Biomechanical changes at the knee after lower limb fatigue in healthy young women

Heather S. Longpré a, Jim R. Potvin b, Monica R. Maly c,⁎

a School of Rehabilitation Science, Institute for Applied Health Sciences, Rm 403, McMaster University, 1400 Main Street West, Hamilton, ON L8S 1C7, Canada
b Department of Kinesiology, Ivor Wynne Centre, Rm 219A, McMaster University, 1400 Main Street West, Hamilton, ON L8S 1C7, Canada
c School of Rehabilitation Science, Institute for Applied Health Sciences, Rm 435/403, McMaster University, 1400 Main Street West, Hamilton, ON L8S 1C7, Canada

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ABSTRACT

Background: The purpose of this study was to identify changes in knee kinematics, kinetics and stiffness that occur during gait due to lower limb neuromuscular fatigue.

Methods: Kinematic, kinetic and electromyographic measures of gait were collected on healthy, young women (n = 20) before and after two bouts of fatigue. After baseline gait analysis, two bouts of fatiguing contractions were completed. Fatigue was induced using sets of 50 isotonic knee extensions and flexions at 50% of the peak torque during a maximum voluntary isometric contraction. Fatigue was defined as a drop in knee extension or flexion maximum voluntary isometric torques of at least 25% from baseline. Gait analyses were completed after each bout of fatigue. Dynamic knee stiffness was calculated as the change in knee flexion moment divided by the change in knee flexion angle from 3 to 15% of the gait cycle. Co-activations of the biceps femoris and rectus femoris muscles were calculated from 3 to 15% and 40 to 52% of gait. Repeat measures analyses of variance assessed differences in discrete gait measures, knee torques, and electromyography amplitudes between baseline and after each bout of fatigue.

Findings: Fatigue decreased peak isometric torque. Fatigue did not alter knee adduction moments, knee extension moments, co-activations, flexion angles, dynamic knee stiffness, or muscle co-activation. Fatigue reduced the peak knee extension moment.

Interpretation: While neuromuscular fatigue of the knee musculature alters the sagittal plane knee moment in healthy, young women during walking, high intensity fatigue is not consistent with known mechanical environments implicated in knee pathologies or injuries.

⁎ Corresponding author.
E-mail addresses: longpre@mcmaster.ca (H.S. Longpré), potvinj@mcmaster.ca (J.R. Potvin), mmalymc@mcmaster.ca (M.R. Maly).

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co-activation of opposing muscle groups will reduce the total net joint moment. This will translate into an apparent decrease in torque generating capacity of the muscles surrounding the joint (Bennell et al., 2008; Busse et al., 2006; Ebenbichler et al., 1998; Heiden et al., 2009). Furthermore, co-activation increases the resistance to angular joint motion (rotational stiffness). This stiffness represents the rotational resistance contributions from the muscles and other soft tissues controlling knee excursion (Dixon et al., 2010; Zeni and Higginson, 2009; Zeni et al., 2009). Higher stiffness, due to greater antagonistic hamstrings activity during the loading portion of gait, occurs in severe knee OA compared to healthy individuals (Zeni and Higginson, 2009). This study aimed to identify the biomechanical changes at the knee that occur during gait in response to neuromuscular fatigue of the knee extensor and flexor muscles in healthy, young women. Our goal was to determine whether the effects of fatigue on knee kinetics and kinematics would further our understanding of the contributions of muscle dysfunction to knee loading patterns. We hypothesized that neuromuscular fatigue around the knee would increase the knee adduction moment, increase co-activation of the quadriceps and hamstrings, and increase dynamic knee stiffness during gait.

