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A functional agility short-term fatigue protocol changes lower extremity mechanics

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Abstract

The purpose of this study was to evaluate the effects of a functional agility fatigue protocol on lower extremity biomechanics between two unanticipated tasks (stop-jump and sidestep). The subjects consisted of fifteen female collegiate soccer athletes (19 ± 0.7 years, 1.67 ± 0.1 m, 61.7 ± 8 kg) free of lower extremity injury. Participants performed five trials of stop-jump and sidestep tasks. A functional short-term agility protocol was performed, and immediately following participants repeated the unanticipated running tasks. Lower extremity kinematic and kinetic values were obtained pre and post fatigue. Repeated measures analyses of variance were conducted for each dependent variable with an alpha level set at 0.05. Knee position post-fatigue had increased knee internal rotation (11.4 ± 7.5° vs. 7.9 ± 6.5° $p = 0.011$) than pre-fatigue, and a decreased knee flexion angle (−36.6 ± 6.2° vs.−40.0 ± 6.3°, $p = 0.003$), as well as hip position post-fatigue had decreased hip flexion angle (35.5 ± 8.7° vs. 43.2 ± 9.5°, $p = 0.002$). A quick functional fatigue protocol altered lower extremity mechanics of Division I collegiate soccer athletes during landing tasks. Proper mechanics should be emphasized from the beginning of practice/game to aid in potentially minimizing the effects of fatigue in lower extremity mechanics.

Keywords

anterior cruciate ligament; biomechanics; knee; hip

Introduction

Anterior Cruciate Ligament injury continues to have a higher rate of incidence in the female population (Arendt, Agel, & Dick, 1999; Mountcastle, Posner, Kragh, & Taylor, 2007). Approximately 70% of ACL tears during athletic activities are a consequence of a non-contact mechanism occurring towards latter stages of the game when fatigue (e.g., central, muscular) is theorized to play a role and are most likely influenced by altered biomechanical control (Boden, Dean, Feagin, & Garrett, 2000; Malinzak, Colby, Kirkendall, Yu, & Garrett, 2001). The incidence of ACL tears in athletics is especially relevant in soccer and basketball female players (Arendt et al., 1999). Further, it has been theorized that with the presence of fatigue and unanticipated stimuli the movement patterns are changed into more stressful positions, potentially increasing the risk of injury (Borotikar, Newcomer, Koppes, & McLean, 2008; Chappell, et al., 2005).

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Research focusing on fatigue has utilized distinct protocols, including single leg squats (McLean & Samorezov, 2009), vertical jumps and short sprints (Chappell et al., 2005), leg press repetitions (Gehring, Melnyk, & Gollhofer, 2009), parallel squat exercises (Kernozek, Torry, & Iwasaki, 2008), step ups and plyometric jumps (McLean, et al., 2007), consecutive squats at 1 Hz (Borotikar et al., 2008), and an intermittent shuttle run protocol (Sanna & O'Connor, 2008). Chappell et al. (2005) reported that after a fatigue protocol consisting of five consecutive vertical jumps and 30 m sprint repeated until the participant could no longer continue, there was an increase in proximal tibial anterior shear force, knee valgus moment, and a decrease in knee flexion angle. In contrast, McLean et al. (2007) did not find any differences in sagittal plane angles and/or loads post-fatigue, but rather found differences in frontal plane knee kinematics and kinetics. The disparity in fatigue protocols may be a reason for the dissimilar findings between studies. Fatigue protocols have been focused either on isolated (e.g., hamstring) (Nyland, Caborn, Shapiro, & Johnson, 1997; Nyland, Shapiro, Caborn, Nitz, & Malone, 1997; Wojtys, Wylie, & Huston, 1996), or central fatigue (e.g., general) (Borotikar et al., 2008; Chappell et al., 2005;Madigan & Pidcoe, 2003; McLean et al., 2007; McLean & Samorezov, 2009). Central fatigue protocols have been speculated to decrease the neuromuscular system of the entire body, and specifically the lower extremity, rather than just knee-related structures (e.g., isolated hamstring fatigue) (Cairns, Knicker, Thompson, & Sjogaard, 2005; Miura et al., 2004). Further, the lack of sport specific drills in some protocols may not be applicable to this specific population.

The utilization of a functional agility protocol that can mimic skills commonly used by athletes is of interest for assessing lower extremity biomechanics during fatigued status (McLean et al., 2007). Few experiments have investigated the effects of fatigue by including soccer skills in the protocol; the lack of drills specific to the athletic event that incorporate multiple directions (e.g., cut right, up-and-down, etc.), and deceleration and accelerations is yet to be applied in fatiguing protocols (Chappell et al., 2005; Kernozek et al., 2008; McLean et al., 2007; McLean & Samorezov, 2009). Hence, combining exercises with various muscular recruitment patterns, a deceleration and acceleration phase that can elicit the entire lower extremity musculature that is normally involved during athletic tasks (e.g., sidestep, stop-jump) is of importance to study the effects of fatigue on lower limb mechanics.

