Journal of Biomechanics 156 (2023) 111688

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech

Effects of induced motor fatigue on walking mechanics and energetics☆

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ARTICLE INFO

Keywords:
Motor fatigue
Walking
Metabolic cost
Efficiency
Joint power

ABSTRACT

Lower-body robotic exoskeletons can be used to reduce the energy demand of locomotion and increase the endurance of wearers. Understanding how motor fatigue affects walking performance may lead to better exoskeleton designs to support the changing physical capacity of an individual due to motor fatigue. The purpose of this study was to investigate the effects of motor fatigue on walking mechanics and energetics. Treadmill walking with progressively increased incline gradient was used to induce motor fatigue. Twenty healthy young participants walked on an instrumented treadmill at 1.25 m/s and 0° of incline for 5 min before (PRE) and after (POST) motor fatigue. We examined lower-limb joint mechanics, metabolic cost, and the efficiency of positive mechanical work (η^{+}_{work}). Compared to PRE, participants had increased net metabolic power by ~14% ($p < 0.001$) during POST. Participants also had increased total-limb positive mechanical power (Total P^{+}_{mech}) by ~4% during POST ($p < 0.001$), resulting in a reduced η^{+}_{work} by ~8% ($p < 0.001$). In addition, the positive mechanical work contribution of the lower-limb joints during POST was shifted from the ankle to the knee while the negative mechanical work contribution was shifted from the knee to the ankle (all $p < 0.017$). Although greater knee positive mechanical power was generated to compensate for the reduction in ankle positive power after motor fatigue, the disproportionate increase in metabolic cost resulted in a reduced walking efficiency. The findings of this study suggest that powering the ankle joint may help delay the onset of the lower-limb joint work redistribution observed during motor fatigue.

1. Introduction

Lower-body robotic exoskeletons can be used to reduce the energy demand of locomotion and increase the endurance of wearers (Sawicki et al., 2020). It was shown that lower-body exoskeletons successfully reduce the net metabolic cost during walking by 6–12 % when assisting at the ankle, by ~5% when assisting at the knee, by 9.3–19.8% when assisting at the hip, or by ~14.9% when assisting multiple joints in healthy young individuals (Sawicki et al., 2020). Slade et al (2022) demonstrated that implementing an adaptive controller that personalizes exoskeleton assistance provided at the ankle can reduce metabolic cost by ~23% while walking at the constant speed on the treadmill or reduce energy cost of transport by ~17% under overground walking conditions where the walking speed might be fluctuated (Slade et al., 2022). While there have been significant advancements made in exoskeleton hardware and controllers to improve locomotion economy (Lv et al., 2018; Sawicki et al., 2020; Siviy et al., 2022; Young and Ferris, 2017), increasing exoskeletons' functionality to support the changing physical capacity of an individual due to motor fatigue may further enhance exoskeleton wearers' performance.

Identifying the effects of motor fatigue on human walking mechanics may lead to better exoskeleton designs to delay the occurrence of motor fatigue during a prolonged walk. Motor fatigue has been shown to reduce neuromuscular performance of an individual (Barry and Enoka, 2007; Proske, 2019), leading to adaptive changes in gait spatiotemporal parameters to maintain stability during walking (Arvin et al., 2015; Kao et al., 2018). Healthy individuals demonstrated a cautious gait strategy following motor fatigue by walking with reduced speed, reduced step length, and/or increased step width (Granacher et al., 2010). In addition, the variability in step width, lower-limb joint kinematics, and the trunk movement during walking was found to be significantly increased after motor fatigue (Kao et al., 2018). Although the changes in gait

☆ The study was partially published as an abstract in the Proceedings of North American Congress on Biomechanics (NACOB), Ottawa, 2022.

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https://doi.org/10.1016/j.jbiomech.2023.111688
Received 2 February 2023; Received in revised form 11 May 2023; Accepted 12 June 2023
Available online 16 June 2023

spatiotemporal or kinematic parameters following motor fatigue were shown to be statistically significant, the magnitude of the changes was modest, ranging from a small to medium effect size, except for the changes in trunk acceleration and velocity variability shown with a large effect size (Kao et al., 2018; Santos et al., 2019a; Santos et al., 2019b). Zhang et al (2022) studied the effects of motor fatigue on walking kinetics at self-selected speeds, showing that participants had reduced the power generation of all three lower-limb joints (Zhang et al., 2022). On the other hand, power absorption was redistributed, decreasing for the ankle and knee joints but increasing at the hip. However, the self-selected speeds for each participant were not matched before and after motor fatigue. Thus, those changes observed in joint kinetics might simply reflect the mechanical requirement for walking at different speeds rather than the effects of motor fatigue. While the effects of motor fatigue on walking kinematics have been studied extensively, the information on how motor fatigue impacts walking mechanics and energetics is limited.