2. Methods

2.1. Participants

A convenience sample of 20 healthy women (age 18 to 30 years) completed this study. Inclusion criteria included regular physical activity, and no contraindications to exercise on the Physical Activity Readiness Questionnaire (Thomas et al., 1992). Exclusion criteria included activity, and no contraindications to exercise on the Physical Activity Questionnaire (Thomas et al., 1992). Exclusion criteria in- cluded pregnancy and a history of knee pain, injury, or surgery. Participants were recruited from the local student population. Written, informed consent was obtained from the participants and this study was approved by the McMaster University Faculty of Health Sciences Research Ethics Board.

2.2. Design

The participants came to the laboratory for two visits, one week apart (Fig. 1). The first visit was an orientation to familiarize participants with the equipment and protocol. This was incorporated in the study design to facilitate maximal performance during the second visit. The second visit consisted of gait analyses before and after two separate bouts of fatigue. Baseline (BL) gait analysis was performed to establish points of reference prior to fatigue. This was followed by a measure of baseline isometric peak knee extensor and flexor torques, followed by the first round of fatiguing contractions. Upon reaching fatigue, the participants performed the first post fatigue gait analysis (PF1). This was followed by a second round of fatiguing contractions to ensure that participants remained fatigued and had not recovered. The final post fatigue gait analysis (PF2) was performed quickly after achieving fatigue.

2.3. Measures

2.3.1. Gait analysis

Marker motion during gait was measured using a Vicon MX 8 camera motion capture system sampling at 100 Hz (Vicon Motion Systems, Oxford, UK) and synchronized with three force platforms measuring ground reaction forces and moments sampled at 1000 Hz (Ives and Wiggssworth, 2003) (Advanced Mechanical Technology Inc., Watertown, MA). Eighteen reflective markers were affixed to the right and left anterior and posterior superior iliac spines, mid thighs, lateral epicondyles, mid shanks, lateral malleoli, calcanei, and 2nd metatarsal heads. Six additional reflective markers were affixed to the right and left iliac crests, greater trochanters, medial epicondyles, and medial malleoli during static standing calibration trials as digital landmarks. These 6 additional markers were removed before beginning the gait trials, but the other 18 markers remained affixed throughout the protocol including all gait analyses and fatiguing contractions. The marker placement and gait protocols used in this study were based on Vicon’s plug-in-gait lower limb model, with the additional markers used for digital landmarking.

Gait analysis was used to capture external knee joint moments and angles. The participants walked barefoot at self-selected speeds. Gait trials were considered successful when the right foot alone fell in full contact with one of the force platforms. Five gait trials were collected for each participant at BL, PF1, and PF2. Gait trials were completed within a 10-minute window of achieving fatigue to avoid recovery (Cheng and Rice, 2005; Parijat and Lockhart, 2008). Kinematic and kinetic gait variables were calculated (C-Motion, Inc., Germantown, MD, USA) using inverse dynamics (Winter, 1984). Marker and force platform data were filtered with a dual-pass fourth order Butterworth low-pass filter at a cut-off frequency of 6 Hz. Knee joint moments were calculated using the Joint Coordinate System floating axis model (Grood and Suntay, 1983). Joint moments were normalized to body mass, and moments and angles were time normalized to the gait cycle (C-Motion, Inc., Germantown, MD, USA). Discrete measures including peaks, maximums, and minimums were extracted from joint moment and angle waveforms from 5 gait trials.