The conflicting results in the literature that have presented divergent lower extremity biomechanical patterns may be due to varied methodological approaches regarding the protocols used. Further, the conflicting use of anticipation and unanticipation stimuli may not make the results comparable between studies. Therefore, the purpose of this study was to evaluate the effect of a functional agility short-term fatigue protocol on lower extremity kinematics and kinetics while performing a stop-jump and sidestep cutting tasks. We hypothesize that at post-fatigue the participants will have decreased knee and hip flexion angles, as well as increased internal knee adduction moment. Further, we also hypothesize that the changes post-fatigue for the stop-jump task will primarily occur in the sagittal plane (e.g., flexion), whereas for the sidestep cutting will be observed in the frontal plane (e.g., adduction).

Methods

Participants

A sample of convenience of fifteen NCAA Division I female soccer players (age = 19.2 ± 0.8 years; height = 1.67 ± 0.05 m; mass = 61.7 ± 8.1 kg) were deemed adequate to participate in this study, based on data from previous studies focusing on the effects of fatigue on lower extremity biomechanics and to achieve 80% statistical power, effect size of

0.7, and with an alpha level of 0.05 (Borotikar et al., 2008; Chappell et al., 2005; McLean & Samorezov, 2009). Institutional Review Board approval and written informed consent for all participants was obtained prior to testing. Participants were free of lower extremity injury, and were excluded if (i) presented a history of lower extremity surgery within the past two years, (ii) sustained an injury to the lower extremity within the past six months, and (iii) currently pregnant. The dominant leg was used for analysis and was defined as the leg that the participant would use to kick a soccer ball as far as possible. Participants wore spandex shorts, tight fitting clothing, and the team running shoes provided at the beginning of the season (Adidas Supernova, AG, Herzogenaurach, Germany).

Testing procedure

Participants had lower extremity three-dimensional joint kinematic and ground reaction force data recorded throughout the execution of unanticipated tasks, prior and immediately after being exposed to a protocol designed to induce fatigue. Prior to testing, participants were allotted time to warm-up, consisting of cycling and self-directed stretching. After the warm-up, maximum vertical jump was obtained using a VERTEC device (Chappell et al., 2005).

Participants had to perform five unanticipated trials of each task, total of ten trials, within the field of view of eight high-speed motion analysis capture cameras (Vicon, Oxford, England), and contact force plates while executing the tasks (Model: 4060–10, Bertec, Columbus OH, USA). To create the unanticipated factor, a light beam was passed across the area where the participants were running and 2 metres prior to the force plates. When the light beam was interrupted it triggered a software program to randomly generate the athletic task (sidestep or stop-jump) and project it onto a screen in front of the participants. The inability to anticipate and the environment allowed mimicking as much as possible a soccer athletic event (Cortes, Blount, Ringleb, & Onate, 2011). A Brower timing system (Brower Timing Systems, Draper UT, USA) was used to control the approach speed.

The sidestep cutting task consisted of a running approach, step with the dominant foot on the force plate, and cut to the contra-lateral side of the dominant foot touching the force plate at an angle of approximately 45° (Cortes, Onate, & Van Lunen, 2011). The stop-jump task consisted of a running approach and planting onto the force plates with one foot on each force plate and jumping straight into the air as if performing a soccer header. Prior to data collection, the participants had three practice trials or more until they felt comfortable with the task. There was a 1-min rest period between trials to minimize fatigue during pre-fatigue assessment, whereas no rest was given during post-fatigue trials. Testing trials were repeated if the participant did not completely land on the force plate, or were unable to execute the trials at a minimum speed of 3.5 m/s. A single experienced experimenter determined if the trials were deemed successful or unsuccessful. After completion of the 10 successful unanticipated trials, five of each task, participants received instructions about the fatigue protocol.