The occurrence of motor fatigue could also reduce an individual's efficiency in task performance. Previous studies investigating the effects of a fatiguing run showed an increase in energy expenditure at the marathon pace (Candau et al., 1998; Petersen et al., 2007), indirectly indicating a reduced mechanical efficiency of running. In addition, it has been shown that the ankle joint reduced its contribution to the total lower-limb positive work while the knee and/or hip joint increased their contribution to the total lower-limb positive work after a fatiguing run, demonstrating a shift of positive work contribution from the distal to proximal joints (Cigoja et al., 2022; Sanno et al., 2018). The findings from these running studies suggested that this redistribution of joint work among the lower-limb joints could partly explain the increased energy expenditure associated with running-induced fatigue due to the requirement of higher muscle volume activation at the proximal joints (Cigoja et al., 2022; Sanno et al., 2018).

The purpose of this study was to investigate how motor fatigue affects walking mechanics and energetics. We examined the changes in lower-limb joint mechanics, metabolic cost, and the efficiency of positive mechanical work before and after walking-induced fatigue. We hypothesized that healthy participants would show increased metabolic cost and reduced mechanical efficiency during walking after motor fatigue. Although walking and running have different patterns of kinetics and kinematics, ankle joint plays a dominant role in generating positive mechanical power and work for walking and running, especially during the stance phase of the locomotion (Farris and Sawicki, 2012; Jin and Hahn, 2018). Based on the findings of prolonged running studies (Cigoja et al., 2022; Sanno et al., 2018), we also expected to see that motor fatigue would change the work contributions of the lower-limb joints during walking, characterized by a shift of positive work contribution from the distal to proximal joints following walking-induced fatigue.

2. Methods

2.1. Participants

Twenty healthy young participants (9 females: age: 24.0 ± 7.0 years, body height: 1.61 ± 0.07 m, body mass: 62.3 ± 8.6 kg; 11 males: age: 22.8 ± 2.8 years, body height: 1.80 ± 0.09 m, body mass: 86.9 ± 12.1 kg, mean ± STD) gave written informed consent to participate in the study, which was approved by the Institutional Review Board of the University (#20–057). Participants were excluded if they had any medical condition, history of major leg injuries, or pain in the legs or spine that limits locomotion or the capacity to perform exercise.

2.2. Experimental procedures

Participants were first given time to familiarize themselves to walking on an instrumented, split-belt treadmill (M−Gait, Motek, Netherlands) prior to the testing. A vertical jump test was used to identify when the participant was being fatigued (Cooper et al., 2020; Sams et al., 2018; Watkins et al., 2017). Participants were instructed to perform 3 vertical squat jumps (i.e., non-countermovement jumps) with one arm crossed the chest and using another arm to reach and tap the vanes of Vertec vertical jump tester (JumpUSA, Sunnyvale, CA, USA). Participants' maximal jump height among all attempts was recorded. Following the jumps, as the baseline, participants walked on the treadmill at 1.25 m/s and 0° of incline for 5 min, which was repeated after participants were being fatigued. During the motor fatiguing protocol, the inclination angle of the treadmill was increased by 2.5° every five minutes until participants had a Borg rating of perceived exertion (RPE) $> 17/20$ and reached ~85% of their maximum age-predicted heart rate or due to voluntary exhaustion, where they expressed to pause the protocol due to fatigue. After the conditions were met, we paused the motor fatiguing protocol and asked participants to perform another 2–3 squat jumps. If participants' vertical jump height was reduced by 20% (i.e., $\leq$ 80% of their maximal vertical jump height), the motor fatiguing protocol would be terminated. Otherwise, the motor fatiguing protocol would be resumed from the last incline setting achieved. The highest inclination angle was 5° for one participant, 7.5° for nine participants and 10° for ten participants. The total treadmill walking time to induce motor fatigue was 37.4 ± 8.6 min, which did not include the 5-minute, post-motor fatiguing walking trial.