2.3.2. Electromyography

Electromyographic (EMG) signals were collected during gait and during peak torque measurements to determine lower limb muscle contributions to these activities. The EMG was pre-amplified through dual differential amplifiers with input impedance > 100,000 MΩ, CMRR > 100 dB at 65 Hz and equivalent input noise of <1.2 μV RMS nominal. Each participant-mounted amplifier had an input impedance of 31 KΩ, signal-to-noise ratio of >50 dB, and a gain of 2000 (MA 300, Motion Lab Systems, Inc., Baton Rouge, LA, USA). EMG data was bandpass filtered between 20 and 450 Hz, synchronized with the motion capture system, and sampled at 1000 Hz. Activity of the rectus femoris, vastus lateralis, and biceps femoris was monitored using stainless steel surface electrodes with a 17 mm inter-electrode distance affixed to the skin along the orientation of the muscle fibers. The rectus femoris and biceps femoris are antagonists, biarticulating the knee and hip. Altered vastus lateralis function resulted in significant changes in knee kinematics during stair climbing; therefore this muscle was selected as a possible explanatory variable for altered knee mechanics during gait (Hinman et al., 2002; Pincivero et al., 2006). A reference electrode was placed over the tibial tubercle. Electrode locations were...
determined through palpation according to the SENIAM guidelines (www.seniam.org, Enschede, Netherlands). The skin was prepared by shaving the electrode locations, vigorously wiping them with rubbing alcohol, and applying electrode gel between the skin and the electrodes. Electrode placement was verified by having the participant perform contractions to elicit activity in each muscle. A quiet trial to establish a baseline activity level for each muscle was obtained with the participants lying supine and relaxed.

2.3.3. Peak torque and fatigue

Strength and fatigue were assessed using a dynamometer (Biodex Medical Systems, Inc., Shirley, NY, USA) with a leg extension attachment. The participants were positioned for testing knee extension and flexion according to the Biodex Multi-Joint System Setup/Operation Manual. The participants performed an isokinetic warm-up of 50 contractions at 60°/s. Strength was measured as the peak knee extension and flexion torque obtained during a set of 5 maximum voluntary isometric contractions (MVICs), at 60° of knee flexion, held for 5 s each with rests of 5 s between each contraction. Peak torque for knee extensors and flexors during maximal efforts typically occurs after the first repetition, but within the first 5 repetitions for women (Pincivero et al., 2003). Peak torque was measured at baseline, and after each bout of fatiguing contractions to test for quadriiceps or hamstrings fatigue.

2.3.4. Fatigue protocol

Neuromuscular fatigue was induced in the right leg using repetitive dynamic isometric knee extensions and flexions throughout the knee range of motion on the dynamometer. An isometric fatigue protocol was chosen because it provides greater insight into functional capacity, and is an effective and efficient method of inducing neuromuscular fatigue (Cheng and Rice, 2005; McNeil and Rice, 2007; Stauber et al., 2000). The participants performed sets of 50 extensions and flexions at 50% of their peak torque during MVIC. Peak torque decreases occur primarily within the first 40–50 contractions and are minimal past 50 knee extension repetitions as individuals enter the stable endurance phase (Larsson et al., 2003; Lindström et al., 1995). Fatigue was assessed by comparing peak isometric knee extension and flexion torques, after the first and second bouts of fatigue, to the baseline values. Fatigue was defined as a drop in either peak isometric knee extension or flexion torque of at least 25% from the baseline value. Measures of peak quadriiceps torque, before and after fatiguing contractions, are a reliable indicator of knee extensor fatigue (Larsson et al., 2003). The participants performed up to 4 sets of fatiguing contractions in a row until fatigue was achieved. If fatigue was not achieved within 4 sets of fatiguing contractions, the participants were withdrawn from the study to avoid lower limb injury.

2.3.5. Dynamic stiffness

Dynamic knee stiffness was calculated as the change in knee flexion moment divided by the change in knee flexion angle during the weight acceptance portion of the gait cycle (3–15%) (Zeni and Higgins, 2009). Dynamic knee stiffness was taken as the slope of a linear regression line fitted to the data points.