Functional agility short-term fatigue protocol

The protocol consisted of a series of agility exercises, which included: step-ups on a 30-cm box, ‘L-drill’, five counter-movement jumps staying within 18–22% of their maximum vertical jump previously obtained, and running back and forth on an agility ladder. Participants started in a three-point stance in front of three cones that were set up in an L shape, with each cone 4.5 yards apart. They sprinted 4.5 yards to one cone, sprinted back to the starting cone, and then headed back to the second cone where they ran around it and cut right to the third cone. Participants then ran in a circle around the third cone from the inside to the outside and ran around the second cone before running to the first cone (Sierer, Battaglini, Mihalik, Shields, & Tomasini, 2008). The protocol started by having the participants perform a series of step-up movements onto a 30-cm height box for 20 seconds with the metronome set at 200 bpm (McLean et al., 2007). Immediately after, they performed one repetition of the ‘L-drill’ they performed 5 consecutive counter-movement jumps. Following the counter- movement jumps, they performed the agility ladder drill. For the first and third set of the protocol, participants ran in the forward direction with both feet touching inside each space of the ladder, whereas in the second and fourth set of the protocol, participants went sideways with both feet touching inside the space of the ladder to a metronome set at 220 bpm. Each set included the four agility drills. They had to perform a total of four sets of the fatiguing protocol with no rest in between. Four sets were chosen based on a pilot study at our laboratory that has shown to obtain the criteria for fatigue. Subsequent to the four sets, participants had to perform the stop-jump and sidestep tasks again for post-fatigue assessment. To be considered in a fatigued condition, subjects’ heart rate had to be at a minimum of 85% of their estimated maximum heart rate.

Biomechanical analysis

Forty reflective markers were placed on specific body landmarks, with 10 being calibration markers. A standing trial and a dynamic trial (circular motion of the pelvis with lower body static) to calculate hip joint center were obtained prior to data collection. For these trials, the participants were standing in a neutral position with arms across the chest, and foot placement standardization was accomplished by drawing foot outlines on the force plate and having the participants standing on top of it. The calibration markers were removed prior to the start of protocol. Pelvic tracking markers were secured with surgical glue, pre-wrap and power flex tape. Thigh and shank cluster markers were also secured with power flex tape. Special plates were made to create a cluster for the foot that would attach to the participants shoes, and were secured with Cramer athletic tape.

Eight high-speed cameras sampling at 300 Hz were used to track marker trajectory, and two force plates sampling at 1200 Hz measured ground reaction forces. From the standing trial a kinematic model (pelvis, thigh, shank, and foot) was created for each participant using Visual 3D software (C-Motion, Inc, Germantown MD, USA) using a least-squares optimization (Lu & O’Connor, 1999). This kinematic model was used to quantify the motion at the hip, knee, and ankle joints, with the rotations being expressed relative to the standing trial. To calculate functional hip joint centers the standing trial with circular motion of the pelvis was used (Begon, Monnet, & Lacouture, 2007; Schwartz & Rozumalski, 2005). A fourth-order Butterworth zero lag filter with 7 and 25 Hz cutoff frequency was used to filter trajectory data and ground reaction force data, respectively. A standard inverse dynamics analysis, using segment inertial characteristics estimated for each participant as per the methods of Dempster (Dempster, 1955), was applied to the kinematic and ground force data to calculate 3D forces and moments (Winter, 2005). Intersegmental joint moments are defined as internal moments and were describe relative to the respective joint-coordinate system (e.g., a knee internal extension moment will resist a flexion load applied to the knee). All data was normalized to 100% of stance, with initial contact being when vertical ground reaction force exceed 10 *N*, and ending with toe-off.

Statistical analysis

Independent variables included tasks (sidestep and stop-jump) and fatigue conditions (pre- and post-fatigue). A 2 (tasks) × 2 (fatigue conditions) factor repeated measures analysis of variance (ANOVA) was conducted for each dependent variable. Tukey post-hoc analyses were conducted where appropriate. The dependent variables included: vertical and posterior ground reaction forces, knee flexion, knee abduction, knee rotation, knee flexion-extension moment, knee abduction-adduction moment, hip flexion, hip abduction, and hip flexion moment. These variables were measured at different points in time that included: initial contact, peak vertical and posterior ground reaction forces, peak knee flexion, and peak stance (McLean et al., 2007). Data were analysed between initial contact and maximum knee flexion, which defines the stop-jump phase. All data were reduced using Visual 3D and a custom made MATLAB (The MathWorks, Inc, Natick MA, USA) program. Each of the five trials were averaged and exported into SPSS (SPSS Inc, Chicago IL, USA) for data analysis. Alpha level was set *a priori* at 0.05.

Results

Heart rate profile during the fatigue protocol is represented in Figure 1. Descriptive statistics with mean and standard deviations are presented in Table I and Table II. Participants completed 6.2 ± 1.7 trials of the sidestep cutting task, and 5.7 ± 1.4 trials of the stop-jump task. No statistical significant interaction was found between fatigue and tasks ($p > 0.05$). The results will be presented based on main effects for fatigue assessment, and tasks.

