18[c]	119 ± 196	31 ± 115	0.33[b]	78	0.40[c]	35 ± 38	136 ± 321	2.48[b]	186	0.31[c]

Data are presented as mean ± SD. Initial foot–ground contact (IC), peak vertical ground reaction force (F_V), peak patellar tendon forces (F_{PT}), and 95% CI defines the range representing the uncertainty in the true value of the (unknown) population mean. For the above rotations: ankle dorsiflexion, forefoot adduction, ankle eversion, knee flexion, knee adduction, knee internal rotation, hip flexion, hip adduction, and hip internal rotation are positive.

[a] Effect size.

[b] Moderate between-fatigue condition difference in the effect size (value, 0.50–0.79).

[c] Significant between-fatigue condition difference, $P < 0.05$.

[d] Large between-fatigue condition difference in the effect size (value, ≥0.80).

APPLIED SCIENCES

TABLE 3. Effect of fatigue on the joint angles (°) and velocities (°·s^{-1}) displayed during the vertical landing phase of a stop–jump task.

	At the Time of IC					At the Time of the Peak F_V					At the Time of the Peak F_{PT}				
Variables	**Nonfatigued**	**Fatigued**	d^a	**CI**	P	**Nonfatigued**	**Fatigued**	d	**CI**	P	**Nonfatigued**	**Fatigued**	d	**CI**	P
Joint angles (°)															
Ankle dorsiflexion	37.1 ± 8.0	41.1 ± 7.9	0.51[b]	2.6	0.00[c]	83.2 ± 8.4	83.1 ± 7.9	0.01	4.3	0.95[c]	79.9 ± 21.9	82.4 ± 18.6	0.12	4.0	0.20[c]
Forefoot adduction	4.9 ± 3.8	4.9 ± 4.3	0.00[b]	1.4	0.99[c]	−2.9 ± 7.4	−2.8 ± 7.9	0.02	2.7	0.89[c]	−6.9 ± 9.8	−6.0 ± 9.4	0.10	3.4	0.57[c]
Ankle eversion	−11.0 ± 6.3	−11.5 ± 5.3	0.08[b]	1.5	0.47[c]	−13.4 ± 6.7	−13.6 ± 4.4	0.04	2.3	0.81[c]	−18.1 ± 18.8	−18.5 ± 14.3	0.02	3.9	0.83[c]
Knee flexion	19.3 ± 6.2	20.3 ± 6.1	0.17[b]	2.1	0.31[c]	46.8 ± 9.5	43.9 ± 8.4	0.30	3.0	0.06[c]	58.6 ± 10.9	60.1 ± 10.0	0.13	5.3	0.58[c]