2.4. Data analysis

The EMG signals were full-wave rectified and low pass filtered at 6 Hz (4th order Butterworth filter). Baseline activity obtained from the bias trial was removed from each signal, and the signals obtained from gait trials were amplitude normalized to % MVIC, meaning that 100% muscle activation is equivalent to the peak muscle activation achieved during MVIC. EMG signals obtained from gait trials were also time normalized to the gait cycle starting from heel strike of the right foot and ending with the following heel strike of the same foot (C-Motion, Inc., Germantown, MD, USA); therefore 0% of the gait cycle refers to the initial heel strike of the right foot, and 100% of the gait cycle represents the subsequent heel strike of that same foot. Muscle amplitude data from all participants was ensemble averaged to obtain one curve per muscle for all participants at BL, PF1, and PF2. Mean and peak EMG amplitudes during knee loading (3–15% of gait) and unloading (40–52% of gait) were extracted at BL, PF1, and PF2. Signals for each muscle during MVIC were passed through a Hanning window and power spectrums were calculated for the middle 2 s (2048 samples) of the 5 s MVIC with a fast Fourier transform (MATLAB, the MathWorks, Natick, MA, USA; MyoResearch XP Master Edition, Noraxon USA Inc., Scottsdale, AZ, USA). For each muscle, the median power frequency (MPF) was calculated as the frequency that divided the total power in half.

Co-activation of the biceps femoris and rectus femoris was calculated across the loading response portion of the gait cycle, from 3 to 15% of gait, and across the unloading portion of the gait cycle, from 40 to 52% of gait. These two muscles were compared since they both articulate the hip and knee, and EMG fatigue patterns are different in mono- and bi-articular muscles (Ebenbichler et al., 1998). The co-activation index was calculated using the following equation:

\[ CI = \frac{\text{EMG}_{\text{ANT}}}{\text{EMG}_{\text{ANT}} + \text{EMG}_{\text{AC}}} \times 100\% \]

where \( \text{EMG}_{\text{ANT}} \) is the magnitude of EMG from the lower activity muscle, and \( \text{EMG}_{\text{AC}} \) is the magnitude of EMG from the higher activity muscle (Kellis et al., 2003).

The mean of 5 gait trials provided one representative curve for each participant for knee moment, knee angle, and muscle activation data. These data from all participants were ensemble averaged to obtain one curve for each variable representing all participants at BL, PF1, and PF2. Repeated measures analyses of variance and pair-wise comparisons among estimated marginal means were used to assess differences in discrete gait measures, knee extension and flexion torques, and EMG amplitudes between BL, PF1, and PF2 (\( \alpha = 0.05 \)). A Bonferroni adjustment accounted for multiple comparisons.

3. Results

Four participants did not achieve a 25% decrease in MVIC after 4 sets of 50 fatiguing contractions and their data were removed from the study. These participants were not different from the rest of the sample in age, body mass, BMI, or baseline peak extension and flexion torques (\( P > 0.05 \)). Twenty participants were considered to be fatigued as they demonstrated at least the criterion 25% decrease in flexor or extensor MVIC torque. These 20 participants were (in mean (standard deviation)) 23.2 (3.1) years old, had a body mass of 63.2 (9.7) kg, and had a BMI of 23.2 (3.0) kg/m². The fatigue protocol significantly decreased peak isometric torque from BL (mean [95% confidence interval], extension: 173.5 [159.1, 187.9] Nm, flexion: 82.4 [75.4, 89.4] Nm) to PF1 (extension: 129.9 [116.4, 143.3] Nm, flexion: 58.1 [53.0, 63.2] Nm) and from BL to PF2 (extension: 128.6 [115.4, 141.9] Nm, flexion: 60.7 [54.4, 67.1] Nm) (\( P < 0.001 \)) (Table 1). Gait speed and stride length were not altered by fatigue (Table 1). Peak joint moments and angles during gait are shown in Table 1. Fatigue did not alter first or second peak knee adduction moments, or peak knee flexion angles (Table 1). Peak knee flexion moments during gait were unchanged with fatigue; however fatigue did decrease the peak knee extension moment from BL (mean [95% confidence interval], \( -0.30 [-0.37, -0.22] \text{Nm/kg} \)) to PF1 (\( -0.24 [-0.31, -0.16] \text{Nm/kg} \)), and from BL to PF2 (\( -0.24 [-0.32, -0.17] \text{Nm/kg} \)) (\( P < 0.001 \)). This decrease occurred between 40 and 50% of the gait cycle, when the heel leaves the ground, preparing for push off.