2.3. Metabolic data collection and analysis

Rates of oxygen consumption and carbon dioxide production (VO_2 and VCO_2) were recorded using a portable gas exchange indirect calorimetry system (K5, Cosmed, Rome, Italy) during a 3-minute baseline standing and the two 5-minute level walking trials before and after the motor fatiguing protocol. We averaged the data from the last minute of the quiet standing trial and the data from the last two minutes of the walking trials to compute metabolic rate. We computed net oxygen consumption rate (net VO_2, ml/min/kg) during walking by subtracting the baseline VO_2 during quiet standing and normalizing to participants' body mass. We also derived net metabolic powers (net P_{met}, W/kg) during walking by converting the rates of oxygen consumption and carbon dioxide production using the abbreviated Weir equation (**Eq. (1)** (Weir, 1990) and subtracting the metabolic power during standing.

$$Metabolic\,power(W/kg) = (3.94{*}V{\cdot}O_2 + 1.10{*}\ V{\cdot}CO_2){*}(1/60)(\min/s){*}4.184 \qquad (1)$$

2.4. Kinematic and kinetic data collection and analysis

We collected three-dimensional kinematics using an 8-camera video system (100 Hz, Motion Analysis Corporation, Santa Rosa, CA, USA) with 40 reflective markers attached to the shoes (on the distal phalanx of the great toes, the first and fifth metatarsal heads, and heels), ankles (lateral and medial malleoli), lateral shanks (4-marker cluster plate), knees (medial and lateral epicondyles), lateral thighs (4-marker cluster plate), pelvis (left and right anterior superior iliac spine, posterior superior iliac spine, top of the iliac crest), and hips (greater trochanters) for seven body segments. Ground reaction force (GRF) data were recorded during walking (1000 Hz, M−Gait, Motek, Netherlands). Kinematic and GRF data post-processing were performed using Visual 3D software (C-Motion Inc., Germantown, MD, USA) such as data filtering (4th order Butterworth low-pass filter with a cutoff frequency of 10 Hz for both marker position and GRF data), gait event detection, and gait cycle normalization (time normalized to 100 % of stride cycle, from heel strike to heel strike), and computation of joint moment and power at ankle, knee, and hip via the inverse dynamic analyses. Overall support moment was derived by summing extensor moments from the hip, knee and ankle joints (Winter, 1980).

Average positive and negative mechanical work values for each joint over the strides were calculated by integrating joint power data of each joint with respect to time over discrete periods of only positive or negative joint power of each stride and summed the work values over those positive or negative periods, respectively (Farris and Sawicki, 2012). Average positive and negative mechanical power values of each joint were calculated by dividing the average positive and negative mechanical work values of each joint by the average step time for the trial. The total limb average positive power (Total P^{+}_{mech}) was computed by summing the average positive mechanical power value at each joint (Total P^{+}_{mech} = ankle P^{+}_{mech} + knee P^{+}_{mech} + hip P^{+}_{mech}) whereas the total limb negative mechanical power (Total P^{-}_{mech}) was computed by summing the average negative mechanical power value at each joint (Nuckols et al., 2020), which were used to derive the percentage contribution of an individual joint to the total limb average mechanical power (e.g., ankle P^{+}_{mech}/Total P^{+}_{mech}, ankle P^{-}_{mech}/Total P^{-}_{mech}). We also computed the efficiency of positive mechanical work by dividing the total limb positive mechanical power to the net metabolic power (η^{+}_{work} = Total P^{+}_{mech}/net P_{met}) (Farris and Sawicki, 2012). Lastly, the average net power for each joint was calculated by summing the average positive and negative power values at each joint (e.g., net P_{mech} of ankle = ankle P^{+}_{mech} + ankle P^{-}_{mech}).

2.5. Statistical analysis

We used paired *t*-test to test for differences in the outcome variables before and after the fatiguing protocol (i.e., PRE and POST). Outcome measures at the whole-body level include the net metabolic power, total limb average mechanical power, and the efficiency of positive mechanical work. Outcome measures at the joint level include the average positive and negative power of individual joint, percentage contribution of individual joint to the total limb mechanical power, and the net mechanical power of individual joint. We set the significance level at 0.05. To account for the comparisons made at each individual joint for those joint-level outcome measures, we applied Bonferroni's correction (adjusted p = 0.017). We used Cohen's *d* to calculate the effect size (ES) for quantifying the magnitude of changes in the outcome variables before and after the fatiguing protocol. ES values of 0.21–0.49 indicate small effect, 0.50–0.79 indicate medium effect, and ≥ 0.80 indicate large effect (Cohen, 1988). All statistical analyses were performed in SPSS version 28 (IBM, New York, USA).