Knee adduction	2.2 ± 2.6	2.4 ± 2.6	0.04[b]	1.1	0.82[c]	1.4 ± 4.3	1.3 ± 4.7	0.03	1.7	0.86[c]	1.1 ± 4.9	0.0 ± 4.4	0.22	1.6	0.18[c]
Knee internal rotation	−12.4 ± 6.8	−12.3 ± 6.6	0.03[b]	2.0	0.85[c]	−4.8 ± 7.8	−4.2 ± 8.3	0.08	2.2	0.54[c]	−3.4 ± 7.4	−3.1 ± 7.3	0.03	4.1	0.90[c]
Hip flexion	14.9 ± 5.2	12.6 ± 6.2	0.44[b]	2.9	0.10[c]	22.4 ± 8.0	19.7 ± 7.1	0.34	3.9	0.16[c]	29.3 ± 9.9	28.6 ± 9.7	0.13	5.3	0.58[c]
Hip adduction	−8.3 ± 2.4	−7.1 ± 3.4	0.48[b]	1.1	0.05[c]	−8.4 ± 4.1	−7.2 ± 4.6	0.30	1.4	0.09[c]	−7.6 ± 4.1	−6.0 ± 4.9	0.22	1.6	0.18[c]
Hip internal rotation	−2.9 ± 5.9	−1.7 ± 5.3	0.21[b]	1.9	0.20[c]	−2.8 ± 5.7	−3.0 ± 6.1	0.02	2.3	0.91[c]	−0.5 ± 5.7	−0.9 ± 5.7	0.03	4.1	0.90[c]
Joint velocities (°·s^{-1})															
Ankle dorsiflexion	−25 ± 73	−103 ± 90	1.08[d]	53	0.01[c]	341 ± 150	319 ± 178	0.14[d]	87	0.60[c]	163 ± 239	73 ± 110	0.38[b]	134	0.17[c]
Forefoot adduction	−21 ± 58	−6 ± 40	0.25[b]	31	0.33[c]	−138 ± 88	−113 ± 157	0.28[d]	83	0.53[c]	62 ± 411	58 ± 395	0.01[b]	313	0.98[c]
Ankle eversion	17 ± 59	14 ± 35	0.05[b]	34	0.85[c]	−49 ± 72	−31 ± 116	0.24[d]	66	0.59[c]	−109 ± 393	−6 ± 190	0.26[b]	233	0.36[c]
Knee flexion	157 ± 66	74 ± 61	1.27[d]	28	0.00[c]	476 ± 144	469 ± 145	0.05[d]	71	0.84[c]	420 ± 128	381 ± 116	0.31[b]	61	0.19[c]
Knee adduction	24 ± 31	4 ± 21	0.55[b]	14	0.02[c]	−18 ± 58	−82 ± 101	1.10[d]	38	0.00[c]	61 ± 131	−22 ± 125	0.64[b]	88	0.06[c]
Kneeinternalrotation	32 ± 62	17 ± 57	0.24[b]	35	0.38[c]	75 ± 142	34 ± 80	0.29[d]	83	0.30[c]	50 ± 289	−81 ± 131	0.45[b]	182	0.14[c]
Hip flexion	50 ± 55	33 ± 56	0.30[b]	26	0.19[c]	233 ± 114	185 ± 154	0.42[d]	80	0.22[c]	210 ± 132	116 ± 157	0.71[b]	98	0.06[c]
Hip adduction	−8 ± 28	−7 ± 23	0.05[b]	17	0.86[c]	34 ± 46	40 ± 82	0.14[b]	47	0.78[c]	−9 ± 82	−4 ± 88	0.07[b]	61	0.84[c]
Hip internal rotation	23 ± 39	23 ± 44	0.00[b]	26	1.00[c]	−8 ± 99	−6 ± 72	0.02[b]	58	0.95[c]	61 ± 93	110 ± 113	0.53[b]	75	0.18[c]