Vastus lateralis data from one participant were excluded. Due to the high level of physical exertion during the data collection and the...
length of data collection time, the vastus lateralis electrode shifted during data collection on this participant. This occurred after the first set of fatiguing contractions. The EMG amplitudes as % MVIC are displayed over the gait cycle in Fig. 3. Mean and peak EMG amplitudes (% MVIC) were not different between BL, PF1, and PF2 across either 3–15% or 40–52% of gait (Table 2). The MPF of rectus femoris decreased 6.3 Hz (8.7%) (P = 0.05) from BL to PF1 (Table 2). No differences in MPF were found between time points for other muscles. Dynamic knee stiffness was not significantly increased with fatigue (Fig. 2, Table 1). Muscle co-activation between the biceps femoris and rectus femoris was not altered with fatigue (Table 2).

4. Discussion

This study sought to identify biomechanical changes at the knee during gait before and after an isotonic fatigue protocol in healthy, young women. Two sets of quadriceps and hamstrings fatiguing contractions did not increase the peak knee adduction moment, dynamic knee stiffness, or anterior–posterior muscle co-activation around the knee; however these contractions decreased the external peak knee extension moment in late stance in these particularly strong, young women. The key implication of this study is that young women with high lower limb muscle strength retain enough strength reserve

Table 1

<table>
<thead>
<tr>
<th></th>
<th>Baseline</th>
<th>Post-fatigue 1</th>
<th>Post-fatigue 2</th>
<th>P-Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Peak isometric torque (Nm) — extension</td>
<td>173.5 (30.7)</td>
<td>129.9 (28.7)</td>
<td>128.6 (28.3)</td>
<td>P &lt; 0.001&lt;sup&gt;a&lt;/sup&gt;</td>
</tr>
<tr>
<td>Peak isometric torque (Nm) — flexion</td>
<td>82.4 (14.9)</td>
<td>58.1 (10.9)</td>
<td>60.7 (13.6)</td>
<td>P &lt; 0.001&lt;sup&gt;b&lt;/sup&gt;</td>
</tr>
<tr>
<td>Gait speed (m/s)</td>
<td>1.24 (0.15)</td>
<td>1.25 (0.13)</td>
<td>1.24 (0.13)</td>
<td>NS</td>
</tr>
<tr>
<td>Stride length (m)</td>
<td>1.37 (0.13)</td>
<td>1.37 (0.16)</td>
<td>1.37 (0.13)</td>
<td>NS</td>
</tr>
</tbody>
</table>

Knee adduction moment (Nm/kg)
- First peak: 0.37 (0.12) vs 0.39 (0.11) vs 0.38 (0.12) NS
- Second peak: 0.20 (0.12) vs 0.22 (0.10) vs 0.23 (0.12) NS

Dynamic knee stiffness (Nm/°)
- First peak flexion: 0.052 (0.027) vs 0.055 (0.017) vs 0.057 (0.018) NS
- First peak extension: -0.30 (0.16) vs -0.24 (0.16) vs -0.24 (0.15) P < 0.001<sup>a</sup>, P < 0.001<sup>b</sup>

Knee flexion angle (degrees)
- First peak flexion: 12.51 (4.49) vs 12.63 (4.41) vs 12.30 (4.02) NS
- First peak extension: -0.30 (0.14) vs -0.16 (0.06) vs -0.16 (0.06) NS

NS = non-significant; significant difference between a baseline and post-fatigue 1, b baseline and post-fatigue 2, and c post-fatigue 1 and post-fatigue 2.