Pre- to post-fatigue assessment

The primary differences were observed in the sagittal plane between pre and post-fatigue. During post-fatigue assessment, the following dependent variables had significantly decreased when compared with the pre-fatigue status: knee rotation at initial contact (Pre: 7.9 ± 7.0, Post: 11.4 ± 7.7, $F_{1,14} = 8.696$, $p = 0.011$) (Figure 2), knee flexion at peak posterior ground reaction force (Pre: −40.0 ± 6.3, Post: −36.6 ± 6.2, $F_{1,14} = 12.601$, $p = 0.003$), peak knee flexion (Pre: −55.3 ± 8.0, Post: −49.9 ± 8.7, $F_{1,14} = 29.675$, $p < 0.001$), hip flexion at initial contact and peak hip flexion (Pre: 43.2 ± 8.2, Post: 35.5 ± 9.5, $F_{1,14} = 14.341$, $p = 0.002$, and Pre: 45.8 ± 9.4, Post: 37.5 ± 9.8, $F_{1,14} = 16.340$, $p = 0.001$, respectively) (Figure 3).

Tasks

The stop-jump task had significantly higher sagittal angles than the sidestep cutting. Hip flexion at initial contact and its peak was significantly higher for the running-stop than the sidestep cutting (Pre: 32.3 ± 9.5, Post: 46.4 ± 13.2, $F_{1,14} = 209.122$, $p < 0.001$, and Pre: 33.8 ± 13.7, 49.4 ± 14.9, $F_{1,14} = 140.221$, $p < 0.001$, respectively). Lastly, a significant difference was attained for peak posterior ground reaction force ($F_{1,14} = 89.378$, $p < 0.001$), with the stop-jump having higher posterior ground reaction force (1.2 ± 0.2 BW) than the sidestep (0.7 ± 0.4 BW), $p < 0.001$.

Discussion

One of the main results to emerge from this study was the altered landing pattern under a fatigued status (e.g., knee rotation, knee flexion) regardless of the task. Furthermore, hip flexion and posterior ground reaction force were significantly changed between tasks and fatigue conditions. These altered knee and hip positions have been theorized to increase the likelihood of injury. We found that during post-fatigue, the participants were in a more extended position at the knee and hip joint. Landing in an extended position has been associated with increased proximal tibia anterior shear force due to increased patellar tendon-tibia shaft angle (Hughes & Watkins, 2006). This increased force has been theorized to increase the load placed on the ACL during landing activities (Chappell et al., 2005; Kernozek et al., 2008). We speculate that our participants had increased load on the ACL that could have increased the load due to their erect posture at post-fatigue. This is most likely due to the mechanical disadvantage suffered by the hamstring muscles that cannot co-contract strongly enough at smaller angles to produce a large posterior force (Pandy & Shelburne, 1997). Previous researchers have reported altered movement patterns during

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post-fatigue; specifically increased knee abduction and internal rotation angles (Borotikar et al., 2008; McLean et al., 2007; McLean & Samorezov, 2009; McNair, Hewson, Dombroski, & Stanley, 2002), increased hip rotation angles (Baca, 1999; Borotikar, et al., 2008; McLean & Samorezov, 2009), hip internal rotation moments (McLean & Samorezov, 2009), and decreased knee flexion angles (Chappell et al., 2005).

Decreased knee flexion angles at peak posterior ground reaction force and peak stance were attained; during pre-fatigue the participants had 40 degrees, whereas at post-fatigue they achieved 36.5 degrees of knee flexion. The reduced knee flexion angle at peak posterior ground reaction force has been theorized to be associated with increased anterior shear force (Yu, Lin, & Garrett, 2006). This anterior shear force is thought to increase the load and strain in the anterior cruciate ligament and thus the probability of rupture. Our participants showed a decline in knee flexion to 36.5 degrees after fatigue, but surpassed the threshold of 30 degrees. The change from pre to post-fatigue was smaller than in previous studies (Chappell et al., 2005). A methodological difference may explain the diminished decrease. Chappell et al. (2005) utilized recreational athletes, while we utilized highly trained Division I soccer players. The level of training and practice with the presence of fatigue, and overall skills in performing such tasks, may have decreased the effects of fatigue on stopping tasks. Another significant finding was the decrease in hip flexion at post-fatigue. The difference in hip flexion angle may have contributed to the different biomechanical demands of each task analysed. The sidestep requires an immediate change of direction with minimal deceleration, whereas the stop-jump requires complete deceleration in order to perform a vertical jump. Rather than focusing on the biomechanical differences between tasks, the focus should be the fact that regardless of the task, fatigue induced decreases in hip flexion during both cases. This may have been due to the fact that fatigue decreases the ability to produce muscular activity as in a non-fatigued status, as well as decreased neuromuscular control.