Data are presented as mean ± SD. Initial foot–ground contact (IC), peak vertical ground reaction force (F_V), peak patellar tendon forces (F_{PT}), and 95% CI defines the range representing the uncertainty in the true value of the (unknown) population mean. For the above rotations: ankle dorsiflexion, forefoot adduction, ankle eversion, knee flexion, knee adduction, knee internal rotation, hip flexion, hip adduction, and hip internal rotation are positive.

[a]Effect size.

[b]Moderate between-fatigue condition difference in the effect size (value, 0.50–0.79).

[c]Significant between-fatigue condition difference, $P < 0.05$.

[d]Large between-fatigue condition difference in the effect size (value, ≥0.80).

TABLE 4. Effect of fatigue on the times of the onset of muscle activation and the peak muscle activity relative to the time of the peak patellar tendon force (F_{PT}) generated during the horizontal landing phase of a stop–jump task.

	Onset of Muscle Activation Relative to the Time of F_{PT}					Peak Muscle Activity Relative to the Time of F_{PT}				
Muscle (ms)	Nonfatigued	Fatigued	d^a	CI	P	Nonfatigued	Fatigued	d	CI	P
Tibialis anterior	−137 ± 47	−132 ± 51	0.14	47	0.55	−30 ± 37	−50 ± 49	0.55[b]	35	0.25[c]
Medial gastrocnemius	−119 ± 63	−95 ± 43	0.38	28	0.09	−42 ± 57	2 ± 42	0.78[b]	37	0.03[c]
Biceps femoris	−103 ± 35	−97 ± 37	0.18	29	0.59	−27 ± 37	17 ± 32	1.21[d]	31	0.01[c]
Semitendinosus	−123 ± 30	−133 ± 46	0.34	30	0.47	−47 ± 29	−62 ± 44	0.52[b]	25	0.21[c]
Vastus medialis	−98 ± 29	−99 ± 20	0.02	16	0.95	4 ± 23	10 ± 36	0.26[b]	21	0.55[c]
Rectus femoris	−93 ± 29	−82 ± 15	0.38	17	0.18	5 ± 30	19 ± 18	0.50[b]	23	0.22[c]
Vastus lateralis	−90 ± 18	−98 ± 16	0.45	15	0.26	4 ± 27	19 ± 28	0.66[b]	16	0.05[c]

Data are presented as mean ± SD. Peak patellar tendon forces (F_{PT}), and 95% CI defines the range representing the uncertainty in the true value of the (unknown) population mean. Negative values indicate muscle activation variable occurred before F_{PT}.
[a]Effect size.
[b]Moderate between-fatigue condition difference in the effect size (value, 0.50–0.79).
[c]Significant between-fatigue condition difference, $P < 0.05$.
[d]Large between-fatigue condition difference in the effect size (value, ≥0.80).

later onset of TA, BF, and ST and a later peak muscle burst activity for TA, MG, and BF relative to the time of the F_{PT} activity compared with when nonfatigued.

DISCUSSION

The primary extrinsic risk factor associated with patellar tendinopathy is repetitive landing (13,32,45). We originally hypothesized that fatigue would exacerbate this risk by increasing the magnitude of the repetitive loads sustained by the patellar tendon during the horizontal landing phase but not the vertical landing phase of the stop–jump task. In partial support of our hypothesis, fatigue did not alter patellar tendon loading during the vertical landing phase. In contrast to our hypothesis, however, when landing horizontally, both the F_{PT} and the LR F_{PT} significantly decreased during the fatigued condition, despite significant increases in both the F_V and the LR F_V generated during the same condition. Suggested reasons for this unexpected finding are discussed below. This result, however, confirms previous research (18) that F_V is not a reliable indicator of F_{PT}. That is, if F_V had been used as a surrogate of F_{PT} in the present study, it would have been errantly concluded that fatigue increased patellar tendon loading during the horizontal landing phase, when fatigue in fact reduced patellar tendon loading. It should be noted that this unexpected finding of a decrease in patellar tendon loading in athletes with normal patellar tendons may not be observed in asymptomatic athletes who are at risk of developing patellar tendinopathy (17,34) because they might be less able to adapt to changes evoked by fatigue, leading to either an unchanged or increased patellar tendon loading.

Most of the significant between-fatigue condition differences were evident during the horizontal landing phase rather than the vertical landing phase of the stop–jump task. This finding highlights the limitation of previous studies that have only used landings with a predominantly vertical motion as the experimental movement to examine lower limb landing strategies in relation to patellar tendinopathy (7,8,43,44). The lower magnitude of patellar tendon force during the vertical landing phase (18) may explain why most of the fatigue effects were observed during the horizontal landing phase. It also helps to explain some of the discrepancies in results previously found in research investigating how fatigue affects landing technique.

Consistent with previous research of a stop–jump task, participants in the present study displayed significantly less knee flexion when they were fatigued compared with when they were nonfatigued during the horizontal landing (9) and during an SSC exercise (29). This reduction in knee flexion during the horizontal landing phase of a stop–jump task when fatigued may be a compensatory effect of fatigue, in which the participants are unable to generate an adequate net knee extension moment upon landing to prevent the knee from collapsing. By landing with the knee more extended, there

TABLE 5. Effect of fatigue on the times of the onset of muscle activation and the peak muscle activity relative to the time of the peak patellar tendon force (F_{PT}) generated during the vertical landing phase of a stop–jump task.