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after a fatigue protocol, which resulted in reductions to maximum torque generating capacity, to perform a submaximal activity such as gait without eliciting potentially damaging knee biomechanics. Peak knee extensor and flexor moments do not exceed approximately 0.6 Nm/kg in young, healthy individuals during walking (Riley et al., 2007). Using the mean body mass as an example from the current sample (63.2 kg) this equates to a peak torque of 37.9 Nm, which is well below the amount of strength retained after the fatigue protocol. Similarly, low muscle activation demands are required for gait. Rectus femoris activity during gait in young, healthy individuals does not typically exceed approximately 20% MVIC during the knee loading portion of the gait cycle (Arsenault et al., 1986).

The gait changes observed after the fatigue protocol, which resulted in a reduction to maximum torque generation in the current study, do not imitate any obvious knee joint injury or pathology. Individuals with knee OA and joint effusion demonstrate a reduced late stance knee extensor moment similar to that associated with the fatigue protocol in the current study. However, individuals with knee OA, but without knee effusion, do not demonstrate changes in this moment (Rutherford et al., 2012). In other pathologies, a reduced knee extensor moment in late stance is typically coupled with alterations in the knee flexion angle, frontal knee moments, and/or quadriceps activity (Henriksen et al., 2010; Rutherford et al., 2012). Murdock and Hubley-Kozey (2012) reported an increased post fatigue external knee adduction moment between early and mid stance, and decreased post fatigue external knee flexion moment during early stance; however no differences were observed during late stance. While the values and waveforms obtained in the current study are similar to Murdock and Hubley-Kozey (2012), the current study found fewer post fatigue gait differences.

The participants in the current study demonstrated substantially greater quadriceps torque at baseline with the mean (standard deviation) being 173.5 (30.7) Nm compared to 143.6 (29.7) Nm in Murdock and Hubley-Kozey (2012), highlighting the difference between a physically active sample and a sedentary sample. Furthermore, Murdock and Hubley-Kozey used a maximal effort isokinetic fatiguing protocol and obtained a 40% decrease in quadriceps torque, while the current study used an isotonic fatiguing protocol at 75% of maximal effort and fatigue obtained a 40% decrease in quadriceps torque, while the current study

### Table 2

<table>
<thead>
<tr>
<th>EMG activation (% MVIC)</th>
<th>Baseline</th>
<th>Post-fatigue 1</th>
<th>Post-fatigue 2</th>
</tr>
</thead>
<tbody>
<tr>
<td>Rectus femoris</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>3–15% gait average</td>
<td>11.7 (15.8)</td>
<td>10.6 (8.2)</td>
<td>10.6 (9.5)</td>
</tr>
<tr>
<td>3–15% gait peak</td>
<td>13.5 (16.2)</td>
<td>12.8 (9.5)</td>
<td>12.7 (10.8)</td>
</tr>
<tr>
<td>40–52% gait average</td>
<td>8.5 (15.8)</td>
<td>8.0 (8.6)</td>
<td>8.0 (9.3)</td>
</tr>
<tr>
<td>40–52% gait peak</td>
<td>10.0 (16.7)</td>
<td>10.3 (10.8)</td>
<td>10.0 (11.4)</td>
</tr>
<tr>
<td>Vastus lateralis</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>3–15% gait average</td>
<td>13.6 (7.2)</td>
<td>13.1 (6.9)</td>
<td>11.8 (5.2)</td>
</tr>
<tr>
<td>3–15% gait peak</td>
<td>17.0 (8.1)</td>
<td>16.4 (9.0)</td>
<td>14.1 (6.2)</td>
</tr>
<tr>
<td>40–52% gait average</td>
<td>6.3 (7.2)</td>
<td>6.3 (7.1)</td>
<td>5.8 (4.5)</td>
</tr>
<tr>
<td>40–52% gait peak</td>
<td>8.0 (9.4)</td>
<td>8.4 (10.0)</td>
<td>6.7 (5.1)</td>
</tr>
<tr>
<td>Biceps femoris</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>3–15% gait average</td>
<td>11.4 (9.5)</td>
<td>10.7 (7.0)</td>
<td>12.0 (8.2)</td>
</tr>
<tr>
<td>3–15% gait peak</td>
<td>13.9 (9.7)</td>
<td>13.3 (8.3)</td>
<td>14.9 (9.3)</td>
</tr>
<tr>
<td>40–52% gait average</td>
<td>9.4 (9.3)</td>
<td>7.8 (5.8)</td>
<td>8.5 (6.7)</td>
</tr>
<tr>
<td>40–52% gait peak</td>
<td>11.5 (11.1)</td>
<td>9.9 (6.9)</td>
<td>10.4 (7.5)</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Co-activation index (%)</th>
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</thead>
<tbody>
<tr>
<td>Rectus femoris</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>3–15% gait</td>
<td>33.3 (12.8)</td>
<td>31.0 (11.7)</td>
<td>29.1 (14.2)</td>
</tr>
<tr>
<td>40–52% gait</td>
<td>27.7 (14.5)</td>
<td>28.3 (12.7)</td>
<td>28.6 (14.9)</td>
</tr>
<tr>
<td>Vastus lateralis</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>3–15% gait</td>
<td>72.7 (15.6)</td>
<td>66.4 (8.4)</td>
<td>67.4 (19.1)</td>
</tr>
<tr>
<td>40–52% gait</td>
<td>73.6 (14.9)</td>
<td>69.3 (12.2)</td>
<td>70.5 (9.7)</td>
</tr>
<tr>
<td>Biceps femoris</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>3–15% gait</td>
<td>64.4 (11.8)</td>
<td>69.2 (12.4)</td>
<td>67.7 (10.7)</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Median power frequency (Hz)</th>
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<tbody>
<tr>
<td>Rectus femoris</td>
<td>72.7 (15.6)</td>
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<td>67.4 (19.1)</td>
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<td>67.7 (10.7)</td>
</tr>
</tbody>
</table>