3. Results

Ankle and hip moment profiles were similar after the motor fatiguing protocol (POST) compared to the baseline (PRE) while the peaks of knee extensor moment during early to mid-stance (PRE: 0.62 ± 0.09 Nm/kg, POST: 0.56 ± 0.15 Nm/kg, p = 0.016) and at late stance during push-off (PRE: 0.24 ± 0.06 Nm/kg, POST: 0.20 ± 0.08 Nm/kg, p = 0.005) were significantly smaller during POST compared to PRE (Fig. 1). These changes resulted in a significant reduction of the overall support moment at the mid-stance during POST compared to PRE (PRE: 0.69 ± 0.21 Nm/kg, POST: 0.61 ± 0.19 Nm/kg, p = 0.002).

For the joint power profiles (Fig. 2), there was a significant increase in the negative ankle power peak at late stance during POST compared to PRE (PRE: −0.95 ± 0.36 W/kg, POST: −1.11 ± 0.35 W/kg, p = 0.001) whereas the peak of positive ankle power at push-off was non-significantly smaller during POST than PRE (PRE: 2.56 ± 0.52 W/kg, POST: 2.47 ± 0.35 W/kg, p = 0.087). For the knee joint power profile, participants had significantly less negative knee power at push-off during POST (PRE: −0.99 ± 0.29 W/kg, POST: −0.83 ± 0.32, p = 0.007). For the hip joint power profile, the peak of positive hip power during early swing was non-significantly smaller during POST compared to PRE (PRE: 0.74 ± 0.22 W/kg, POST: 0.68 ± 0.21 W/kg, p = 0.078).

For the average positive mechanical power (Table 1, Fig. 3), participants had significantly greater average positive knee power (knee P^{+}_{mech}) (PRE: 0.17 ± 0.03 W/kg, POST: 0.21 ± 0.04 W/kg, p < 0.001) while having similar average positive ankle and hip power (ankle P^{+}_{mech} and hip P^{+}_{mech}) during POST compared to PRE, resulting in a significant increase in total limb average positive power (Total P^{+}_{mech}) during POST by 4% (PRE: 0.95 ± 0.15 W/kg, POST: 0.99 ± 0.16 W/kg, p < 0.001). Correspondingly, participants had increased net VO_2 (PRE: 10.91 ± 1.87 ml/min/kg, POST: 12.45 ± 2.07 ml/min/kg, p < 0.001) and net P_{met} (PRE: 3.66 ± 0.64 W/kg, POST: 4.14 ± 0.70 W/kg, p < 0.001) significantly during POST compared to PRE by ~14% (Table 1). As a result, these changes resulted in a significant reduction in the efficiency of the positive mechanical work (η^{+}_{work}) during POST by 8% (PRE: 0.26 ± 0.05, POST: 0.24 ± 0.05, p < 0.001). For the average negative mechanical power (Table 1, Fig. 3), participants had significantly greater average negative ankle power (ankle P^{-}_{mech}) (PRE: −0.43 ± 0.09 W/kg, POST: −0.48 ± 0.10 W/kg, p = 0.011) but had significantly less average negative knee power (knee P^{-}_{mech}) (PRE: −0.52 ± 0.08 W/kg, POST:

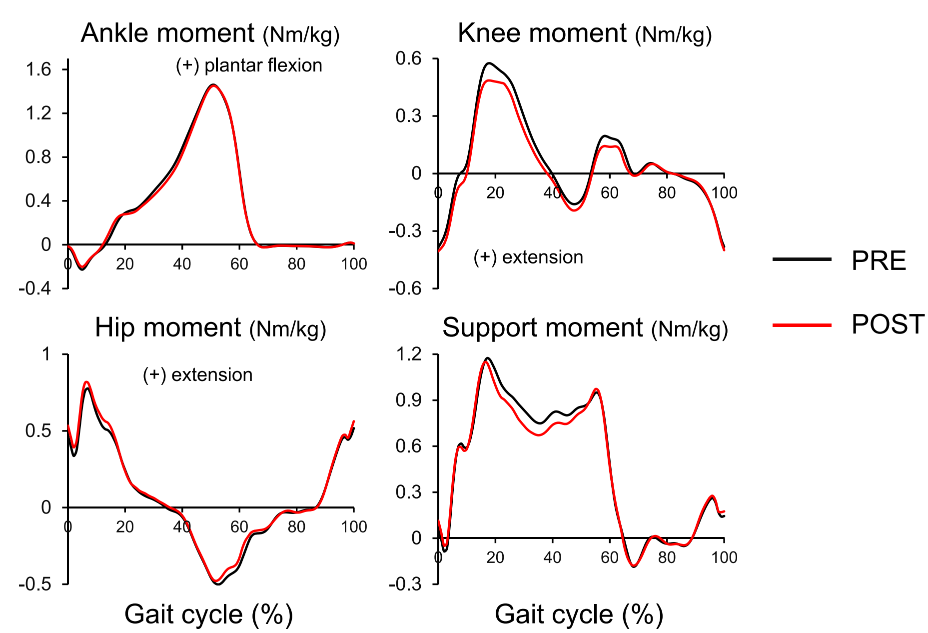

Fig. 1. Joint moments of the ankle, knee, and hip, and the overall support moment before (PRE, black lines) and after (POST, red lines) the motor fatiguing protocol. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

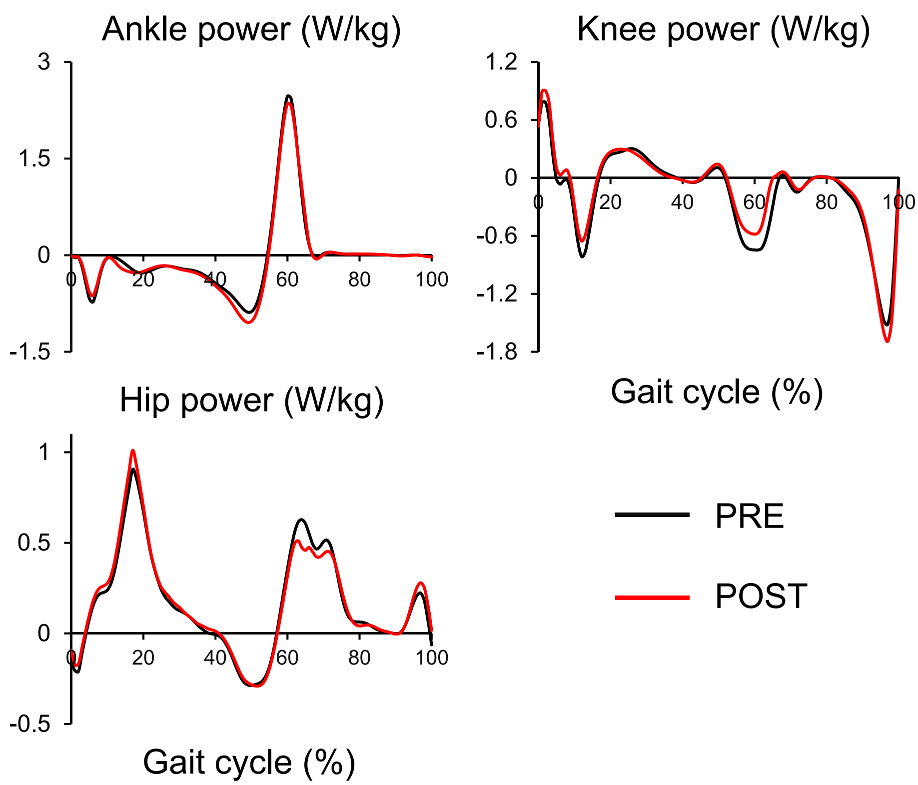

Fig. 2. Joint power profiles of the ankle, knee, and hip before (PRE, black lines) and after (POST, red lines) the motor fatiguing protocol. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

Table 1
Energetics and mechanics data when walking at 1.25 m/s before (PRE) and after (POST) the motor fatiguing protocol (mean ± STD).

parameters	PRE	POST	p-value	Cohen's d
Total P^{+}_{mech} (W/kg)	0.95 ± 0.15	0.99 ± 0.16	**< 0.001***	0.93
Ankle P^{+}_{mech} (W/kg)	0.33 ± 0.06	0.32 ± 0.05	0.069	0.35
Knee P^{+}_{mech} (W/kg)	0.17 ± 0.03	0.21 ± 0.04	**< 0.001#**	1.51
Hip P^{+}_{mech} (W/kg)	0.44 ± 0.12	0.47 ± 0.14	0.025	0.47
Total P^{-}_{mech} (W/kg)	−1.06 ± 0.14	−1.06 ± 0.16	0.453	0.03
Ankle P^{-}_{mech} (W/kg)	−0.43 ± 0.09	−0.48 ± 0.10	**0.011#**	0.56
Knee P^{-}_{mech} (W/kg)	−0.52 ± 0.08	−0.48 ± 0.09	**0.002#**	0.73
Hip P^{-}_{mech} (W/kg)	−0.11 ± 0.06	−0.11 ± 0.05	0.389	0.06
net VO_2 (ml/min/kg)	10.91 ± 1.87	12.45 ± 2.07	**< 0.001***	1.86
net P_{met} (W/kg)	3.66 ± 0.64	4.14 ± 0.70	**< 0.001***	1.78
η^{+}_{work}	0.26 ± 0.05	0.24 ± 0.05	**< 0.001***	1.01

*: $p < 0.05$; #: $p < 0.017$ (Bonferroni correction).