	Onset of Muscle Activation Relative to the Time of F_{PT}					Peak Muscle Activity Relative to the Time of F_{PT}				
Muscle (ms)	Nonfatigued	Fatigued	d^a	CI	P	Nonfatigued	Fatigued	d	CI	P
Tibialis anterior	−99 ± 34	−70 ± 46	0.84[b]	26	0.04[c]	−15 ± 35	13 ± 32	0.80[b]	23	0.02[c]
Medial gastrocnemius	−149 ± 35	−146 ± 44	0.11[d]	20	0.68[c]	−81 ± 31	−51 ± 39	0.96[b]	22	0.01[c]
Biceps femoris	−109 ± 30	−82 ± 38	0.90[b]	17	0.00[c]	−43 ± 35	−6 ± 34	1.04[b]	25	0.01[c]
Semitendinosus	−133 ± 53	−92 ± 49	0.76[d]	24	0.00[c]	−50 ± 69	−22 ± 63	0.41[d]	33	0.08[c]
Vastus medialis	−101 ± 21	−101 ± 21	0.00[d]	15	1.00[c]	−9 ± 36	−4 ± 18	0.16[d]	14	0.40[c]
Rectus femoris	−95 ± 38	−88 ± 22	0.17[d]	17	0.42[c]	−16 ± 40	5 ± 31	0.52[d]	30	0.15[c]
Vastus lateralis	−96 ± 22	−97 ± 17	0.05[d]	12	0.85[c]	−12 ± 27	−6 ± 29	0.20[d]	17	0.49[c]

Data are presented as mean ± SD. Peak patellar tendon forces (F_{PT}) and 95% CI define the range representing the uncertainty in the true value of the (unknown) population mean. Negative values indicate muscle activation variable occurred before F_{PT}.
[a]Effect size.
[b]Large between-fatigue condition difference in the effect size (value, ≥0.80).
[c]Significant between-fatigue condition difference, $P < 0.05$.
[d]Moderate between-fatigue condition difference in the effect size (value, 0.50–0.79).

APPLIED SCIENCES

was less reliance on generating a high knee extensor moment when fatigued to stabilize the knee upon landing. As the direction of the patellar tendon force is a function of knee joint angle (26), landing with less knee flexion results in less compressive load on the patellar tendon.

Irrespective of this reduced knee flexion, using the SSC during a movement task increases the F_{PT} because of the countermovement, which passively and then actively lengthens the muscles to take advantage of the muscle's length–tension properties to enhance velocity and power in the subsequent takeoff phase (21). Our previous research has highlighted that in a stop–jump task, the need to prepare to perform a vertical jump immediately after the horizontal landing phase may cause the participants to use greater knee flexion than would typically be used in an isolated horizontal landing movement, in which the knee joint is relatively extended at IC (18). Furthermore, fatigue of the SSC causes participants to land with less knee flexion at IC, decreases the knee extensor positive peak power, and decreases the performance of an SSC exercise (29). We speculate that in the present study, landing with less knee flexion during the horizontal landing phase when fatigued reduced the participant's efficiency in using the SSC muscle action during the subsequent takeoff phase, reflected in the decrease in net knee joint extension moment, but would have protected the knee by significantly reducing the magnitude of the patellar tendon load.