Fatigue can be defined as decrease in performance (Stamford, Weltman, Moffatt, & Fulco, 1978); with muscle fatigue being associated with the accumulation of lactic acid in muscles (Jacobs, 1981). Our participants, during a 6-min fatigue protocol, surpassed their anaerobic threshold since they were at or above 86% of their estimated maximum heart rate (Koutedakis & Sharp, 1986). This suggests that most likely the accumulation of lactic acid in their muscles could have been one of the reasons for the detriment in performance. Associating the decrease in the muscles' ability to perform with the decision process to successfully complete the task presented by the simulation software (e.g., unanticipated), it may be plausible to assume that the combination of the central (decision process) and peripheral fatigue (muscles) may have contributed to the altered biomechanical patterns seen post-fatigue. McLean and associates have proposed that the central process is diminished with such fatigue models (McLean & Samorezov, 2009). Further, it has been shown that soccer players usually attain 87% of their maximum heart rate during a competitive soccer match (Krustrup, Mohr, Ellingsgaard, & Bangsbo, 2005). With our fatigue protocol, we were able to mimic similar heart rate percentages potentially achieved during a soccer match (Krustrup et al., 2005), and induce fatigue to alter landing mechanics (i.e., erect posture) (Chappell et al., 2005; McLean et al., 2007; McLean & Samorezov, 2009) as typically occurs towards the later stages of a game (Boden et al., 2000).

It is noteworthy that our functional agility short-term fatigue protocol took approximately 6 min. This short-time frame to induce fatigue had the same effects than other long lasting protocols (e.g., decreased knee and hip flexion, and increased knee internal rotation) (Greig & Siegler, 2009). This raises two points: (i) the neuromuscular adaptations are dependent on the intensity of the exercise and duration of the activity, thus athletes may be injured earlier in the game/practice if played at a high level of intensity during short period, and (ii) training programs should focus on developing strategies to accommodate the biomechanical

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changes during fatiguing activities. An alternative explanation, perhaps our fatigue protocol, Greig & Siegler (2009) fatigue protocol, and other protocols are not soliciting the demands that are observed during an actual game (Borotikar et al., 2008; Chappell et al., 2005; McLean & Samorezov, 2009). Potentially, in order to develop a fatigue protocol that mimics the conditions of an actual game, it would be necessary to quantify the fatigue that athletes are experiencing when ACL tears occur. To our knowledge, no intervention program has been specific in providing instructions when athletes were fatigued or towards the end of the practice. These intervention programs have primarily focused on warm-up and early stages of practice (Gilchrist et al., 2008;Mandelbaum et al., 2005; Steffen, Bakka, Myklebust, & Bahr, 2008; Steffen, Myklebust, Olsen, Holme, & Bahr, 2008). Furthermore, screening tools such as the Landing Error Scoring System (LESS) can potentially be implemented after such a short-term fatigue protocol to attempt to identify those athletes at high and low risk of ACL injury during a non-fatigue and a fatigue status (Onate, Cortes, Welch, & Van Lunen, 2010).

Conclusion

It was shown that regardless of the task analysed, lower extremity changes were noted at post-fatigue. It is also important to note the short time frame needed to alter biomechanics of the lower extremity. Given that changes in knee and hip position occurred within a 6-min time frame, proper mechanics from the beginning of practice/game are necessary to aid in keeping the lower extremity in a relatively safe position. A quick functional fatigue protocol negatively altered the landing mechanics of Division I collegiate soccer athletes. This information may be useful in developing future ACL prevention programs. As mentioned previously, clinicians may use this fatigue protocol, combined with clinical tools (i.e., Landing Error Scoring System − LESS), as a quick assessment tool in order to identify individuals at high(er) risk for ACL injury. Instructing on proper mechanics to those individuals at higher risk of injury is a definitive factor in working towards ACL prevention.