−0.48 ± 0.09 W/kg, p = 0.002) while having similar average negative hip power (hip P^{-}_{mech}) during POST, resulting in similar total limb average negative power (Total P^{-}_{mech}) between POST and PRE.

The average net power of the ankle was negative in both conditions and the magnitude of the negative net ankle power increased significantly after being fatigued (POST) (PRE: −0.10 ± 0.12 W/kg, POST: −0.16 ± 0.11 W/kg, $p = 0.009 < 0.017$) (Table 2). For the knee joint, the average net power was also negative; however, the magnitude of the negative net knee power was reduced significantly during POST (PRE: −0.34 ± 0.08 W/kg, POST: −0.27 ± 0.10 W/kg, $p < 0.001$). For the hip joint, the average net power was positive and was non-significantly increased during POST (PRE: 0.33 ± 0.15 W/kg, POST: 0.36 ± 0.17 W/kg, $p = 0.045 > 0.017$).

For the relative contribution of each joint to the total limb average mechanical power (Table 3), ankle joint had significantly reduced its contribution to the total limb positive power (PRE: 35.5 ± 6.2 %, POST: 32.7 ± 5.7%, $p = 0.002 < 0.017$) but increased its contribution to the total limb negative power (PRE: 40.8 ± 5.2 %, POST: 44.9 ± 4.7 %, $p < 0.001$) during POST compared to PRE. On the contrary, knee joint increased its contribution to the total limb positive power (PRE: 18.5 ± 2.7 %, POST: 21.0 ± 3.8 %, $p < 0.001$) but reduced its contribution to the total limb negative power during POST (PRE: 49.1 ± 6.1 %, POST: 45.0 ± 4.9%, $p < 0.001$). The contribution of hip to total limb positive or negative power was not changed at POST. To further investigate the joint work redistribution between the ankle and knee joints, we computed the ratio between ankle and knee average positive and negative power (ankle P^{+}_{mech}/Knee P^{+}_{mech} and ankle P^{-}_{mech}/Knee P^{-}_{mech}, respectively). After being fatigued (POST), the ratio between the ankle and knee positive power (ankle P^{+}_{mech}/Knee P^{+}_{mech}) was significantly reduced (PRE: 1.94 ± 0.39, POST: 1.58 ± 0.25, $p < 0.001$, Cohen's *d*: 1.07) while the ratio between ankle and knee negative power (ankle P^{-}_{mech}/Knee P^{-}_{mech}) was significantly increased (PRE: 0.85 ± 0.21, POST: 1.02 ± 0.20, $p < 0.001$, Cohen's *d*: 1.21) compared to the baseline (PRE).

4. Discussion

This study investigated the effects of walking-induced fatigue on walking mechanics and energetics. Our findings support the hypothesis that healthy participants would show increased metabolic cost and reduced mechanical efficiency during walking after motor fatigue. We found that healthy participants had significantly increased energy expenditure when walking at the same speed after motor fatigue, indicating a deteriorated walking economy. We also found that there was an increase in the total-limb mechanical power generation during walking after motor fatigue. However, the disproportionate increase in net metabolic power relative to the increase in mechanical power generation following motor fatigue resulted in a reduced efficiency of positive mechanical work during walking. This is the first study demonstrating that motor fatigue has a direct impact on the mechanical efficiency of walking.