* Significant difference (*P* < 0.05) between baseline and post-fatigue 1 values.

Fig. 3. Muscle activation (% MVIC) of the a. biceps femoris, b. rectus femoris, and c. vastus lateralis over the gait cycle at BL, PF1, and PF2, where n = 20 for rectus femoris and biceps femoris, and n = 19 for vastus lateralis.
discrepancy could be explained by differences in the sex, type of fatigue, muscle groups, and level of physical activity between the participants in these studies. Men experience a greater peak external knee flexion moment immediately following heel strike during gait than women (Kerrigan et al., 1998). The gender difference at this point in the stance cycle may have attenuated the results of fatigue in the current study compared to previous studies. Murdock and Hubley-Kozey (2012) studied 20 sedentary young adults, half of whom were women, while Parijat and Lockhart (2008) evaluated 10 men and 6 women who were healthy, but no further details on physical fitness were given. In the current study, all participants were physically active, young women. Furthermore, participants in the current study walked barefoot to eliminate alterations in gait patterns caused by footwear including inflated lower extremity joint loads (Shakoor and Block, 2006), while other studies may have analyzed gait while wearing shoes (Murdock and Hubley-Kozey, 2012; Parijat and Lockhart, 2008).

The decreased peak extension moment during late stance was not explained by a difference in co-activation or knee angle before and after fatigue. A possible explanation is an increase in the acceleration of the lower limb during this portion of stance, as a mechanism to decrease the duration of load applied to the knee. Participants retained at least 75% of their maximum strength, leaving plenty of reserve for gait (Table 1). However, post fatigue gait did not require a significant increase in muscle activation (average or peak) (Table 2, Fig. 3) to maintain pre fatigue mechanics. This indicates that the strenuous fatigue protocol did not alter the knee biomechanics in a way that is consistent with gait patterns associated with knee injury or pathology.

Dyskinesia during knee loading was not altered by the fatigue protocol. Mechanical knee stiffness is poorly correlated with self-reported knee stiffness (Dixon et al., 2010); however it indicates the amount of torque being used to rotate the knee in the sagittal plane. The minimal biomechanical changes in this study were observed only during knee unloading, pre-swing. This measure of knee stiffness focuses on the knee loading portion of gait; therefore it did not capture the portion of the gait cycle that was altered by the fatigue protocol.