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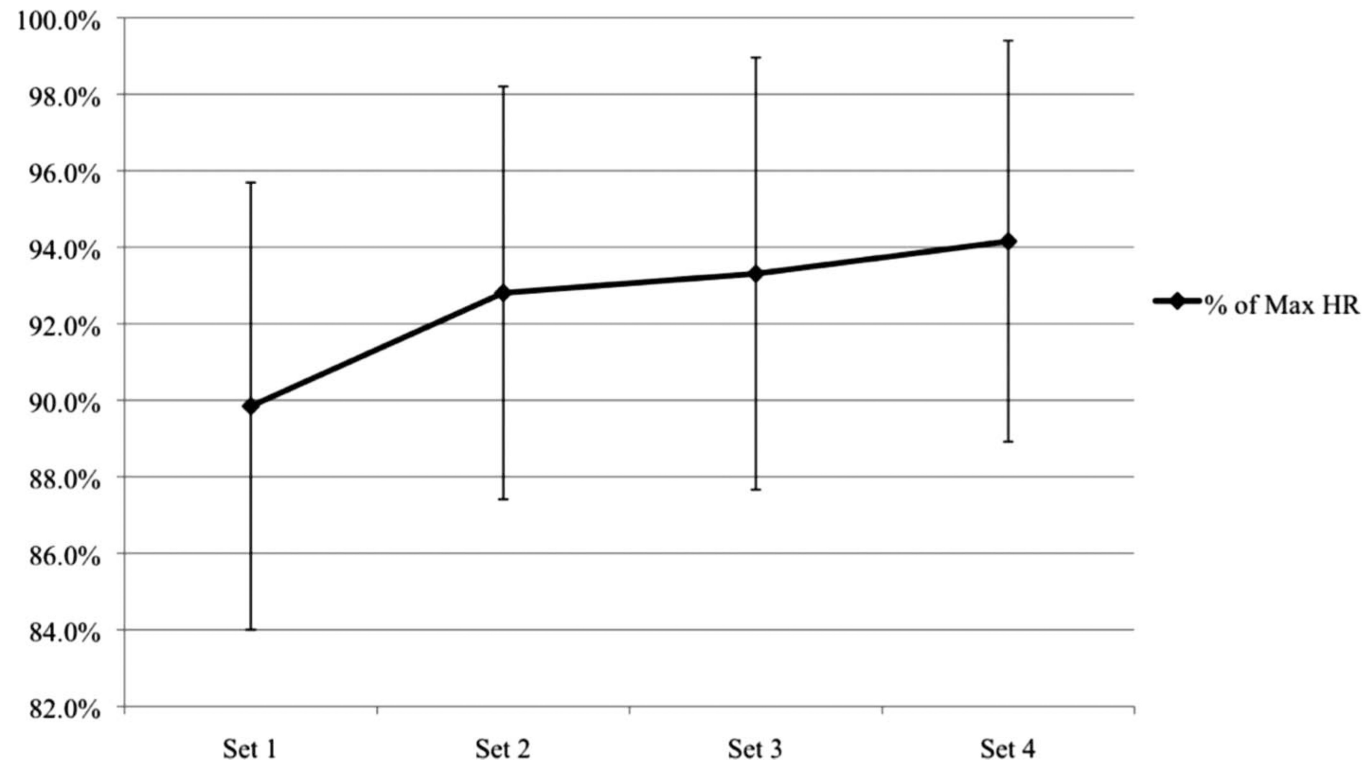

Figure 1.
Percentage of Maximum Heart Rate for each set throughout the functional fatigue protocol. Mean and standard deviation presented.

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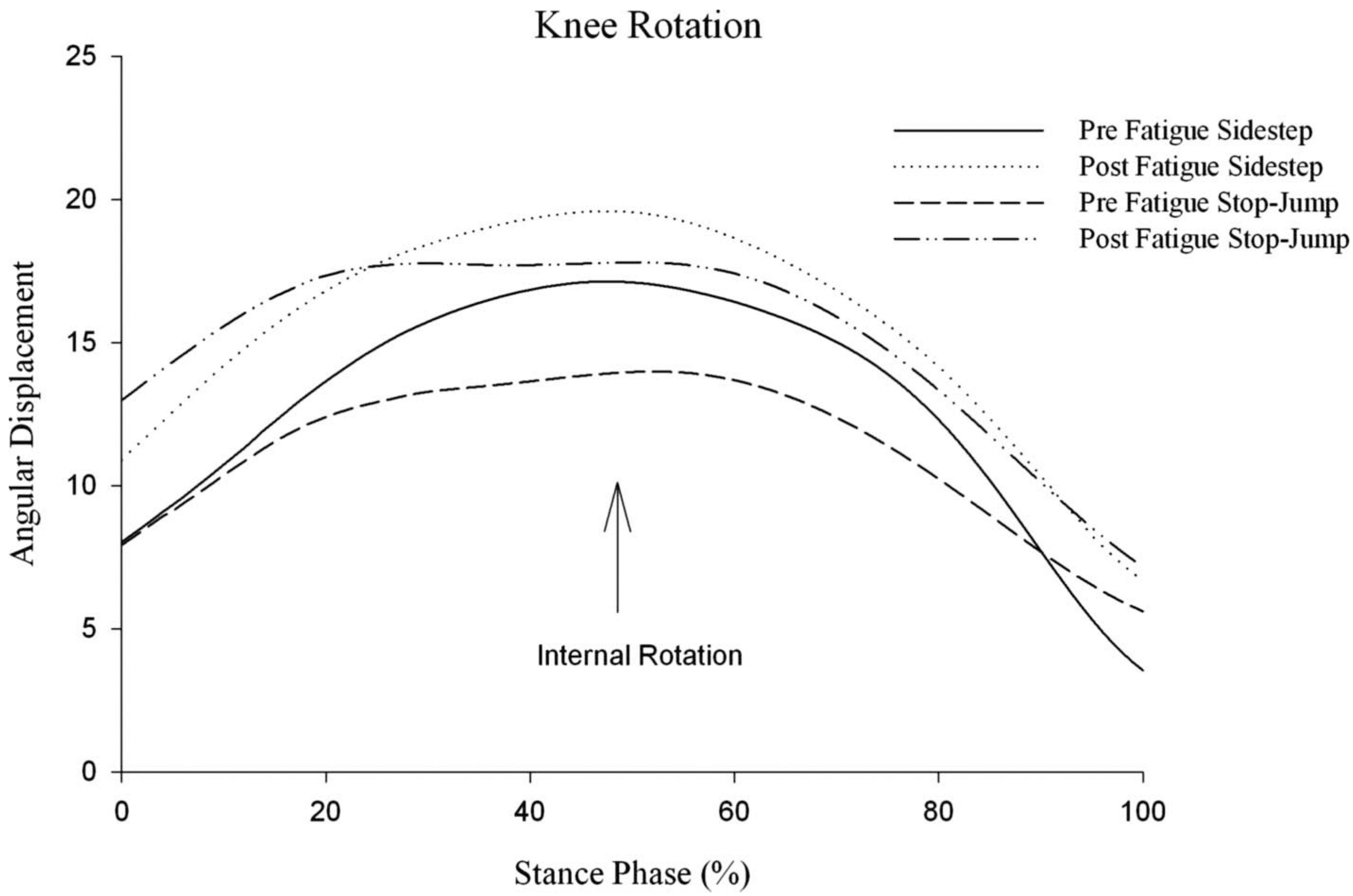