The F_{PT} is derived by dividing the net knee extension–flexion joint moment by the patellar tendon moment arm and thus can be influenced by changes in knee flexion used in an activity. In this present study, variations in the patellar tendon moment arm could partially explain the between-fatigue condition differences in the F_{PT} during the horizontal landing phase. That is, as the peak patellar tendon moment arm occurs at 29° of knee flexion (26), by the participants landing when fatigued with less knee flexion and at an angle that used a larger patellar tendon moment arm at the times of the F_V and F_{PT}, this could have, in turn, contributed to the decreased F_{PT} and LR F_{PT} compared with when nonfatigued. However, the significantly lower knee joint moment generated when fatigued was the primary contributor to the significantly lower F_{PT} found during the horizontal landing phase compared with when nonfatigued. This decrease in knee joint moment could not be attributed to the higher F_V and LR F_V when fatigued compared with when nonfatigued, as these both increased and would consequently increase the knee joint moment. We speculate that the lower knee joint moment was attributed to the change in the orientation and position of the body segments, which in turn altered the direction of the vector of the resultant ground reaction force. This change was characterized by landing with less knee and hip flexion during the horizontal landing phase, which would position the participant's center of mass closer relative to their base of support, necessitating smaller forward translation of the center of mass during the landing phase compared with when nonfatigued. By using this landing strategy when fatigued, the participants were less efficient in using the SSC and unable to generate an adequate net knee extension moment upon landing. Our previous research has shown that landing with less knee flexion and less hip flexion contributes to a significantly lower net knee extension joint moment (18). The patellar tendon forces were calculated using the net knee extension–flexion joint moment, which does not take into account muscular cocontraction. Therefore, the lower net knee extension joint moment may have resulted from the later peak VL muscle burst activity, which occurred after the time of the F_{PT}. This, in turn, may have resulted in a lower knee extension moment and/or the later peak MG and BF muscle burst activity, which occurred after the time of the F_{PT}, and may have resulted in a lower flexion extension moment. In addition, fatigue also led to the participants performing the stop–jump task with a slower approach speed, which may have also contributed to the lower net knee joint extension moment.

Consistent with previous landing studies, the use of a muscle activation strategy in preparation to land was evident during both landing phases (14–16), and differences existed in these muscle recruitment patterns between the two landing phases (15,18). During the horizontal landing phase of the stop–jump task, although the participants maintained a similar lower limb muscle recruitment order and time of onset, substantial between-fatigue condition differences were observed, including a later peak BF and VL activity relative to the time of the F_{PT}. We speculate that although this later muscle recruitment when fatigued had a positive effect on reducing patellar tendon loading, it could also have a negative effect on dissipating the impact load and knee joint stability due to these muscles acting too late. The significantly later peak MG activity relative to the time of the peak F_{PT} displayed when fatigued during the horizontal landing phase cannot be attributed to any ankle joint kinematics due to no significant between-fatigue condition differences. Nevertheless, the later BF and MG activity, which acts as eccentrically to control the rate of knee extension before IC, relative to the time of the peak F_{PT} when fatigued compared with when nonfatigued, may have contributed to the decreased knee flexion at IC. The muscle activation strategy that may be attributed to the decrease in hip flexion displayed during through the landing phase remains unknown. That is, as BF is a two-joint muscle that can also act eccentrically to control the rate of hip flexion during landing, the delayed BF muscle activity relative to the time of the peak F_{PT} could potentially act to increase hip flexion. However, when fatigued participants displayed decreased hip flexion compared with when nonfatigued. It remains unknown whether this decreased hip flexion displayed throughout landing when fatigued compared with nonfatigued condition was due to an earlier gluteus maximum or later tensor fasciae latae muscle activation strategy, which functions to control hip joint motion, as the activity of these muscles was not measured.