We also found that motor fatigue would change the mechanical work

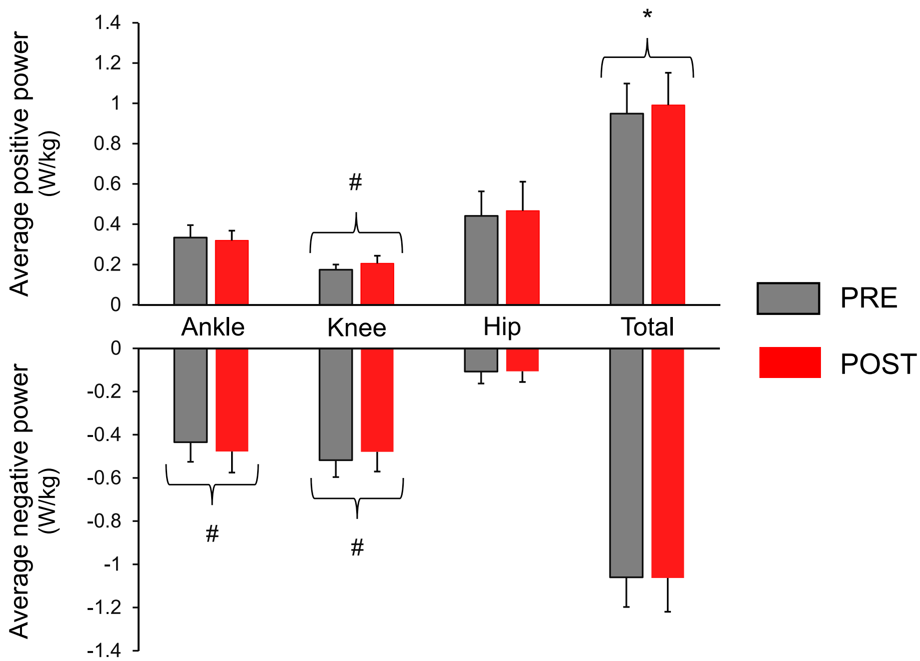

Fig. 3. Average positive and negative powers of the ankle, knee, hip, and the total limb before (PRE, grey bars) and after (POST, red bars) the motor fatiguing protocol. Error bars represent 1 STD. * indicates significant difference between the PRE and POST ($p < 0.05$). # indicates significant difference between the PRE and POST at the joint level ($p < 0.017$). (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

Table 2
Lower limb joint average net power (net P_{mech}, W/kg) of ankle, knee, and hip joints during walking before (PRE) and after (POST) the motor fatiguing protocol (mean ± STD).

Net power (net P_{mech}, W/kg)	PRE	POST	p-value	Cohen's d
Ankle	−0.10 ± 0.12	−0.16 ± 0.11	**0.009**[#]	0.58
Knee	−0.34 ± 0.08	−0.27 ± 0.10	**< 0.001**[#]	1.28
Hip	0.33 ± 0.15	0.36 ± 017	0.045	0.40

[#] : $p < 0.017$ (Bonferroni correction).

contribution of the lower-limb joints after walking-induced motor fatigue, similar to the previous findings on the effects of a prolonged run (Cigoja et al., 2022; Sanno et al., 2018). Our results show a shift of positive work contribution from the ankle joint to the knee joint while having a shift of negative work contribution from the knee joint to the ankle joint. These findings indicate that ankle joint's capability in generating positive mechanical power was affected by the walking-induced motor fatigue the most and the reduced ankle push-off might have been compensated by an increase in mechanical power generation at the knee joint, resulting in higher metabolic cost during walking following motor fatigue. Activating proximal leg muscles (e.g., knee extensors) was thought to be more energetically costly compared to activating distal leg muscles (e.g., ankle plantar flexors) due to the increased muscle volume activation (Beck et al., 2019; Cigoja et al., 2022; Sawicki et al., 2009). Our results, along with previous findings from the prolonged run studies, suggest that the occurrence of this distal-to-proximal joint work redistribution could potentially serve as a precursor of motor fatigue during walking or running.

In addition to the effects of a prolonged walk or run, similar joint work redistribution among the lower-limb joints was also observed when walking at a constant speed with reduced propulsive force generation at push-off, accompanied by higher metabolic costs (Browne and Franz, 2017; Pieper et al., 2021). Pieper et al (2021) found that walking with diminished push-off imposed metabolic penalty due to an increase in the total limb work produced by proximal leg muscles (Pieper et al., 2021). Conversely, when walking with increased ankle push-off, Lewis and Ferris (2008) demonstrated a lower hip flexor moment and hip positive power (Lewis and Ferris, 2008). These results suggest that amplifying ankle push-off via the exoskeleton assistance during walking may help prevent excessive effort from the proximal leg muscles in addition to reducing plantar flexor muscle activations (Kao et al., 2010).