Figure 2.

Knee rotation angular displacement throughout the stance phase during pre and post fatigue conditions for the sidestep cutting and stop-jump task. Lines represent normative values of the entire sample.

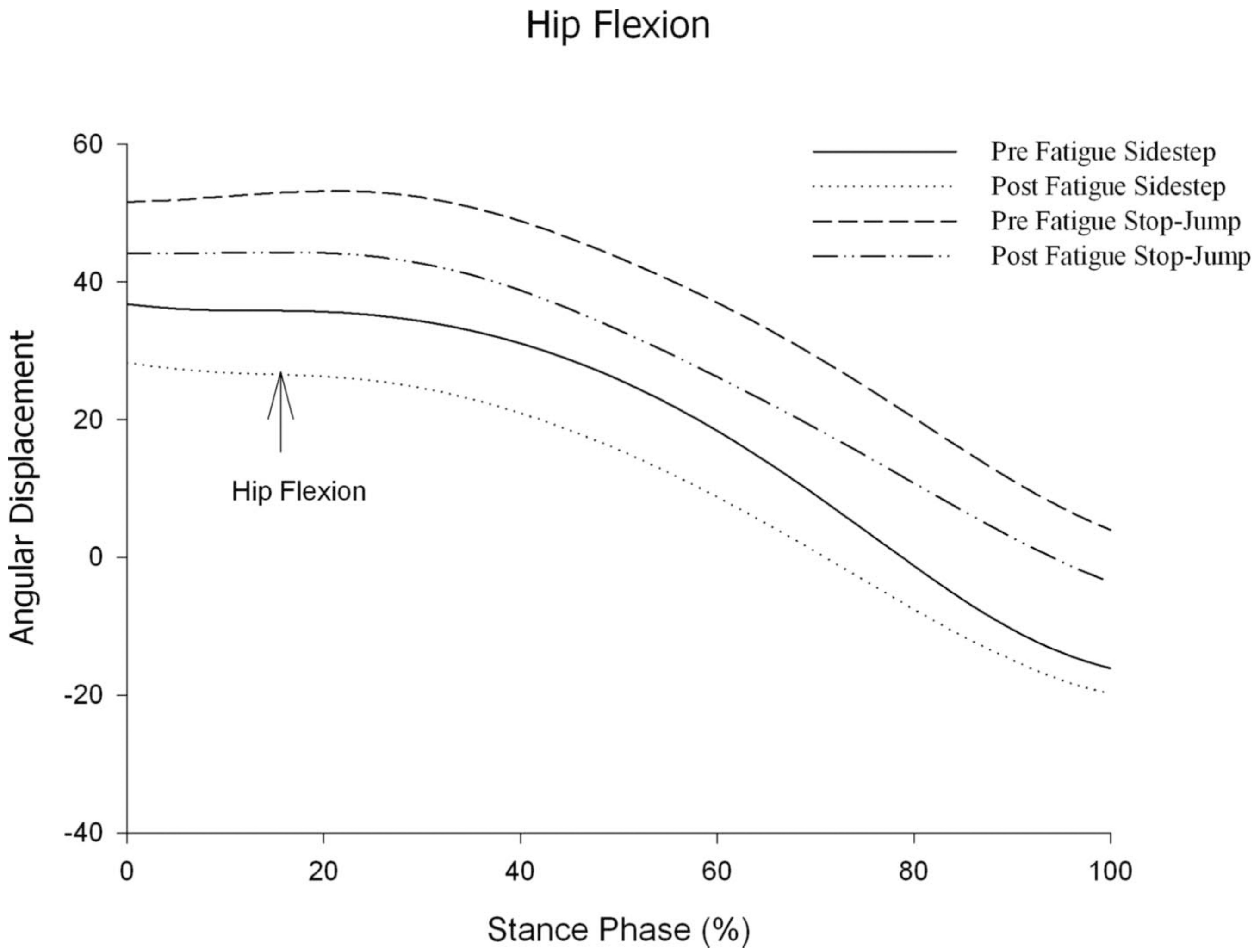

Figure 3.

Hip flexion angular displacement throughout the stance phase during pre and post fatigue conditions for the sidestep cutting and stop-jump task. Lines represent normative values of the entire sample.

Table I

Descriptive statistics (mean and standard deviation) for kinematic variables between fatigue states and tasks at initial contact, peak vertical ground reaction force (PVGRF), peak posterior ground reaction force (PPGRF), and peak stance. All variables measured in degrees.

	Sidestep Cutting				Stop-Jump			
	Pre test		Post Test		Pre test		Post Test	
	Mean	SD	Mean	SD	Mean	SD	Mean	SD
Initial Contact								
Knee Flexion (−)/Extension (+)	−25.5	8.0	−22.3	7.1	−25.9	8.1	−25.2	9.1
Knee Abduction (−)/Adduction (+)	−2.0	2.5	−2.4	3.7	−1.3	4.2	−2.3	5.6
Knee Internal (+)/External Rotation (−)*	14.2	6.2	10.9	5.6	7.9	7.8	11.9	9.8
Hip Flexion*#	36.5	8.0	28.1	9.3	49.9	8.4	42.9	9.6
Hip Abduction (−)/Adduction (+)	−10.2	5.7	−12.6	5.8	−5.5	4.5	−6.4	5.8
PVGRF								
Knee Flexion (−)/Extension (+)	−41.6	7.7	−37.4	6.8	−38.4	5.0	−35.7	5.6
PPGRF								
Knee Flexion (−)/Extension (+)*	−34.2	8.6	−31.2	7.7	−38.5	5.5	−34.9	6.7
Peak Stance								
Knee Flexion (−)/Extension (+)*	−53.1	7.0	−48.3	7.4	−57.6	8.9	−51.6	10.1