During the vertical landing phase of the stop–jump task, despite fatigue causing only minor changes to the kinetics and kinematics displayed by the participants, fatigue caused substantial alterations to the participants' lower limb muscle activation strategies. Interestingly, muscle activity onset relative to the time of F_{PT} was delayed for TA and the hamstring muscle group when fatigued but remained unchanged for the antigravity muscles, MG, and the quadriceps muscle group. This suggests that coactivation of the hamstrings/quadriceps and MG-TA muscles was delayed when fatigued, potentially leading to a moderate decrease in F_{PT} and active stabilization of the knee and ankle, respectively, during the vertical landing phase of a stop–jump task. Furthermore, as the hamstring muscles act to assist in stabilizing the knee, to decelerate anterior tibial translation, and to internally/externally rotate the hip, the later onset of BF and ST, and later peak BF activity relative to the time of F_{PT} when fatigued during the vertical landing phase may have contributed to the increased F_{AP} compared with when nonfatigued. Nevertheless, it remains unknown whether the faster knee abduction velocity throughout landing when fatigued compared with nonfatigued condition was due to a delayed hip adductor muscle activation strategy, which functions to control lower limb alignment, as this was not measured. We also speculate that the later MG peak muscle activity relative to the time of F_{PT} contributed to the less plantarflexion and a higher plantarflexion velocity at IC due to delayed MG eccentric control of ankle dorsiflexion during landing. It should be noted, however, that the differences in muscle recruitment order displayed during the two landing phases are likely to have been moderated by the preparation for the takeoff phase in the horizontal landing phase, whereas there was no follow-on movement after the vertical landing phase.

CONCLUSIONS

Fatigue was found to significantly decrease the F_{PT} and the LR F_{PT} during the horizontal landing phase but not vertical landing phase of a stop–jump task in asymptomatic athletes with normal patellar tendons. This reduction in patellar tendon loading when landing horizontally while fatigued was primarily attributed to the way the participants moderated their landing technique, including reducing their net knee joint extension moment as a consequence of using less knee flexion and at an angle that used a larger patellar tendon moment arm, less hip flexion, and less efficient use of the SSC during the horizontal landing phase. This decrease in patellar tendon loading when fatigued, caused by altered landing technique during the horizontal landing phase, may be an inherent protective strategy to potentially decrease patellar tendon loading during repetitive landing.

The authors acknowledge the New South Wales Sporting Injuries Committee (Australia) for financial support of this research.

Professor Cook was supported by the Australian Centre for Research into Sports Injury and its Prevention, which is one of the International Research Centres for Prevention of Injury and Protection of Athlete Health supported by the International Olympic Committee (IOC).

There is no conflict of interest for any of the authors.

The results of the present study do not constitute endorsement by the American College of Sports Medicine.