Quadiceps fatigue has previously been induced by repetitive isokinetic contractions at maximal effort (Murdock and Hubley-Kozey, 2012; Parijat and Lockhart, 2008; Pincivero et al., 2001). This study aimed to better simulate fatigue caused by repetitive performance of activities of daily living (ADLs). ADLs have been characterized as bursts of activity requiring varying velocities of joint motion, under relatively constant loads applied to both knee extensors and flexors (Cheng and Rice, 2005); however activities such as walking are not strictly isokinetic or isotonic tasks as contraction velocities, and both internal and external loads, change throughout the cyclical movement. As a step toward a more functionally relevant representation of these ADLs, an isotonic fatigue protocol was used where the load was held constant while undergoing variations in angular velocity, applied to knee extensors and flexors. The current study attempted to induce fatigue in a similar manner to that induced by ADLs and ensure that fatigue was maintained throughout the gait trials; however achieving fatigue quickly in a laboratory setting may elicit faster rates of recovery compared to fatigue caused by repetitive ADLs. The high intensity, short duration, and repetitive contractions used to the knee in this study are not particularly representative of the fatigue that affects individuals with knee injury or pathology. Typically, fatigue occurs from low intensity, long duration, and repetitive loads on the knee, through ADLs such as squatting, lunging, lifting, or long distance walking. This is different to the genesis and management of knee osteoarthritis (rheum. dis. clin. n. am. 34 (3), 731–754). and external loads, change throughout the cyclical motion. As a step toward a more functionally relevant representation of these ADLs, an isotonic fatigue protocol was used where the load was held constant while undergoing variations in angular velocity, applied to knee extensors and flexors. The current study attempted to induce fatigue in a similar manner to that induced by ADLs and ensure that fatigue was maintained throughout the gait trials; however achieving fatigue quickly in a laboratory setting may elicit faster rates of recovery compared to fatigue caused by repetitive ADLs. The high intensity, short duration, and repetitive contractions used to the knee in this study are not particularly representative of the fatigue that affects individuals with knee injury or pathology. Typically, fatigue occurs from low intensity, long duration, and repetitive loads on the knee, through ADLs such as squatting, lunging, lifting, or long distance walking. This is difficult to reproduce in a laboratory setting; however future studies should attempt to quantify biomechanical changes induced by low intensity, repetitive fatigue.

Analysis techniques in the currently study differ from previous studies of knee mechanics during gait. The current study used discrete analyses to identify differences, before and after the fatigue protocol, between waveform peaks and means during specific portions of the gait cycle. This analysis method disregards differences between waveform shape, magnitude, and timing. Our sample size limited the analyses to discrete parameters rather than performing full-wave analyses, such as principal component analysis (Deluzio and Astephene, 2007; Deluzio et al., 1997). This discrete analysis did not detect substantial differences in waveform peaks and means, likely due to variability between participants.

The mechanism for fatigue alters the rate of change in muscle activation in both young and older adults, and varies between muscle groups. In young adults, the rectus femoris, vastus lateralis, and vastus medialis display different activation properties under isometric and dynamic conditions (Callahan et al., 2009; Pincivero et al., 2006). Relative to the absolute strength of the participants in the current study, tasks such as walking are low demand in terms of strength requirements. The strong, healthy participants clearly maintained enough strength to perform the fatigue activity without substantial changes to their knee biomechanics, which could be considered harmful. Alterations in knee mechanics due to fatigue may be more prominent in higher demand tasks such as running or jumping.

In conclusion, this study indicates that a high intensity lower limb fatigue-inducing activity did not alter the mechanical environment of the knee joint during walking in young, active women in a way that could increase their risk for injury.

Conflict of interest statement

The authors of this paper are not in any conflict of interest with regard to the work presented.

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