Hip Flexion*#	38.3	8.8	29.3	9.7	53.3	10.0	45.6	9.8
Knee Abduction (−)/Adduction (+)	−7.11	4.8	−7.2	6.4	−6.1	4.9	−7.0	5.9

*Statistically significant difference between pre and post fatigue ($P < 0.05$).

#Statistically significant difference between sidestep cutting and stop-jump ($P < 0.05$).

Table II

Descriptive statistics (mean and standard deviation) for kinetic variables between two tasks at initial contact, peak vertical ground reaction force (PVGRF), peak posterior ground reaction force (PPGRF), and peak stance. Ground reaction forces measures in multiples of body weight; internal moments measures in Nm · Kgm.

	Sidestep Cutting				Running Stop			
	Pre test		Post Test		Pre test		Post Test	
	Mean	SD	Mean	SD	Mean	SD	Mean	SD
Initial Contact								
Knee Abduction (−)/Adduction (+) Moment	0.070	0.090	0.070	0.080	0.05	0.20	0.071	0.23
Knee Flexion (−)/Extension (+) Moment	0.001	0.170	0.01	0.19	0.04	0.18	0.06	0.23
Posterior Ground Reaction Force	0.2	0.1	0.2	0.1	0.19	0.14	0.21	0.15
PVGRF								
Vertical Ground Reaction Force	2.7	0.5	2.8	0.4	2.9	0.64	3.0	0.8
PPGRF								
Posterior Ground Reaction Force#	0.7	0.2	0.7	0.2	1.2	0.4	1.3	0.3
Peak Stance								
Knee Flexion (−)/Extension (+) Moment	1.920	0.280	1.880	0.290	1.98	0.33	1.92	0.39
Knee Abduction (−)/Adduction (+) Moment	0.430	0.360	0.370	0.270	0.33	0.13	0.41	0.22

Statistically significant difference between sidestep cutting and stop-jump ($P < 0.05$).