REFERENCES

1. Australian Bureau of Statistics. *Sports and physical recreation: a statistical overview, Australia*, Oct 2008. [cited April 23, 2013] from AusStats Database.
2. Basmajian JV, Gopal DN, Ghista DN. Electrodiagnostic model for motor unit action potential (MUAP) generation. *Am J Phys Med*. 1985;64(6):279–94.
3. Benjaminse A, Habu A, Sell TC, et al. Fatigue alters lower extremity kinematics during a single-leg stop-jump task. *Knee Surg Sports Traumatol Arthrosc*. 2008;34(1):43–54.
4. Bennell K, Talbot R, Wajswelner H, Techovanich W, Kelly D. Intra-rater and inter-rater reliability of a weight-bearing lunge measure of ankle dorsiflexion. *Aust J Physiother*. 1998;44(3): 175–80.
5. Besier TF, Lloyd DG, Ackland TR, Cochrane JL. Anticipatory effects on knee joint loading during running and cutting maneuvers. *Med Sci Sports Exerc*. 2001;33(7):1176–81.
6. Bisseling RW, Hof AL. Handling of impact forces in inverse dynamics. *J Biomech*. 2006;39(13):2438–44.
7. Bisseling RW, Hof AL, Bredeweg SW, Zwerver J, Mulder T. Relationship between landing strategy and patellar tendinopathy in volleyball. *Br J Sports Med*. 2007;41(7):e8.
8. Bisseling RW, Hof AL, Bredeweg SW, Zwerver J, Mulder T. Are the take-off and landing phase dynamics of the volleyball spike jump related to patellar tendinopathy? *Br J Sports Med*. 2008; 42(6):483–9.
9. Chappell JD, Herman DC, Knight BS, Kirkendall DT, Garrett WE, Yu B. Effect of fatigue on knee kinetics and kinematics in stop-jump tasks. *Am J Sports Med*. 2005;33(7):1022–9.
10. Chappell JD, Yu B, Kirkendall DT, Garrett WE. A comparison of knee kinetics between male and female recreational athletes in stop-jump tasks. *Am J Sports Med*. 2002;30(2):261–7.
11. Cohen J. *Statistical Power Analysis for the Behavioral Sciences*. Hillsdale (NJ): Lawrence Erlbaum Associates; 1988, p. 20–7.
12. Cook JL, Khan KM, Kiss ZS, Coleman BD, Griffiths L. Asymptomatic hypoechoic regions on patellar tendon ultrasound: a 4-yr clinical and ultrasound followup of 46 tendons. *Scand J Med Sci Sports*. 2001;11(6):321–7.
13. Cook JL, Khan KM, Kiss ZS, Griffiths L. Patellar tendinopathy in junior basketball players: a controlled clinical and ultrasonographic study of 268 patellar tendons in players aged 14–18 years. *Scand J Med Sci Sports*. 2000;10(4):216–20.
14. Cowling E, Steele JR. Is lower limb muscle synchrony during landing affected by gender? Implications for variations in ACL injury rates. *J Electromyogr Kinesiol*. 2001;11(4):263–8.
15. Edwards S, Steele JR, Cook JL, Purdam CR, McGhee DM. Lower limb movement symmetry cannot be assumed when investigating the stop-jump landing. *Med Sci Sports Exerc*. 2012;44(6):1123–30.
16. Edwards S, Steele JR, McGhee DE. Does a drop landing represent a whole skill landing and is this moderated by fatigue? *Scand J Med Sci Sports*. 2010;20(3):516–23.
17. Edwards S, Steele JR, McGhee DE, Beattie S, Purdam C, Cook JL. Landing strategies of athletes with an asymptomatic patellar tendon abnormality. *Med Sci Sports Exerc*. 2010;42(11):2072–80.
18. Edwards S, Steele JR, McGhee DE, Cook JL, Purdam C, Munro BJ. Characterising patellar tendon loads during dynamic landings. *Scand J Med Sci Sports*. 2012;22(1):2–11.