Ankle joint was found to be a less metabolic-costly source for mechanical power generation during walking compared to other lower-extremity joints (Sawicki and Ferris, 2008; Sawicki et al., 2009). It was shown that the efficiency of the positive mechanical work at the ankle was much higher than the estimated efficiency of the positive mechanical work at the hip and knee (Sawicki et al., 2009), suggesting

Table 3
The percentage of total limb average positive and negative power contributed at the ankle, knee, and hip joints during walking before (PRE) and after (POST) the motor fatiguing protocol (mean ± STD).

Contribution of individual joint (%)	PRE			POST			p-value and Cohen's *d*		
	Ankle	Knee	Hip	Ankle	Knee	Hip	Ankle	Knee	Hip
Total limb positive power (Total P^{+}_{mech})	35.5 ± 6.2	18.5 ± 82.7	45.9 ± 87.3	32.7 ± 85.7	21.0 ± 83.8	46.3 ± 88.5	**0.002**[#] 0.71	**< 0.001**[#] 0.99	0.343 0.09
Total limb negative power (Total P^{-}_{mech})	40.8 ± 85.2	49.1 ± 86.1	10.1 ± 85.1	44.9 ± 84.7	45.0 ± 84.9	10.1 ± 85.0	**< 0.001**[#] 1.18	**< 0.001**[#] 1.05	0.498 0.00

[#] : $p < 0.017$ (Bonferroni correction).

that powering the ankle joint during walking is less “metabolically” beneficial (Sawicki and Ferris, 2008). However, our results suggest that providing exoskeleton assistance at the ankle joint to meet the requirement of mechanical work during walking may help delay the joint work redistribution from distal to proximal joints observed during motor fatigue and thus, reducing energy expenditure of walking following motor fatigue.

One major limitation of this study is that we used vertical jump height reduction as the primary criterion for achieving motor fatigue in addition to the Borg rating of RPE and heart rate level. Although we did not provide participants with the information on the criteria for terminating the motor fatiguing protocol, the reduction of squat jump height could have been also affected by the perceived fatigability (e.g., psychological state such as motivation, homeostasis state such as hydration) in addition to motor fatigue (Enoka and Duchateau, 2016). In other words, some of our participants might not have been as “fatigued” as we planned to be and thus, our results could have underestimated the effects of motor fatigue on walking mechanics and energetics.

Another major limitation of this study is that this study only examined joint mechanical work in the sagittal plane. It was shown that the average and variability of step width, the center of mass (COM) displacement, and peak COM velocity in the mediolateral direction were increased after being fatigued (Chen and Chou, 2022; Kao et al., 2018). It is possible that these gait changes following motor fatigue might also affect non-sagittal plane joint kinetics and alter the total work produced at each lower-limb joint or summed for the entire lower-extremity, which would have contributed to the increased metabolic cost to some extent (Donelan et al., 2001; O’Connor et al., 2012). Thus, our findings on the reduced mechanical efficiency or joint work redistribution after motor fatigue would be only valid when referred to the sagittal-plane work but not the overall mechanical work. Lastly, the current study aimed to understand the effects of motor fatigue on walking mechanics and energetics in healthy young adults, with a potential application in exoskeleton designs for providing mechanical assistance based on the wearers’ physical status. Testing other populations will be warranted for future studies to examine how other factors (e.g., perceived fatigability, efficiency in neuromuscular control) may have also contributed to the changes in walking performance due to motor fatigue observed in the elderly or patient populations.

5. Conclusions

The current study investigated the effects of walking-induced motor fatigue on walking mechanics and energetics in healthy young adults. Our results demonstrate that motor fatigue impacts both walking economy and mechanical work efficiency. Motor fatigue also changes the mechanical work contribution of the lower-limb joints, showing a shift of positive work contribution from the ankle joint to the knee joint. Our results suggest that providing exoskeleton assistance at the ankle joint may help delay the onset of this lower-limb joint work redistribution observed during motor fatigue, which can potentially reduce the energy expenditure during a prolonged walk.

CRediT authorship contribution statement

Pei-Chun Kao: Conceptualization, Data curation, Formal analysis, Funding acquisition, Investigation, Methodology, Supervision, Visualization, Writing – original draft, Writing – review & editing. **Colin Lomasney:** Data curation, Formal analysis, Methodology, Project administration. **Yan Gu:** Funding acquisition, Methodology, Writing – review & editing. **Janelle P. Clark:** Methodology, Project administration, Writing – review & editing. **Holly A. Yanco:** Funding acquisition, Methodology, Supervision, Writing – review & editing.

Declaration of Competing Interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

Acknowledgements

Supported by NSF IIS-1955979. We thank Amie Russell for assisting with data collection.

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