19. Enoka RM. Mechanisms of muscle fatigue: central factors and task dependency. *J Electromyogr Kinesiol*. 1995;5(3):141–9.
20. Falvo MJ, Bloomer RJ. Review of exercise-induced muscle injury: relevance for athletic populations. *Res Sports Med*. 2006;14(1):65–82.
21. Finni T, Komi PV, Lepola V. In vivo human triceps surae and quadriceps femoris muscle function in a squat jump and counter movement jump. *Eur J Appl Physiol*. 2000;83(4–5):416–26.
22. Ford KR, Myer GD, Hewett TE. Valgus knee motion during landing in high school female and male basketball players. *Med Sci Sports Exerc*. 2003;35(10):1745–50.
23. Ford KR, Myer GD, Toms HE, Hewett TE. Gender differences in the kinematics of unanticipated cutting in young athletes. *Med Sci Sports Exerc*. 2005;37(1):124–9.
24. Grau S, Maiwald C, Krauss I, Axmann D, Janssen P, Horstmann T. What are causes and treatment strategies for patellar-tendinopathy in female runners? *J Biomech*. 2008;41(9):2042–6.
25. Hawkins RD, Hulse MA, Wilkinson C, Hodson A, Gibson M. The association football medical research programme: an audit of injuries in professional football. *Br J Sports Med*. 2001;35(1):43–7.
26. Herzog W, Read LJ. Lines of action and moment arms of the major force-carrying structures crossing the human knee joint. *J Anat*. 1993;182(2):213–30.
27. Hopkins WG. A spreadsheet for deriving a confidence interval, mechanistic inference and clinical inference from a *P* value. *Sportscience*. 2007;11:16–20.
28. Hopkins WG, Marshall SW, Batterham AM, Hanin J. Progressive statistics for studies in sports medicine and exercise science. *Med Sci Sports Exerc*. 2009;41(1):3–13.
29. Horita T, Komi PV, Nicol C, Kyröläinen H. Stretch shortening cycle fatigue: interactions among joint stiffness, reflex, and muscle mechanical performance in the drop jump. *Eur J Appl Physiol*. 1996;73(5):393–403.
30. Horita T, Komi V, Hamalainen I, Avela J. Exhausting stretch-shortening cycle (SSC) exercise causes greater impairment in SSC performance than in pure concentric performance. *Eur J Appl Physiol*. 2003;88(6):527–34.
31. Krustrup P, Mohr M, Steensberg A, Bencke J, Kjaer M, Bangsbo J. Muscle and blood metabolites during a soccer game: implications for sprint performance. *Med Sci Sports Exerc*. 2006;38(6):1165–74.
32. Lian O, Engebretsen L, Ovrebø RV, Bahr R. Characteristics of the leg extensors in male volleyball players with jumper's knee. *Am J Sports Med*. 1996;24(3):380–5.
33. Lian OB, Engebretsen L, Bahr R. Prevalence of jumper's knee among elite athletes from different sports: a cross-sectional study. *Am J Sports Med*. 2005;33(4):561–7.
34. Mann KJ, Edwards S, Drinkwater EJ, Bird SP. A lower limb assessment tool for athletes at risk of developing patellar tendinopathy. *Med Sci Sports Exerc*. 2013;45(3):527–33.
35. McInnes SE, Carlson JS, Jones CJ, McKenna MJ. The physiological load imposed on basketball players during competition. *J Sports Sci*. 1995;13(5):387–97.
36. McLean SG, Felin RE, Suedekum N, Calabrese G, Passerallo A, Joy S. Impact of fatigue on gender-based high-risk landing strategies. *Med Sci Sports Exerc*. 2007;39(3):502–14.
37. McLean SG, Lipfert SW, van den Bogert AJ. Effect of gender and defensive opponent on the biomechanics of sidestep cutting. *Med Sci Sports Exerc*. 2004;36(6):1008–16.
38. Moran KA, Marshall BM. Effect of fatigue on tibial impact accelerations and knee kinematics in drop jumps. *Med Sci Sports Exerc*. 2006;38(10):1836–42.
39. Nicol C, Avela J, Komi PV. The stretch-shortening cycle: a model to study naturally occurring neuromuscular fatigue. *Sports Med*. 2006;36(11):977–99.
40. Nisell R, Ekholm J. Patellar forces during knee extension. *Scand J Rehabil Med*. 1985;17(2):63–74.
41. Ostenberg A, Roos H. Injury risk factors in female European football. A prospective study of 123 players during one season. *Scand J Med Sci Sports*. 2000;10(5):279–85.
42. Perneger TV. What's wrong with Bonferroni adjustments. *Brit Med J*. 1998;316(7139):1236–8.
43. Richards DP, Ajemian SV, Wiley JP, Brunet JA, Zernicke RF. Relation between ankle joint dynamics and patellar tendinopathy in elite volleyball players. *Clin J Sport Med*. 2002;12(5):266–72.
44. Richards DP, Ajemian SV, Wiley JP, Zernicke RF. Knee joint dynamics predict patellar tendinitis in elite volleyball players. *Am J Sports Med*. 1996;24(5):676–83.
45. Sanchis-Alfonso V, Rosello-Sastre E, Subias-Lopez A. Neuroanatomic basis for pain in patellar tendinosis ("jumper's knee"): a neuroimmunohistochemical study. *Am J Knee Surg*. 2001;14(3):174–7.
46. Sanna G, O'Connor KM. Fatigue-related changes in stance leg mechanics during sidestep cutting maneuvers. *Clin Biomech*. 2008;23(7):946–54.
47. Sell TC, Ferris CM, Abt JP, et al. The effect of direction and reaction on the neuromuscular and biomechanical characteristics of the knee during tasks that simulate the noncontact anterior cruciate ligament injury mechanism. *Am J Sports Med*. 2006;34(1):43–54.